Physics and Imaging in Radiation Oncology 16 (2020) 95–98

Contents lists available at ScienceDirect

Physics and Imaging in Radiation Oncology

journal homepage: www.sciencedirect.com/journal/physics-and-imaging-in-radiation-oncology

Technical Note

Influence of beam quality on reference dosimetry correction factors in magnetic resonance guided radiation therapy

Stefan Pojtinger[a,b,*], Marcel Nachbar[b], Ralf-Peter Kapsch[a], Daniela Thorwarth[b,1]

[a] *Physikalisch-Technische Bundesanstalt, Braunschweig, Germany*
[b] *Section for Biomedical Physics, Department of Radiation Oncology, University of Tübingen, Germany*

ARTICLE INFO

Keywords:
MR-linac
Dosimetry
Ionization chambers
Magnetic field correction factors
EGSnrc
Monte Carlo

ABSTRACT

Correction factors for reference dosimetry in magnetic resonance (MR) imaging-guided radiation therapy ($k_{\vec{B},M,Q}$) are often determined in setups that combine a conventional 6 MV linac with an electromagnet. This study investigated whether results based on these measurements were applicable for a 7 MV MR-linac using Monte Carlo simulations. For a Farmer-type ionization chamber, $k_{\vec{B},M,Q}$ was assessed for different tissue-phantom ratios ($TPR_{20,10}$). $k_{\vec{B},M,Q}$ differed by 0.0029(43) between $TPR_{20,10} = 0.6790(23)$(6 MV linac) and $TPR_{20,10} = 0.7028(14)$ (7 MV MR-linac) at 1.5T. The agreement was best in an orientation in which the secondary electrons were deflected to the stem of the ionization chamber.

1. Introduction

Magnetic resonance-guided radiation therapy (MRgRT) is a new modality in image guided radiotherapy where the target position is determined by magnetic resonance imaging (MRI) in real-time before and during the irradiation. Because of the good MRI soft tissue contrast, this leads to an improved patient positioning and therefore also potentially to improved tumor conformity [1–4].

On the other hand, the MRI scanner of the MR-linac introduces challenges for absolute dosimetry. This is because it is impossible to turn off the magnetic field in clinical practice, due to the high cost and time demands. However, ionization chambers, which are typically used for this kind of measurements are influenced by magnetic fields [5]. To compensate for this effect, an additional magnetic field correction factor must be applied to the reading of ionization chambers. Recently, several authors have published simulated as well as experimentally determined magnetic field correction factors for Farmer-type ionization chambers [5–14]. The magnetic field correction factor $k_{\vec{B},Q}$ can be defined as [13]:

$$k_{\vec{B},Q} = k_{\vec{B},M,Q}\, c_{\vec{B}} \qquad (1)$$

Here, $k_{\vec{B},M,Q}$ is the ratio between the reading of the ionization chamber in the presence of a magnetic field and the reading without a magnetic field. For the calculation of $k_{\vec{B},M,Q}$ the readings must be corrected for other influences, such as air pressure and temperature. The dose conversion factor $c_{\vec{B}}$ is calculated as the ratio of the absorbed dose-to-water at the point of measurement in the absence of a magnetic field to the absorbed dose-to-water at the point of measurement in the presence of a magnetic field.

One approach to determine $k_{\vec{B},M,Q}$ was proposed based on experiments that were carried out in setups that combine a conventional 6 MV linac with an electromagnet [5,7,14]. In such experimental setups, it is not possible to determine $k_{\vec{B},Q}$ directly. To measure $c_{\vec{B}}$ experimentally, a detector which is not influenced by the magnetic field would need to be used. On the other hand, the determination of $k_{\vec{B},M,Q}$ is possible in experimental setups that combine an electromagnet with a conventional linac [7,13]. In this case, for the calculation of $k_{\vec{B},Q}$, $k_{\vec{B},M,Q}$ must be multiplied with $c_{\vec{B}}$ in an additional step. To do this, $c_{\vec{B}}$ must be calculated by Monte Carlo simulations for the respective beam quality of the MR-linac, as done in [13]. The spread in the results for magnetic field correction factors that were determined by different methods is approximately 1% [15].

---

[*] Corresponding author at: Physikalisch-Technische Bundesanstalt, Braunschweig, Germany.
E-mail address: Stefan.Pojtinger@ptb.de (S. Pojtinger).

[1] Dr. Daniela Thorwarth, a co-author of this paper, is an Editor-in-Chief of Physics & Imaging in Radiation Oncology. The editorial process for this manuscript was managed independently from Dr. Thorwarth and the manuscript was subject to the Journal's usual peer-review process.

https://doi.org/10.1016/j.phro.2020.10.005
Received 15 June 2020; Received in revised form 7 October 2020; Accepted 7 October 2020

As the beam quality of conventional linacs differs from the beam quality of the two clinically available MR-linacs, that is 6 MV flattening filter free (FFF) for the 0.35 T MR-linac [16] and 7 MV FFF for the 1.5 T MR-linac [17], the determination $k_{\vec{B},M,Q}$ in a setup using a conventional linac is problematic. Due to the fact that the effects of the magnetic field can be explained by a change of the secondary electron's trajectories due to the Lorentz force [15], a change in the energy spectrum of these secondary particles might result in a change of $k_{\vec{B},M,Q}$.

A recent study investigated the effect of the beam quality on magnetic field correction factors and found significant differences for Farmer type ionization chambers [10]. However, the authors did not investigate the situation in which the ionization chamber is perpendicular to the magnetic field as well as the photon beam, which is the typical orientation for setups that combine an electromagnet with a conventional linac. In addition, this study was based on Monte Carlo simulations utilizing a simplified model of the dead volume inside the ionization chamber. The dead volume is defined as the part of the sensitive volume of the ionization chamber where charged particles are not collected by the electric field. Taking dead volumes into account has been shown to influence simulation results significantly [7].

Theoretically, experiments on the determination of magnetic field correction factors can also be performed at MR-linac sites after shutting down the magnetic field [12,13], but these experiments are very time-consuming and costly and can therefore usually not be carried out in clinical departments. In addition, the treatment beam is usually further tuned after the magnetic field has been turned on, which might impact the energy spectrum of the treatment beam. Therefore, it would be beneficial, if magnetic field correction factors that have been determined in a 6 MV setup were also valid for use also at 7 MV FFF MR-linacs.

The aim of this study was to investigate whether $k_{\vec{B},M,Q}$ for Farmer type ionization chambers differs significantly between a beam quality of 6 MV in a conventional linac and 7 MV FFF in a 1.5 T MR-linac.

2. Materials & methods

2.1. Calculation of $k_{\vec{B},M,Q}$

EGSnrc, a Monte Carlo code system [18] was used to simulate the energy deposition in the sensitive volume of a Farmer-type ionization chamber that was positioned inside a water tank. This was undertaken for both beam qualities (6 MV and 7 MV FFF) and for different magnetic flux densities.

To do this, a model of a PTW 30013 (PTW, Freiburg, Germany) Farmer-type ionization chamber was created in EGSnrc using the egs_chamber user code [19]. The dead volume inside the sensitive volume of the ionization chamber was modelled by a finite element simulation of the electric field inside the ionization chamber, as described in detail in our previous work [7].

Accelerator head models of a 6 MV linac (Elekta Precise, Elekta AB, Stockholm, Sweden) and the 7 MV FFF MR-linac (Unity, Elekta AB, Stockholm, Sweden) were created for the EGSnrc user code BEAMnrc [20,21]. Both accelerator head models were compiled as a shared library, so they could be used as a source in egs_chamber.

The models of the Farmer type ionization chamber as well as of the 6 MV linac were already used in a previous investigation [7]. In this study, Monte Carlo simulated results based on these models were successfully benchmarked with experimental results [7]. The accelerator head model of the 7 MV FFF MR-linac was described and benchmarked against MR-linac measurements [21]. All simulation parameters were set in accordance previous publications [6,7].

In the simulation geometry, the ionization chamber was placed at a 10cm depth inside a $30 \times 30 \times 20\mathrm{cm}^3$ water cuboid. The ionization chamber was placed in an orientation, in which the ionization chamber axis was perpendicular to the magnetic field vector. The beam direction was set perpendicular to the magnetic field vector, as well as to the ionization chamber axis (Fig. 1).

The magnitude of the magnetic flux density (B) was defined in negative values, if the secondary electrons were deflected to the tip of the ionization chamber, and this was defined in positive values, if the secondary electrons were deflected to the stem of the ionization chamber. Simulations were done for magnetic flux densities up to $B = 1.5\mathrm{T}$ in steps of 0.15T (Fig. 2).

The results were used to calculate $k_{\vec{B},M,Q}$ in dependence of the magnetic flux density:

$$k_{\vec{B},M,Q} = \frac{E_{\mathrm{dep}}}{E_{\mathrm{dep},\vec{B}}} \tag{2}$$

Here, E_{dep} is the deposited energy scored in the sensitive volume of the ionization chamber for $B = 0\mathrm{T}$ and $E_{\mathrm{dep},\vec{B}}$ is the deposited energy scored for a specific magnetic flux density $\vec{B}$, for the same irradiation conditions.

2.2. Calculation of the tissue-phantom-ratio

The beam quality of the implemented accelerator head models was quantified by calculating the tissue-phantom ratio (TPR). To do this, the deposited energy was scored in a cylindrical volume that was placed in the middle of the water cuboid at a 10cm depth ($E_{dep}^{\mathrm{cyl},10}$). The thickness and radius of the cylinder were set to 0.2mm and 1mm, respectively. A second simulation was carried out, in which the size of the water phantom was symmetrically increased to $30 \times 30 \times 40\mathrm{cm}^3$. For this, the position of the cylinder was not changed, so that the water cylinder was placed at a depth of 20cm. Again, the deposited energy was scored in the cylindrical volume ($E_{\mathrm{dep}}^{\mathrm{cyl},20}$).

For all simulations, the cylinder was placed at the isocentre of the corresponding accelerator. The distance between the source and the scoring volume was therefore set to 100cm for the 6 MV linac beam model and 143.5cm for the 7 MV FFF MR-linac beam model. The TPR at the depths of 20cm and 10cm ($TPR_{20,10}$) was calculated to quantify the beam quality for a field size of $10 \times 10\mathrm{cm}^2$ [22]:

$$TPR_{20,10} = \frac{E_{\mathrm{dep}}^{\mathrm{cyl},20}}{E_{\mathrm{dep}}^{\mathrm{cyl},10}} \tag{3}$$

For the simulation of $TPR_{20,10}$, the magnetic flux density was set to $B = 1.5\mathrm{T}$.

The uncertainties given in this work were calculated by multiplying the standard deviation of the Monte Carlo results by a factor of 2.

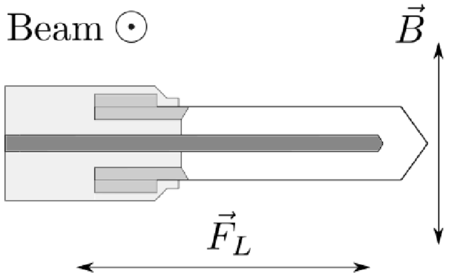

Fig. 1. Schematic representation of the orientation of the magnetic field vector (B→), the radiation beam direction, the Lorentz force (F L) and the ionization chamber axis.

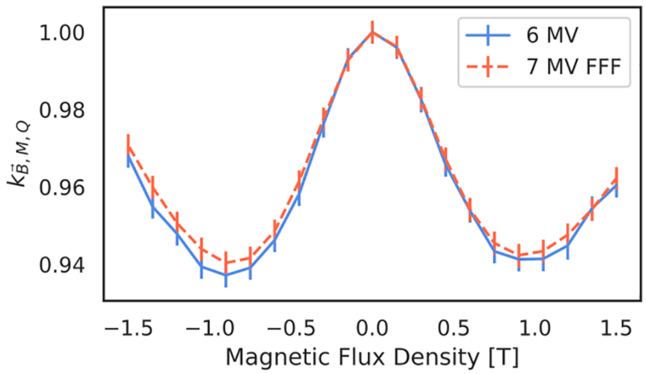

Fig. 2. Simulation results for *kB*→, Q for the Farmer-type ionization chamber for both beam qualities.

3. Results

The maximum difference for the different beam qualities was found for $B = -1.35$T. For this magnetic flux density, the effect of the magnetic field was more prominent for the 6 MV beam quality and differed by 0.0055(44) from the 7 MV FFF beam quality. Also, for $B = -1.05$T, the results differed by 0.0052(44). For all other magnetic flux densities, the differences were within the uncertainty intervals, that never exceeded 0.0047. For $B = -1.5$T and $B = 1.5$T the results differed by 0.0020(43) and 0.0029(43). The calculation of $TPR_{20,10}$ resulted in $TPR_{20,10} = 0.7028(14)$ for the 7 MV MR-linac and $TPR_{20,10} = 0.6790(23)$ for the 6 MV linac.

4. Discussion

In this work, we investigated the influence of the beam quality on $k_{\overrightarrow{B},M,Q}$ for the Farmer ionization chamber PTW 30013. In addition, the beam quality was quantified in terms of $TPR_{20,10}$. It was found that there is only a minor effect of the change of beam quality on $k_{\overrightarrow{B},M,Q}$.

In our simulations, the change of the beam quality did not influence $k_{\overrightarrow{B},M,Q}$ more than 0.5%. The agreement was best, when the ionization chamber was simulated in an orientation in which the secondary electrons were deflected to the stem of the ionization chamber. It can be speculated, that the effect of the change of beam quality on the secondary electron spectrum inside the ionization chamber is small, for a change of beam quality between 6 MV and 7 MV FFF.

These results can be compared to the experimental results of Gohil et al. [14]. The authors found that $k_{\overrightarrow{B},M,Q}$ differed by 0.51%, comparing the 6 MV beam quality to the 8 MV beam quality. Gohil et al. [14] did not report any beam quality specifier. Therefore, it cannot be concluded how close the 6 MV and 8 MV beam quality of the conventional linac used by Gohil et al. [14] is to the 6 MV beam quality that was used for the Monte Carlo simulations in this work. The same applies the 7 MV FFF beam quality of the 1.5 T MR-linac. However, it can be speculated, that the change in the photon spectrum is more dominant for a change of beam quality from 6 MV to 8 MV (conventional linac), compared to 6 MV (conventional linac) to 7 MV FFF (MR-linac). In this sense, the study of Gohil et al. [14] supports our conclusion, that the change of the beam quality between a conventional 6 MV linac and a 7 MV MR-linac does not influence $k_{\overrightarrow{B},M,Q}$ by more than approximately 0.5%.

For the 7 MV FFF beam quality, the simulated beam quality specifier $TPR_{20,10}$ is in excellent agreement with experimental results that can be found in the literature [12,23]. De Prez *et al.* [12] found $TPR_{20,10} = 0.701(2)$ and Snyder *et al.* [23] published a value of $TPR_{20,10} = 0.704$. In contrast, the results of this work do not agree with the Monte Carlo simulated values of other authors [9,10]. These authors published $TPR_{20,10} = 0.695$ and $TPR_{20,10} = 0.691(1)$, respectively.

The limited space between the pole shoes of an electromagnet might result in an additional problem as this limits the maximum field size and the size of the phantom. A recent publication showed that a reduction of the field size to $3 \times 10\text{cm}^2$ can influence the results for $k_{\overrightarrow{B},Q}$ by 0.4%[8].

In conclusion, this study revealed that the change of the beam quality between $TPR_{20,10} = 0.6790(23)$ (6 MV linac) and $TPR_{20,10} = 0.7028(14)$ (7 MV MR-linac) did not influence $k_{\overrightarrow{B},M,Q}$ more than 0.5%, for the PTW 30013 ionization chamber in perpendicular orientation.

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

Acknowledgements

The Tübingen MR-linac programme is funded by the German Research Council (DFG), grant no. ZI 736/2-1. Marcel Nachbar and Daniela Thorwarth report institutional collaborations with Elekta, Philips and PTW Freiburg. We acknowledge the support provided by the state of Baden-Württemberg through bwHPC and by the German Research Foundation (DFG) through Grant No. INST 39/963-1 FUGG. This work is part of the project Traceable Dosimetry for Small fields in MR-guided Radio Therapy (MRgRT-DOS) that has received funding from the EMPIR programme co-financed by the participating states and from the European Union's Horizon 2020 research and innovation programme.

This paper is part of a special issue that contains contributions originally submitted to the scientific meeting MR in RT, which was planned to take place 05/2020, organized by the German Research Center (DKFZ) in Heidelberg. We acknowledge funding by DKFZ for the publication costs of this special issue.

References

[1] Nachbar M, Mönnich D, Boeke S, Gani C, Weidner N, Heinrich V, et al. Partial breast irradiation with the 1.5 T MR-Linac: first patient treatment and analysis of electron return and stream effects. Radiother Oncol 2020;145:30–5. https://doi.org/10.1016/j.radonc.2019.11.025.

[2] Nachbar M, Mönnich D, Kalwa P, Zips D, Thorwarth D, Gani C. Comparison of treatment plans for a high-field MRI-linac and a conventional linac for esophageal cancer. Strahlenther Onkol 2019;195:327–34. https://doi.org/10.1007/s00066-018-1386-z.

[3] Finazzi T, van Sörnsen de Koste JR, Palacios MA, Spoelstra FOB, Slotman BJ, Haasbeek CJA, et al. Delivery of magnetic resonance-guided single-fraction stereotactic lung radiotherapy. Phys Imag Radiat Oncol 2020;14:17–23. https://doi.org/10.1016/j.phro.2020.05.002.

[4] Winkel D, Bol GH, Kroon PS, van Asselen B, Hackett SS, Werensteijn-Honingh AM, et al. Adaptive radiotherapy: the Elekta Unity MR-linac concept. Clin Transl Radiat Oncol 2019;18:54–9. https://doi.org/10.1016/j.ctro.2019.04.001.

[5] Meijsing I, Raaymakers BW, Raaijmakers AJE, Kok JGM, Hogeweg L, Liu B, et al. Dosimetry for the MRI accelerator: the impact of a magnetic field on the response of a Farmer NE2571 ionization chamber. Phys Med Biol 2009;54:2993–3002. https://doi.org/10.1088/0031-9155/54/10/002.

[6] Pojtinger S, Dohm OS, Kapsch R-P, Thorwarth D. Ionization chamber correction factors for MR-linacs. Phys Med Biol 2018;63:11NT03. https://doi.org/10.1088/1361-6560/aac4f2.

[7] Pojtinger S, Kapsch R-P, Dohm OS, Thorwarth D. A finite element method for the determination of the relative response of ionization chambers in MR-linacs: simulation and experimental validation up to 1.5 T. Phys Med Biol Biol 2019;64:135011. https://doi.org/10.1088/1361-6560/ab2837.

[8] Spindeldreier CK, Schrenk O, Bakenecker A, Kawrakow I, Burigo L, Karger CP, et al. Radiation dosimetry in magnetic fields with Farmer-type ionization chambers: determination of magnetic field correction factors for different magnetic field strengths and field orientations. Phys Med Biol 2017;62:6708–28. https://doi.org/10.1088/1361-6560/aa7ae4.

[9] O'Brien DJ, Roberts DA, Ibbott GS, Sawakuchi GO. Reference dosimetry in magnetic fields: formalism and ionization chamber correction factors. Med Phys 2016;43:4915–27. https://doi.org/10.1118/1.4959785.

[10] Malkov VN, Rogers DWO. Monte Carlo study of ionization chamber magnetic field correction factors as a function of angle and beam quality. Med Phys 2018;45:908–25. https://doi.org/10.1002/mp.12716.

[11] Smit K, van Asselen B, Kok JGM, Aalbers AHL, Lagendijk JJW, Raaymakers BW. Towards reference dosimetry for the MR-linac: magnetic field correction of the ionization chamber reading. Phys Med Biol 2013;58:5945–57. https://doi.org/10.1088/0031-9155/58/17/5945.

[12] de Prez L, Woodings S, de Pooter J, van Asselen B, Wolthaus J, Jansen B, et al. Direct measurement of ion chamber correction factors, k Q and k B, in a 7 MV MRI-linac. Phys Med Biol 2019;64:105025. https://doi.org/10.1088/1361-6560/ab1511.

[13] van Asselen B, Woodings SJ, Hackett SL, Van Soest TL, Kok JGM, Raaymakers BW, et al. A formalism for reference dosimetry in photon beams in the presence of a magnetic field. Phys Med Biol 2018;63:125008. https://doi.org/10.1088/1361-6560/aac70e.

[14] Gohil P, Agnew J, Berresford J, Budgell G, Billas I, Duane S. Measurement of magnetic field correction factors for multiple radiation detectors for MR linacs (abstr). Radiother Oncol 2018;127:S468–9. https://doi.org/10.1016/s0167-8140(18)31195-2.

[15] de Pooter JA, Billas I, de Prez LA, Duane S, Kapsch R-P, Karger C, et al. Reference dosimetry in MRI-linacs: evaluation of available protocols and data to establish a code of practice. Phys Med Biol 2020. https://doi.org/10.1088/1361-6560/ab9efe. In press.

[16] Klüter S. Technical design and concept of a 0.35 T MR-Linac. Clin Transl Radiat Oncol 2019;18:98–101. https://doi.org/10.1016/j.ctro.2019.04.007.

[17] Raaymakers BW, Jürgenliemk-Schulz IM, Bol GH, Glitzner M, Kotte ANTJ, van Asselen B, et al. First patients treated with a 1.5 T MRI-Linac: clinical proof of concept of a high-precision, high-field MRI guided radiotherapy treatment. Phys Med Biol 2017;62:L41–50. https://doi.org/10.1088/1361-6560/aa9517.

[18] Kawrakow I, Mainegra-Hing E, Rogers DWO, Tessier F, Walters BRB. The EGSnrc Code System: Monte Carlo Simulation of Electron and Photon Transport. NRCC PIRS-701 2019.

[19] Wulff J, Zink K, Kawrakow I. Efficiency improvements for ion chamber calculations in high energy photon beams. Med Phys 2008;35:1328–36. https://doi.org/10.1118/1.2874554.

[20] Rogers DWO, Walters B, Kawrakow I. BEAMnrc Users Manual. NRCC PIRS-0509(A) RevL 2018.

[21] Friedel M, Nachbar M, Mönnich D, Dohm O, Thorwarth D. Development and validation of a 1.5 T MR-Linac full accelerator head and cryostat model for Monte Carlo dose simulations. Med Phys 2019;46:5304–13. https://doi.org/10.1002/mp.13829.

[22] Almond PR, Biggs PJ, Coursey BM, Hanson WF, Huq MS, Nath R, et al. AAPM's TG-51 protocol for clinical reference dosimetry of high-energy photon and electron beams. Med Phys 1999;26:1847–70. https://doi.org/10.1118/1.598691.

[23] Snyder JE, St-Aubin J, Yaddanapudi S, Boczkowski A, Dunkerley DAP, Graves SA, et al. Commissioning of a 1.5T Elekta Unity MR-linac: a single institution experience. J Appl Clin Med Phys 2020;21:160–72. https://doi.org/10.1002/acm2.12902.