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RESEARCH ARTICLE

MEDICAL PHYSICS

# Experimental determination of magnetic field quality conversion factors for eleven ionization chambers in 1.5 T and 0.35 T MR-linac systems

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## Abstract

**Background:** The static magnetic field present in magnetic resonance (MR)-guided radiotherapy systems can influence dose deposition and charged particle collection in air-filled ionization chambers. Thus, accurately quantifying the effect of the magnetic field on ionization chamber response is critical for output calibration. Formalisms for reference dosimetry in a magnetic field have been proposed, whereby a magnetic field quality conversion factor $k_{B,Q}$ is defined to account for the combined effects of the magnetic field on the radiation detector. Determination of $k_{B,Q}$ in the literature has focused on Monte Carlo simulation studies, with experimental validation limited to only a few ionization chamber models.

**Purpose:** The purpose of this study is to experimentally measure $k_{B,Q}$ for 11 ionization chamber models in two commercially available MR-guided radiotherapy systems: Elekta Unity and ViewRay MRIdian.

**Methods:** Eleven ionization chamber models were characterized in this study: Exradin A12, A12S, A28, and A26, PTW T31010, T31021, and T31022, and IBA FC23-C, CC25, CC13, and CC08. The experimental method to measure $k_{B,Q}$ utilized cross-calibration against a reference Exradin A1SL chamber. Absorbed dose to water was measured for the reference A1SL chamber positioned parallel to the magnetic field with its centroid placed at the machine isocenter at a depth of 10 cm in water for a 10 × 10 cm$^2$ field size at that depth. Output was subsequently measured with the test chamber at the same point of measurement. $k_{B,Q}$ for the test chamber was computed as the ratio of reference dose to test chamber output, with this procedure repeated for each chamber in each MR-guided radiotherapy system. For the high-field 1.5 T Elekta Unity system, the dependence of $k_{B,Q}$ on the chamber orientation relative to the magnetic field was quantified by rotating the chamber about the machine isocenter.

**Results:** Measured $k_{B,Q}$ values for our test dataset of ionization chamber models ranged from 0.991 to 1.002, and 0.995 to 1.004 for the Elekta Unity and ViewRay MRIdian, respectively, with $k_{B,Q}$ tending to increase as the chamber sensitive volume increased. Measured $k_{B,Q}$ values largely agreed within uncertainty to published Monte Carlo simulation data and available experimental data. $k_{B,Q}$ deviation from unity was minimized for ionization chamber orientation parallel or antiparallel to the magnetic field, with increased deviations observed

at perpendicular orientations. Overall ($k$ = 1) uncertainty in the experimental determination of the magnetic field quality conversion factor, $k_{B,Q}$ was 0.71% and 0.72% for the Elekta Unity and ViewRay MRIdian systems, respectively.
**Conclusions:** For a high-field MR-linac, the characterization of ionization chamber performance as angular orientation varied relative to the magnetic field confirmed that the ideal orientation for output calibration is parallel. For most of these chamber models, this study represents the first experimental characterization of chamber performance in clinical MR-linac beams. This is a critical step toward accurate output calibration for MR-guided radiotherapy systems and the measured $k_{B,Q}$ values will be an important reference data source for forthcoming MR-linac reference dosimetry protocols.

**KEYWORDS**
ionization chamber, $k_{B,Q}$, magnetic field correction factor, magnetic field quality conversion factor, MR-guided radiotherapy, MR-linac, reference dosimetry

## 1 | INTRODUCTION

Magnetic resonance (MR)-guided radiotherapy systems combine the high soft tissue contrast of MR imaging with integrated linear accelerator beam delivery, allowing for accurate dose delivery, organ-at-risk sparing, and the potential for online adaptive radiotherapy. Current widely-used commercially-available MR-guided radiotherapy systems, or MR-linear accelerators (MR-linacs) include the Elekta Unity[1] (Elekta Solutions AB, Stockholm, Sweden) and ViewRay MRIdian[2] (ViewRay Technologies Inc., Oakwood, OH, USA). Other systems entering the market such as the MagnetTx Aurora-RT[3] (MagnetTx Oncology Solutions Ltd., Edmonton, AB, Canada) will not be discussed in this manuscript but may become important as clinical availability and use increases.

In the presence of strong static magnetic fields utilized in these MR-guided radiotherapy systems, electrons set in motion by the linear accelerator photon beam in the medium have a curved path influenced by the Lorentz force. The magnetic field influence on charged particle paths can thus impact dose deposition.[4] Furthermore, the magnitude of curvature in the charged particle path is increased in low density material such air in comparison to a water-equivalent medium. The altered pathlength of electrons in the air-filled sensitive volumes of ionization chambers when in the presence of a magnetic field can influence charge collection and thus dose measurement.[5] It is critical for output calibration of MR-linac systems that the effect of the magnetic field on charge collection is accurately quantified and reference dosimetry formalisms are extended to account for the presence of the magnetic field.

Several groups have proposed reference dosimetry frameworks for MR-guided radiotherapy,[6–10] most commonly as an extension of the accepted AAPM TG-51 protocol and its addendum for clinical reference dosimetry of high-energy photon beams.[11,12] The working formalism for reference dosimetry of high-energy photon beams in the presence of a strong magnetic field can be written as follows[8,10]:

$$D_W^{B,Q} = M k_Q k_{B,Q} N_{D,W}^{Co60}, \tag{1}$$

where $D_W^{B,Q}$ is the absorbed dose to water at the point of measurement for a beam quality $Q$ in the presence of a permanent magnetic field $B$; $k_Q$ is the beam quality conversion factor which converts the detector calibration coefficient for a reference Cobalt-60 beam quality to one for beam quality $Q$ in the absence of the magnetic field; $N_{D,W}^{Co60}$ is the detector calibration coefficient for a Cobalt-60 reference beam; and $M$ is the corrected charge reading, defined as:

$$M = M_{raw} P_{TP} P_{ion} P_{pol} P_{elec} P_{leak} P_{rp}, \tag{2}$$

with no changes from the TG-51 addendum definitions.[12] Correction factors include the temperature-pressure correction ($P_{TP}$), ion recombination correction ($P_{ion}$), polarity correction ($P_{pol}$), electrometer correction ($P_{elec}$), leakage correction ($P_{leak}$), and radial profile correction ($P_{rp}$). The new term in this formalism, $k_{B,Q}$, is known as the magnetic field quality conversion factor which is defined for beam quality $Q$ in the presence of a permanent magnetic field $B$. This factor corrects for ionization chamber response changes due to the presence of a magnetic field and depends on the strength of the permanent magnetic field, the beam quality, the chamber type, and the orientation of the ionization chamber relative to the radiation beam and the magnetic field.[8]

There has been extensive work characterizing the magnetic field quality conversion factor, $k_{B,Q}$, for commonly used air-filled ionization chambers using Monte Carlo simulation.[7,8,13–15] O'Brien et al. simulated ionization chamber response in MR-linac for Exradin A19, NE2571, and several PTW chambers, with further experimental validation in a preclinical 1.5 T MR-linac.[7] Malkov and Rogers determined magnetic field quality conversion factors for 32 cylindrical ionization chambers as a function of angle for both a Cobalt-60 beam with a 0.35 T magnetic field and a 7 MV beam with a 1.5 T magnetic field.[8] Pojtinger et al. utilized a finite element method in combination with Monte Carlo simulation to determine PTW 30013 Farmer ionization chamber response in MR-linacs for magnetic field strengths from 0 to 1.5 T.[13] Cervantes et al. simulated ionization chamber response for small-cavity chambers (PTW31010, PTW31021, and PTW31022) in a 7 MV FFF beam and 1.5 T magnetic field.[14] Finally, Margaroni et al. calculated the magnetic field quality conversion factors for 12 ionization chambers at varying angle relative to the magnetic field direction through Monte Carlo simulation of the 1.5T Elekta Unity MR-linac.[15]

At present, experimental validation is limited to only a few chamber models. van Asselen et al. and de Prez et al. experimentally measured the response of PTW 310013 and IBA FC65-G ionization chambers in a pre-clinical 1.5 T MR-linac, with de Prez et al. utilizing a custom-built water calorimeter as a primary standard to measure an absorbed dose.[9,16] D'Souza et al. designed and built an MR-compatible water calorimeter to directly measure the absorbed dose and determine the magnetic field quality conversion factor for an Exradin A1SL ionization chamber in the Elekta Unity MR-linac.[17] In subsequent work, Iakovenko et al. evaluated magnetic field quality conversion factors as a function of angle relative to the magnetic field in an Elekta Unity MR-linac for Exradin A19 and A1SL as well as IBA FC65-G and CC13 ionization chambers using a cross-calibration technique.[18] For the 0.35 T ViewRay MRIdian MR-linac, Krauss et al. utilized water calorimetry to directly measure absorbed dose and determine magnetic field quality conversion factors for eight cylindrical ionization chambers including IBA FC65-G, PTW 31010, and PTW31021.[19]

In this paper, we experimentally measure the magnetic field quality conversion factor, $k_{B,Q}$, for eleven commercially available cylindrical ionization chambers in both the Elekta Unity and ViewRay MRIdian MR-guided radiotherapy systems. In addition, we characterize ionization chamber response for different orientations relative to the static 1.5 T magnetic field of the Elekta Unity system and provide a comprehensive uncertainty budget analysis.[20] This experimental characterization is critical for validating Monte Carlo simulation studies and will be important as a reference data source for forthcoming MR-linac reference dosimetry protocols.

## 2 | MATERIALS AND METHODS

Experiments were conducted using two clinical MR-linac systems, the Elekta Unity and ViewRay MRIdian, representing the two commercially available and most commonly used MR-guided radiotherapy systems. The Elekta Unity system features a 1.5 T static magnetic field and single-energy 7 MV flattening-filter-free (FFF) linear accelerator beam,[1] while the ViewRay MRIdian system utilizes a 0.35 T static magnetic field strength and six MV FFF beam.[2] In both systems, the radiation beam is oriented perpendicular to the static magnetic field.

Eleven ionization chamber models lacking experimental $k_{B,Q}$ validation were characterized for this study. The 11 chambers were selected based on three main criteria that were deemed necessary to make them "appropriate" for use in MR-linac: (1) chambers were reference class and suitable for reference dosimetry; (2) chambers were previously characterized with published $k_Q$ values; and (3) chambers were fully waterproof so that direct in water reference dosimetry measurements were possible. The ionization chamber model, sensitive cavity volume and length, distance from the tip of the chamber to the centroid of the cavity, and corresponding $k_Q$ values are provided in Table 1. Ten commercially available chamber models were loaned from ionization chamber manufacturers Standard Imaging (Standard Imaging Inc., Middleton, WI, USA), IBA (IBA Dosimetry GmbH, Schwarzenbruck, Germany), and PTW (PTW Freiburg GmbH, Freiburg, Germany) for the purposes of this study. The detector calibration coefficient, $N_{D,W}^{Co60}$, for each of these ionization chambers was determined in a Cobalt-60 reference radiation beam at the University of Wisconsin Accredited Dosimetry Calibration Laboratory (ADCL). For the final chamber model, IBA CC13, $k_{B,Q}$ has been previously experimentally determined for the Elekta Unity,[18] so measurements for this study were only completed using the ViewRay MRIdian system. The CC13 chamber used in this study was previously calibrated and is currently in clinical use at the University of Wisconsin-Madison.

The ratio of tissue phantom ratios (TPR) at depths of 20 and 10 cm in water with a field size of 10 × 10 cm$^2$ for the same detector position, known as $TPR_{10}^{20}$, was determined for each MR-guided radiotherapy system as a beam quality specifier. Using $TPR_{10}^{20}$, the beam quality conversion factor, $k_Q$, for each ionization chamber apart from the IBA CC08 chamber was computed using the method described in Andreo et al.[21,22] This publication combines both experimental and Monte Carlo data for a number of ionization chambers to provide fit parameters to calculate their consensus beam quality conversion factors as a function of $TPR_{10}^{20}$.[21] Calculation of $k_Q$ for the PTW T31022 chamber was described in a PTW internal report, which also utilized the method described by Andreo et al.[21,22] Due to its exclusion from

**TABLE 1** List of ionization chamber models used in this study, including the chamber cavity sensitive volume and length, as well as the distance from the tip of the chamber to the centroid of the cavity. The $k_Q$ values used and their sources are provided for the Elekta Unity (7 MV FFF) and ViewRay MRIdian (6 MV FFF) MR-guided radiotherapy systems calculated using $TPR^{20}_{10}$ values of 0.699 and 0.643, respectively.

| Ion chamber model | Cavity volume ($cm^3$) | Cavity length (mm) | Centroid of cavity from tip (mm) | Elekta unity $k_Q$ | ViewRay MRIdian $k_Q$ |
|---|---|---|---|---|---|
| Exradin A1SL | 0.053 | 6.0 | 4.1 | 0.9874[a] | 0.9943[a] |
| Exradin A12 | 0.64 | 24.8 | 12.9 | 0.9888[a] | 0.9954[a] |
| Exradin A12S | 0.24 | 10.6 | 5.8 | 0.9886[a] | 0.9953[a] |
| Exradin A28 | 0.125 | 8.3 | 4.5 | 0.9888[a] | 0.9953[a] |
| Exradin A26 | 0.015 | 2.4 | 1.8 | 0.9893[a] | 0.9957[a] |
| PTW T31010 | 0.125 | 6.5 | 4.5 | 0.9836[a] | 0.9923[a] |
| PTW T31021 | 0.07 | 4.8 | 3.5 | 0.9858[a] | 0.9932[a] |
| PTW T31022 | 0.016 | 2.9 | 2.4 | 0.9877[b] | 0.9947[b] |
| IBA FC23-C | 0.23 | 9.0 | 4.8 | 0.9875[a] | 0.9956[a] |
| IBA CC25 | 0.25 | 10.0 | 5.4 | 0.9891[a] | 0.9956[a] |
| IBA CC13 | 0.13 | 5.8 | 3.5 | 0.9877[a] | 0.9949[a] |
| IBA CC08 | 0.08 | 4.0 | 2.4 | 0.9865[c] | 0.9928[c] |

Calculated using.
[(a)]Andreo et al. (2020).[21] (b) Würfel et al. (2022).[22] (c) Muir and Rogers (2010).[23]

the Andreo report, the beam quality conversion factor, $k_Q$, for the IBA CC08 ionization chamber was calculated using the Monte Carlo-based formula provided by Muir and Rogers, also as a function of $TPR^{20}_{10}$.[23] When using $TPR^{20}_{10}$ as a beam quality specifier, the $k_Q$ values calculated by Muir and Rogers agree to within 0.2% of $k_Q$ values reported by Andreo et al.[21,23] Table 1 provides $k_Q$ values for each ionization chamber examined in this study calculated using $TPR^{20}_{10}$ values of 0.699 and 0.643 for the Elekta Unity and ViewRay MRIdian, respectively.

## 2.1 Experimental kB,Q determination

This study describes the experimental measurement of magnetic field quality conversion factors, $k_{B,Q}$, through cross calibration of test ionization chambers against a reference Exradin A1SL ionization chamber. The Exradin A1SL chamber was chosen as a reference due to its well-validated $k_{B,Q}$ data, including both Monte Carlo[8] and experimental data.[17,18] While we did not use a primary standard such as calorimetry to directly measure absorbed dose to water, the Exradin A1SL experimental data can be traced back to water calorimetry work that was performed under the same Elekta Unity MR-linac that is used in this work.[17] An average of the available $k_{B,Q}$ data, weighted based on uncertainty, was computed to provide reference $k_{B,Q}$ values for the A1SL chamber of 0.9973 for the Elekta Unity system[8,18] and 0.9985 for the ViewRay MRIdian system,[8,19] applicable to chamber orientation parallel to the magnetic field and perpendicular to the radiation beam.

First, the reference A1SL ionization chamber was positioned with its point of measurement, the centroid of the sensitive collecting volume, at the machine isocenter at a depth of 10 cm in water. Chamber alignment procedures are specific to the MR-guided radiotherapy system and will be described in further detail in Sections 2.2 and 2.3. All measurements utilized a 10 × 10 $cm^2$ field size, defined at the machine isocenter, and a gantry angle of 0°.

Based on both Monte Carlo simulations and experimental work, the magnetic field quality conversion factor's difference from unity is minimized when the ionization chamber is positioned parallel to the magnetic field and perpendicular to the radiation beam.[8,18] This corresponds to the so-called 0° orientation as shown in the beams-eye-view photograph of our experimental setup (Figure 1), where the tip of the ionization chamber is pointing in the direction of the magnetic field.

Measurements were first obtained with the reference A1SL positioned in parallel orientation at the machine isocenter to determine the absorbed dose to water using the formula shown in Equation 1. This measurement of dose at the chamber point of measurement in the parallel orientation served as the reference for test chamber output measurements at all angles with respect to the magnetic field. For all measurements on both the Elekta Unity and ViewRay MRIdian systems, a PTW Model T10010 UNIDOS E (PTW Freiburg GmbH, Freiburg, Germany) electrometer was used with a standard –300 V polarization voltage and positive collecting electrode. Fixed measurement protocols were employed across both MR-linacs. Before each measurement series and after each polarization voltage change, between 2000 MU and 3000 MU were delivered to stabilize the ionization chamber and electrometer, as well as to ensure a leakage current that measured less than 0.1% of

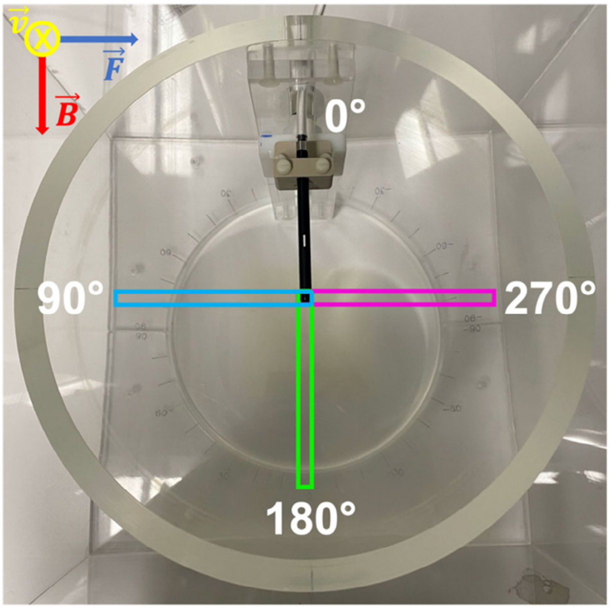


**FIGURE 1** Photograph of our experimental setup demonstrating A1SL ionization chamber orientation. The magnetic field is directed inferiorly toward the bottom of the treatment couch and the x-ray beam is directed vertically into the treatment couch, with the Lorentz force directed laterally toward patient left. 0° orientation refers to the chamber being positioned parallel with the magnetic field pointing out of the bore toward the bottom of the couch. The cylindrical insert allows rotation of the chamber to positions at 90° (blue), 180° (green), and 270° (magenta).

the average measured charge. A minimum of three measurements were obtained at the standard −300 V polarization voltage, and measurements were repeated until the coefficient of variation was less than 0.1%. 400 MU per measurement was delivered for small-volume ionization chambers with less than 0.1 cm$^3$ sensitive volume, such as the A1SL chamber, and 200 MU per measurement was delivered for ionization chambers with larger than 0.1 cm$^3$ sensitive volume. Measurements were repeated at polarization voltages of −150 V and +300 V to compute $P_{ion}$ and $P_{pol}$ correction factors, respectively, following the method described in AAPM's TG-51 protocol.[11] Temperature and pressure were continuously monitored to compute $P_{TP}$ correction factors at different time points, using the formula provided in AAPM's TG-51 protocol.[11] Finally, due to both MR-linacs using flattening filter-free (FFF) beams, the radial profile correction factor, $P_{rp}$, introduced in the TG-51 addendum[12] was computed for each ionization chamber and each MR-linac system. To compute $P_{rp}$, in-line beam profiles taken along the long axis of the ionization chamber were averaged over the chamber-specific sensitive volume length provided in Table 1.

Following measurements with the reference A1SL chamber, the new test chamber was placed with its point of measurement at the same position at the machine's isocenter. The procedure described in the previous paragraph was repeated to measure the output for the test chamber as well as to compute $P_{TP}$, $P_{ion}$, $P_{pol}$, $P_{leak}$, and $P_{rp}$ correction factors specific to the test chamber. By taking the ratio of the absorbed dose to water as measured using the reference A1SL ionization chamber and the product of the corrected measured charge, beam quality conversion factor, and Cobalt-60 detector calibration coefficient for the new test chamber, as shown in Equation 3,

$$k_{B,Q}^{test} = \frac{\left(D_W^{B,Q}\right)^{A1SL}}{\left(Mk_Q N_{D,W}^{Co60}\right)^{test}} = \frac{\left(Mk_Q k_{B,Q} N_{D,W}^{Co60}\right)^{A1SL}}{\left(Mk_Q N_{D,W}^{Co60}\right)^{test}}, \tag{3}$$

the magnetic field quality conversion factor $k_{B,Q}^{test}$ for the test ionization chamber could be determined. This procedure was repeated for each ionization chamber in each MR-linac system. Reference A1SL output measurements were repeated at the start and end of each measurement session to assess MR-linac output stability for uncertainty analysis.

Uncertainty budget analysis for experimental $k_{B,Q}$ determination using the cross-calibration method described above for the Elekta Unity and ViewRay MRIdian systems was conducted. A breakdown of uncertainty components as well as overall ($k = 1$) uncertainty for each MR-linac system is provided in Table 2.

## 2.2 | Elekta Unity experimental setup

For the Elekta Unity measurements, a custom-built 28 × 28 × 35 cm$^3$ MR-compatible water tank was used, as described in Iakovenko et al.[18] Ionization chambers were held in place using custom chamber holders fixed to a cylindrical insert (radius = 25 cm) within the water tank as shown in Figure 1. The chamber holder is mounted on a stage capable of moving vertically to help align the chamber at the machine isocenter.

Utilization of the calibrated onboard MV electronic portal imaging device (EPID) allowed for positioning of the centroid of the ionization chamber sensitive volume at the machine isocenter and cylindrical insert centroid with sub-millimeter accuracy. EPID images acquired at a gantry angle of 0° allowed for precise lateral and longitudinal alignment with the machine isocenter, as shown in Figure 2a. The chamber axis was aligned rotationally by matching the etched markings on the rotating cylindrical insert with the etched markings on the base of the phantom. This alignment was confirmed with the EPID images obtained at a gantry angle of 0°, allowing for an overall rotational alignment accuracy of better than 0.5°. By acquiring EPID images at a gantry angle of 90°, the chamber point of measurement could be aligned with the machine isocenter in the vertical plane. The water depth was assessed by placing a ruler along the base of the phantom for a consistent point of reference.

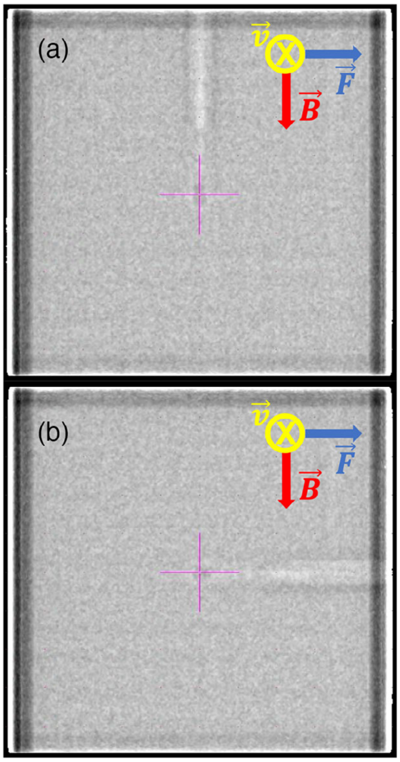


**FIGURE 2** Example MV EPID images of the A1SL ionization chamber acquired at 10 cm depth in water with 10 × 10 $cm^2$ field size and 0° gantry angle for (a) 0° orientation and (b) 270° orientation relative to the static magnetic field, showing alignment of the centroid of the chamber sensitive volume and machine isocenter, represented by the crosshair.

## 2.3 | ViewRay MRIdian experimental setup

For the ViewRay MRIdian measurements, a CNMC WP-3040 water tank (CNMC Company Inc., Nashville, TN, USA) was used (30 × 40 × 38 $cm^3$) with the tank's universal chamber holder used to position each ionization chamber. Each chamber was aligned to the lasers outside the bore of the machine, and shifted longitudinally, as needed, to align the centroid of the measurement volume with wall-mounted lasers. The correspondence between the laser position and imaging/beam isocenter is verified daily. If the chamber or cap had an external marking of the active chamber volume, that mark was used to set the longitudinal position. The tank was then translated by moving the couch a known displacement of 155 cm longitudinally into the bore for placement at the machine isocenter.

## 2.4 | Angular dependence of $k_{B,Q}$

Orientation of the ionization chamber relative to the magnetic field direction has been shown to impact the magnetic field quality conversion factor and thus ionization chamber response in high magnetic field MR-guided radiotherapy systems.[18] To account for potential variations in setup technique at different centers, a characterization of the angular dependence of $k_{B,Q}$ was completed. The magnetic field quality conversion factor, $k_{B,Q}$, was characterized for the ionization chamber axis positioned at each cardinal angle (0°, 90°, 180°, and 270°) relative to the static magnetic field. Figure 1 demonstrates the angular orientation labeling system in a beams-eye-view photograph of our experimental setup, showing the outline of the ionization chamber when positioned at each cardinal angle. The lower 0.35T magnetic field strength of the MRIdian MR-guided radiotherapy system has been shown to reduce the magnitude of angular dependence on $k_{B,Q}$ compared to the high-field Elekta Unity system.[8,13,24] As a result, angular dependence on $k_{B,Q}$ values was only quantified for the Elekta Unity.

As described in Section 2.2, the custom MR-compatible water tank used for Elekta Unity measurements included a cylindrical insert. This insert, shown in Figure 1, allowed for rotation of the ionization chamber about the cylinder centroid and thus machine isocenter. MV EPID imaging was used to align the centroid of the chamber-sensitive volume with the machine isocenter and the centroid of the rotating cylindrical insert. Using the etched angle markings on the cylindrical insert and base of the water phantom, the ionization chamber could be rotated about the machine isocenter to positions at each cardinal angle. MV EPID images were acquired after each rotation to confirm alignment of the chamber point of measurement with the machine isocenter. An example EPID image with the ionization chamber at the 270° orientation is shown in Figure 2b.

With the ionization chamber at each cardinal angle orientation, the measurement procedure as described in Section 2.1 was repeated to determine the magnetic field quality conversion factor, $k_{B,Q}$, specific to that orientation. $P_{pol}$ and $P_{ion}$ correction factors were taken from the 0° orientation measurements and were thus not repeated at each cardinal angle. Output measurements with the new test chambers were compared to the absorbed dose measured with the reference A1SL at the standard 0° orientation.

# 3 | RESULTS

Characterization of uncertainty is a critical component of experimental determination of magnetic field quality conversion factors. A comprehensive uncertainty budget analysis for the experimental $k_{B,Q}$ determination via cross-calibration to a reference A1SL ionization chamber is provided in Table 2, including the combined ($k = 1$) uncertainty. Separate uncertainty budgets are provided for the Elekta Unity and ViewRay MRIdian MR-guided radiotherapy systems due to differences

**TABLE 2** Uncertainty budget ($k = 1$) for $k_{B,Q}$ determination via cross calibration to a reference A1SL ionization chamber for the Elekta Unity and ViewRay MRIdian MR-guided radiotherapy systems.

| Component of uncertainty | Elekta Unity (%) | ViewRay MRIdian (%) |
|---|---|---|
| General components | | |
| $(N_{D,W}^{Co60})^{A1SL}/(N_{D,W}^{Co60})^{test}$ | 0.26% | 0.26% |
| $(k_Q)^{A1SL}/(k_Q)^{test}$ | 0.47% | 0.47% |
| Linac output stability | 0.14% | 0.30% |
| **Reference A1SL chamber** | | |
| Standard error of the mean | 0.04% | 0.06% |
| Depth setting | 0.10% | 0.10% |
| Alignment of chamber axis | 0.07% | 0.01% |
| $k_{B,Q}$ | 0.35% | 0.27% |
| $P_{pol}$ | 0.05% | 0.05% |
| $P_{ion}$ | 0.10% | 0.10% |
| $P_{TP}$ | 0.05% | 0.05% |
| $P_{leak}$ | 0.05% | 0.05% |
| $P_{rp}$ | 0.05% | 0.05% |
| **Test chamber** | | |
| Standard error of the mean | 0.04% | 0.06% |
| Depth setting | 0.10% | 0.10% |
| Alignment of chamber axis | 0.07% | 0.01% |
| $P_{pol}$ | 0.05% | 0.05% |
| $P_{ion}$ | 0.10% | 0.10% |
| $P_{TP}$ | 0.05% | 0.05% |
| $P_{leak}$ | 0.05% | 0.05% |
| $P_{rp}$ | 0.05% | 0.05% |
| **Combined ($k = 1$)** | **0.71%** | **0.72%** |

in system performance and experimental setup. The combined ($k = 1$) uncertainty in the experimental determination of the magnetic field quality conversion factor, $k_{B,Q}$, for the Elekta Unity and ViewRay MRIdian MR-linacs was 0.71% and 0.72%, respectively. Due to the cross-calibration technique used in this study, most uncertainty components are duplicated for the reference A1SL chamber and the test chamber. Uncertainty components for the reference A1SL chamber and the test chamber were largely considered independent in this uncertainty budget analysis, making this a conservative total uncertainty. Most of the uncertainty comes from the reference values $N_{D,W}^{Co60}$, $k_Q$, and $k_{B,Q}$, which we could not control. The total $N_{D,W}^{Co60}$ uncertainty quoted from the University of Wisconsin ADCL report is 0.7%, with 0.6% attributed to the ADCL standard ionization chamber calibration at the primary standards lab. Removing this and all other correlated components which were identical for the reference and test chambers, the total uncertainty of the $(N_{D,W}^{Co60})^{A1SL}/(N_{D,W}^{Co60})^{test}$ ratio was reduced to 0.26%. Similarly, while the total uncertainty in $k_Q$ taken from Andreo et al.[21] is 0.6%, when considering the ratio of $k_Q$ values for different chambers the uncertainty in W/e of 0.5%[23] is removed. $k_{B,Q}$ uncertainty is obtained from the weighted average of experimental and Monte Carlo derived $k_{B,Q}$ uncertainties for the A1SL chamber. Removing these terms and considering only the uncertainty due to our experimental procedure, the total user-dependent uncertainty ($k = 1$) was 0.28% and 0.27% for the Elekta Unity and ViewRay MRIdian MR-linacs, respectively. Experimental determination of $k_{B,Q}$ at each cardinal angle relative to the static magnetic field followed the same procedure as the standard 0° orientation, so uncertainty is the same as the overall Elekta Unity uncertainty reported below.

**TABLE 3** Magnetic field quality conversion factor, $k_{B,Q}$, values for 11 ionization chamber models, experimentally determined using ViewRay MRIdian and Elekta Unity MR-guided radiotherapy systems. Chambers were positioned parallel to the magnetic field (0° orientation).

| | $k_{B,Q}$ | |
|---|---|---|
| Ion chamber model | Elekta Unity[a] | ViewRay MRIdian[b] |
| Exradin A12 | 0.999 | 1.001 |
| Exradin A12S | 0.998 | 1.000 |
| Exradin A28 | 0.999 | 0.995 |
| Exradin A26 | 0.996 | 0.996 |
| PTW T31010 | 0.996 | 0.997 |
| PTW T31021 | 0.994 | 0.998 |
| PTW T31022 | 0.991 | 0.998 |
| IBA FC23-C | 1.002 | 1.003 |
| IBA CC25 | 1.002 | 1.002 |
| IBA CC13 | 0.996[c] | 1.001 |
| IBA CC08 | 1.002 | 1.004 |

[a]A combined uncertainty of 0.71% ($k = 1$) applies to all $k_{B,Q}$ data for Elekta Unity.
[b]A combined uncertainty of 0.72% ($k = 1$) applies to all $k_{B,Q}$ data for ViewRay MRIdian.
[c]CC13 $k_{B,Q}$ value has been previously determined by Iakovenko et al. (2020) in the exact Elekta Unity system used in this work and is being reproduced in this table.[18]

Magnetic field quality conversion factors, $k_{B,Q}$, for 11 ionization chamber models experimentally determined via cross-calibration against a reference A1SL chamber are provided in Table 3. $k_{B,Q}$ values were determined using clinical Elekta Unity and ViewRay MRIdian MR-guided radiotherapy systems, with the values in Table 3 corresponding to 0° orientation where the long axis of the ionization chamber is parallel to the static magnetic field direction.

For the high-field 1.5 T Elekta Unity MR-linac, ionization chamber magnetic field quality conversion factors, $k_{B,Q}$, were determined with the ionization chamber long axis positioned at the four cardinal angles relative to the magnetic field. Table 4 shows the resulting $k_{B,Q}$ values measured at 0°, 90°, 180°, and 270° orientations relative to the 1.5 T magnetic field for 10 ionization chambers.

**TABLE 4** Magnetic field quality conversion factor, $k_{B,Q}$, for the ionization chamber long axis positioned at orientations of 0°, 90°, 180°, and 270° relative to the static magnetic field. The percent difference in $k_{B,Q}$ with respect to the standard 0° orientation are shown in brackets. The angular dependence of $k_{B,Q}$ was assessed only for the high-field Elekta Unity system.

| | $k_{B,Q}$ | | | |
|---|---|---|---|---|
| **Ion chamber model** | **0°** | **90°** | **180°** | **270°** |
| Exradin A12 | 0.999 | 0.943 (−5.6%) | 0.998 (−0.1%) | 0.954 (−4.5%) |
| Exradin A12S | 0.998 | 0.945 (−5.3%) | 1.002 (0.4%) | 0.982 (−1.6%) |
| Exradin A28 | 0.999 | 0.959 (−4.0%) | 0.999 (-) | 1.004 (0.5%) |
| Exradin A26 | 0.996 | 1.011 (1.5%) | 0.994 (−0.2%) | 1.015 (1.9%) |
| PTW T31010 | 0.996 | 1.011 (1.5%) | 0.996 (-) | 0.998 (0.2%) |
| PTW T31021 | 0.994 | 1.036 (4.2%) | 0.992 (−0.2%) | 1.008 (1.4%) |
| PTW T31022 | 0.991 | 1.036 (4.5%) | 0.991 (-) | 1.001 (1.0%) |
| IBA FC23-C | 1.002 | 0.983 (−1.9%) | 1.000 (−0.2%) | 0.999 (−0.3%) |
| IBA CC25 | 1.002 | 0.970 (−3.2%) | 1.002 (-) | 0.985 (−1.7%) |
| IBA CC08 | 1.002 | 0.976 (−2.6%) | 1.002 (-) | 1.028 (2.6%) |

## 4 | DISCUSSION

Comparing magnetic field quality conversion factors measured in the high-field 1.5 T Elekta Unity and low-field 0.35 T ViewRay MRIdian, the average $k_{B,Q}$ values were closer to one for the MRIdian system, supporting the conclusion that in general, increased magnetic field strength results in increased impact on charged particle collection. As expected, it was also observed that as the ionization chamber sensitive volume increased, the magnetic field quality conversion factor also tended to increase. Interestingly, the measured $k_{B,Q}$ values in the parallel orientation for all IBA ionization chambers were greater than one, indicating less charge collection in the presence of the magnetic field. In contrast, measured $k_{B,Q}$ values for all PTW and Exradin chambers apart from the A12 chamber were less than one, indicating increased charge collection in the magnetic field. While this may be the result of the ionization chamber design, the percent difference in $k_{B,Q}$ values between IBA and PTW/Exradin chambers of similar volume is within uncertainty meaning these differences could be coincidental.

Characterization of ionization chamber response as a function of angular orientation for the 1.5T Unity MR-linac showed that the impact of the magnetic field on charge collection is minimized when the chamber is aligned parallel to the magnetic field, confirming previously reported results.[8,18] Furthermore, charge collection was the same in the parallel (0°) and antiparallel (180°) orientations, with an identical average $k_{B,Q}$ value. As such, output calibration for MR-guided radiotherapy systems could be completed in either parallel or antiparallel orientation depending on a center's preferences. For chambers positioned in the so-called 90° orientation, where the chamber is perpendicular to the magnetic field and pointing in the direction of the Lorentz force, the magnetic field had the largest impact on charge collection with $k_{B,Q}$ deviating on average 3.2% from unity. In this 90° orientation, charge collection was shown to either increase or decrease depending on the chamber, making this response both non-trivial and chamber-specific. While the impact of angular dependence on $k_{B,Q}$ values is smaller for the 0.35T MRIdian system, it may still have a significant impact.[8,13,24] Further experimental validation in the 0.35T MRIdian MR-linac beam is required to quantify this effect for different ionization chamber models.

Table 5 provides a comparison of our experimentally determined magnetic field quality conversion factor, $k_{B,Q}$, values to available Monte Carlo simulation and experimental results in the literature. Malkov and Rogers derived the magnetic field quality conversion factor, $k_{B,Q}$, of standard cylindrical ionization chambers through Monte Carlo simulation of the chamber geometry and MR-linac magnetic field and radiation beam.[8] Compared to the Monte Carlo simulation data, experimental $k_{B,Q}$ values measured using the Elekta Unity MR-linac had an overall average percent difference of 0.26 ± 0.17%, with all $k_{B,Q}$ values agreeing to within uncertainty ($k = 1$). Previously reported Monte Carlo simulation data for the ViewRay MRIdian system was based on three Cobalt-60 sources in place of the 6 MV FFF linear accelerator beam.[8] Regardless, the magnetic field strength (0.35 T) and orientation are comparable, meaning comparisons between our experimental data gathered on a 6 MV linear accelerator-based MRIdian system and the reported Monte Carlo results calculated for the Co-60 MRIdian system are still relevant. Compared to the Monte Carlo simulation data, the experimental $k_{B,Q}$ values measured using the ViewRay MRIdian 6 MV FFF MR-linac showed an overall average percent difference of 0.19 ± 0.18%. Once again, all measured values agreed within uncertainty ($k = 1$) to the Monte Carlo derived data. This

**TABLE 5** Comparison of our experimentally determined magnetic field quality conversion factor, $k_{B,Q}$, values to Monte Carlo calculations and experimental results in the literature. The percent difference for each literature value compared to our experimental value is presented in brackets. All comparisons are for chambers positioned parallel to the magnetic field (0° orientation).

| Ion chamber model | Source | Method | $k_{B,Q}$ (1.5 T) | $k_{B,Q}$ (0.35 T) |
|---|---|---|---|---|
| Exradin A12 | This Study | Experiment | 0.999 | 1.001 |
| | Malkov and Rogers[8] | Monte Carlo | 0.9983 (0.07%) | 1.0001 (0.09%) |
| | Margaroni et al.[15] | Monte Carlo | 1.006 (−0.70%) | – |
| Exradin A12S | This Study | Experiment | 0.998 | 1.000 |
| | Malkov and Rogers[8] | Monte Carlo | 0.9984 (−0.04%) | 0.9999 (0.01%) |
| Exradin A28 | This Study | Experiment | 0.999 | 0.995 |
| | Krauss et al.[19] | Experiment | – | 0.9962 (−0.12%) |
| Exradin A26 | This Study | Experiment | 0.996 | 0.996 |
| | Krauss et al.[19] | Experiment | – | 0.9939 (0.21%) |
| | Margaroni et al.[15] | Monte Carlo | 1.0034 (−0.74%) | – |
| IBA FC23-C | This Study | Experiment | 1.002 | 1.003 |
| | Malkov and Rogers[8] | Monte Carlo | 0.9980 (0.40%) | 1.000 (0.30%) |
| IBA CC25 | This Study | Experiment | 1.002 | 1.002 |
| | Malkov and Rogers[8] | Monte Carlo | 0.9987 (0.33%) | 0.9988 (0.32%) |
| IBA CC13 | This Study | Experiment | – | 1.001 |
| | Malkov and Rogers[8] | Monte Carlo | 0.9990 | 1.000 (0.10%) |
| IBA CC08 | This Study | Experiment | 1.002 | 1.004 |
| | Malkov and Rogers[8] | Monte Carlo | 0.9975 (0.45%) | 0.9992 (0.48%) |
| PTW T31010 | This Study | Experiment | 0.996 | 0.997 |
| | Malkov and Rogers[8] | Monte Carlo | 0.9933 (0.27%) | 0.9969 (0.01%) |
| | Cervantes et al.[14] | Monte Carlo | 0.9950 (0.10%) | – |
| | Margaroni et al.[15] | Monte Carlo | 0.9986 (−0.26%) | – |
| | Krauss et al.[19] | Experiment | – | 0.9933 (0.37%) |
| PTW T31021 | This Study | Experiment | 0.994 | 0.998 |
| | Cervantes et al.[14] | Monte Carlo | 1.0160 (−2.17%) | – |
| | Margaroni et al.[15] | Monte Carlo | 1.0019 (−0.79%) | – |
| | Krauss et al.[19] | Experiment | – | 0.9937 (0.43%) |
| PTW T31022 | This Study | Experiment | 0.991 | 0.998 |
| | Cervantes et al.[14] | Monte Carlo | 0.9970 (−0.60%) | – |
| | Margaroni et al.[15] | Monte Carlo | 1.0022 (−1.12%) | – |

close agreement between Co-60 MRIdian-based Monte Carlo simulation results and our 6 MV FFF linac-based MRIdian experimental data may also suggest that beam quality does not have a large impact on the magnetic field quality conversion factor. It is important to note that the reference $k_{B,Q}$ values for our reference A1SL ionization chamber were calculated by weighted average in which the simulated $k_{B,Q}$ values by Malkov and Rogers[8] were one of two sources. This link may have influenced the observed close agreement to our experimental results.

Through Monte Carlo simulation of the 1.5 T Elekta Unity MR-linac, Cervantes et al. characterized the performance of three small-cavity ionization chambers (PTW T31010, T31021, and T31022) at different orientations relative to the magnetic field.[14] At the standard 0° orientation with the chamber parallel to the magnetic field, percent differences for our experimentally measured magnetic field quality conversion factors compared to the Monte Carlo simulated data were 0.06%, −2.18%, and −0.65% for the T31010, T31021, and T31022 chambers, respectively. While $k_{B,Q}$ values for the T31010 and T31022 chambers agreed within uncertainty, the T31021 $k_{B,Q}$ value was considerably smaller when determined using our experimental approach. This disagreement is interesting due to otherwise excellent agreement with Monte Carlo simulations for the other chambers characterized by Cervantes et al. as well for chambers characterized by Malkov and Rogers. Furthermore, the experimental $k_{B,Q}$ value of 0.994 is in line with experimental $k_{B,Q}$ values for other small-volume ionization chambers determined in this study. Considering the

90° and 270° chamber orientations relative to the magnetic field, our experimental $k_{B,Q}$ values agreed to within uncertainty compared to Monte Carlo simulations for the T31010 and T31022 chambers, but not for the T31021 chamber.

Margaroni et al. calculated magnetic field quality conversion factors for 12 ionization chamber models through Monte Carlo simulation of the 1.5T Elekta Unity MR-linac.[15] Five of the simulated chamber models overlapped with our experimental results. In the standard 0° parallel orientation, the average percent difference between the simulated and experimental $k_{B,Q}$ values was 0.72 ± 0.31%. Interestingly, agreement was better for the 180° anti-parallel orientation, with an average percent difference of 0.47 ± 0.27%, in either case demonstrating good agreement. It was also observed that congruence between Monte Carlo calculated $k_{B,Q}$ from Margaroni et al. and our experimental results were better for large volume chambers, with an average percent difference of 0.48 ± 0.31% for chamber volume >0.1 cc and 0.88 ± 0.21% for chamber volume <0.1 cc in the parallel orientation. Considering the perpendicular 270° and 90° orientations, agreement to the simulated $k_{B,Q}$ values were worse, with average percent differences of 1.52 ± 0.57% and 1.47 ± 0.67%, respectively.

For the ViewRay MRIdian MR-linac, Krauss et al. utilized water calorimetry as a primary standard to directly measure absorbed dose, facilitating direct determination of the magnetic field quality conversion factor, $k_{B,Q}$, for several cylindrical ionization chambers.[19] In addition, the use of water calorimetry as a primary standard allowed for reduced uncertainty, which was determined to be less than 0.5%.[19] While the A26 and A28 chambers characterized by Krauss et al. were the MR-safe variant, the chamber geometry is identical. Compared to $k_{B,Q}$ values determined by Krauss et al., our cross-calibration based $k_{B,Q}$ values had an overall average percent difference of 0.28 ± 0.14%. This excellent agreement compared to the $k_{B,Q}$ values determined based on water calorimetry as a primary standard provide increased confidence in our cross-calibration approach.

There is a lack of published $k_{B,Q}$ data available for comparison. For the ionization chambers characterized in our study, there is no experimental data for the Elekta Unity MR-linac, experimental data is limited to only a few ionization chambers for the ViewRay MRIdian system, and there are no published Monte Carlo simulations for the updated MRIdian system with an integrated 6 MV FFF linear accelerator, to our knowledge. The $k_{B,Q}$ values provided in this manuscript are relevant only to 6 MV FFF linear accelerator-based MRIdian systems. Further experimental validation is required to determine magnetic field quality conversion factors for Co-60 MRIdian systems. Small differences in the construction of individual chambers of the same model may have an impact on the magnetic field quality conversion factor.[25] As we had access to only one ionization chamber per model type, we were unable to assess this chamber-to-chamber variation in $k_{B,Q}$. Woodings et al. analyzed the consistency of magnetic field quality conversion factors in 12 PTW 30013 and 13 FC65-G chambers, demonstrating standard deviations of 0.19% and 0.15%, respectively, in the parallel orientation.[25] Although this spread is a function of chamber type, if utilizing our reported $k_{B,Q}$ values during MR-linac output calibration, we recommend that the user adds an additional 0.2% uncertainty in quadrature to account for chamber-to-chamber variation. With the advent of MR-guided radiotherapy systems, vendors have begun to offer MR-safe equivalent versions of standard ionization chambers. Future work includes a comprehensive characterization of MR-safe ionization chamber performance in magnetic fields compared to the equivalent standard chambers, including experimental determination of the magnetic field quality conversion factors, $k_{B,Q}$.

## 5 | CONCLUSIONS

The magnetic field quality conversion factor, $k_{B,Q}$, was experimentally determined for a comprehensive selection of eleven commercially available ionization chambers via cross-calibration with a reference A1SL ionization chamber. $k_{B,Q}$ was characterized for both the Elekta Unity and ViewRay MRIdian, representing the two commercially available and most commonly used MR-guided radiotherapy systems. For the 0° orientation, experimental $k_{B,Q}$ values largely agreed within uncertainty to Monte Carlo simulations, providing additional validation of this simulated data. Furthermore, the magnetic field quality conversion factors were characterized for various chamber orientations relative to the high-field 1.5T Unity magnetic field, providing reference data for alternative calibration setups or for quantification of ionization chamber sensitivity to orientation during reference setup. This characterization of ionization chamber response in the magnetic field of MR-linacs is critical for accurate output calibration, and the data provided in this report could be useful as a reference for upcoming MR-guided radiotherapy reference dosimetry protocols, where experimental data has previously been lacking.

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## CONFLICT OF INTEREST STATEMENT

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