Skin dose in longitudinal and transverse linac-MRIs using Monte Carlo and realistic 3D MRI field models

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(Received 1 June 2012; revised 5 August 2012; accepted for publication 10 September 2012; published 3 October 2012)

Purpose: The magnetic fields of linac-MR systems modify the path of contaminant electrons in photon beams, which alters patient skin dose. To accurately quantify the magnitude of changes in skin dose, the authors use Monte Carlo calculations that incorporate realistic 3D magnetic field models of longitudinal and transverse linac-MR systems.

Methods: Finite element method (FEM) is used to generate complete 3D magnetic field maps for 0.56 T longitudinal and transverse linac-MR magnet assemblies, as well as for representative 0.5 and 1.0 T Helmholtz MRI systems. EGSnrc simulations implementing these 3D magnetic fields are performed. The geometry for the BEAMnrc simulations incorporates the Varian 600C 6 MV linac, magnet poles, the yoke, and the magnetic shields of the linac-MRIs. Resulting phase-space files are used to calculate the central axis percent depth-doses in a water phantom and 2D skin dose distributions for 70 μm entrance and exit layers using DOSXYZnrc. For comparison, skin doses are also calculated in the absence of magnetic field, and using a 1D magnetic field with an unrealistically large fringe field. The effects of photon field size, air gap (longitudinal configuration), and angle of obliquity (transverse configuration) are also investigated.

Results: Realistic modeling of the 3D magnetic fields shows that fringe fields decay rapidly and have a very small magnitude at the linac head. As a result, longitudinal linac-MR systems mostly confine contaminant electrons that are generated in the air gap and have an insignificant effect on electrons produced further upstream. The increase in the skin dose for the longitudinal configuration compared to the zero B-field case varies from ~1% to ~14% for air gaps of 5–31 cm, respectively. (All dose changes are reported as a % of Dmax.) The increase is also field-size dependent, ranging from ~3% at 20 × 20 cm² to ~11% at 5 × 5 cm². The small changes in skin dose are in contrast to significant increases that are calculated for the unrealistic 1D magnetic field. For the transverse configuration, the entrance skin dose is equal or smaller than that of the zero B-field case for perpendicular beams. For a 10 × 10 cm² oblique beam the transverse magnetic field decreases the entry skin dose for oblique angles less than ±20° and increases it by no more than 10% for larger angles up to ±45°. The exit...
I. INTRODUCTION

Currently there are two working prototype linac-MR systems that would allow for patient imaging in real-time, providing soft-tissue contrast suitable for real-time tumor tracking.\(^1\text{–}\text{3}\) The linac-MR prototype described in Ref. 3 employs a split-solenoid cylindrical MRI system that generates a main magnetic (\(B\)) field along the superior–inferior patient axis and transverse (i.e., perpendicular) to the linac’s x-ray beam. Our group’s design (Refs. 1 and 2) is a rotating biplanar (RBP) system producing a main MRI magnetic field perpendicular to the head–foot patient axis. In this case, two orientations of the linac’s x-ray beam direction relative to the magnetic (\(B\)) field are possible: perpendicular for the transverse configuration,\(^4\) and parallel in the longitudinal configuration.\(^5\) It has been established that systems with transverse geometry may suffer from significant dose perturbations in the patient, particularly when higher magnetic field strengths are used. These perturbations are caused by several effects related to the direction of the Lorentz force acting on high energy electrons. These effects include changes to percentage depth-dose, lateral shifts in dose distributions, and electron return effects (EREs) that increase exit dose and cause cold and hot spots at lung/tissue interfaces.\(^4\text{–}\text{6}\text{,}\text{8}\) With a longitudinal geometry many of these dosimetric issues within the patient are eliminated, since the Lorentz force instead restricts the radial spread of secondary electrons in the patient when the \(B\)-field is parallel to the photon beam axis.\(^9\text{,}\text{10}\) Use of a parallel magnetic field also reduces the penumbral width.\(^3\text{,}\text{11}\) Further, in a recent study Kirkby et al.\(^5\) showed that the longitudinal rotating biplanar system exhibits an increase in dose to the planning target volume (PTV), potentially offering reduced normal tissue dose for the same PTV dose. This effect is more pronounced for higher magnetic field strengths and in low density tissue such as lung.

Another potentially important aspect of linac-MR dosimetry is the impact on patient skin dose. The main contribution to skin dose at the beam entrance comes from the contaminant electrons present in megavoltage photon beams. The magnetic field of the MRI unit perturbs the fluence of contaminant electrons and hence alters the skin dose in both linac-MR configurations. In a transverse system the Lorentz force sweeps the contaminants away from the incident path of the x-ray beam and a reduction in the skin dose would be expected. Changes in the entrance and exit doses from a transverse magnetic field in a fixed cylindrical MRI geometry have been studied extensively.\(^12\text{–}\text{14}\) For longitudinal linac-MR systems only a small increase in the entrance skin dose is predicted, due to the rapid decay of the realistic magnetic fringe fields. For transverse linac-MR systems, changes to the entrance skin dose are small for most scenarios. For the same geometry, on the exit side a fairly large increase is observed for perpendicular beams, but significantly drops for large oblique angles of incidence. The observed effects on skin dose are not expected to limit the application of linac-MR systems in either the longitudinal or transverse configuration.

Conclusions: For longitudinal linac-MR systems only a small increase in the entrance skin dose is predicted, due to the rapid decay of the realistic magnetic fringe fields. For transverse linac-MR systems, changes to the entrance skin dose are small for most scenarios. For the same geometry, on the exit side a fairly large increase is observed for perpendicular beams, but significantly drops for large oblique angles of incidence. The observed effects on skin dose are not expected to limit the application of linac-MR systems in either the longitudinal or transverse configuration.
et al.\textsuperscript{15} means that the significance of skin dose increases in a longitudinal linac-MR can only be determined with further comprehensive investigation based on a more definite determination of the fringe fields. The magnetic fields modeled in their work were generally unrealistic for three reasons: (1) the large extension of the uniform MRI field to the linac head is unnecessary and impractical from a design perspective, (2) the fringe fields are much too large as a result of the effects of the yoke and magnetic shields not being incorporated in their models, and (3) no transverse ($x$ and $y$) components of the $B$-fields were considered.

In this work, we report on the entry and exit skin dose for linac-MRI system based on RBP geometry. First, we calculate realistic model magnetic fields of several magnet designs, using finite element method (FEM), to determine the characteristics of realistic fringe fields. Second, by using the realistic 3D magnetic fields, we accurately predict the increase in skin dose in a longitudinal RBP linac-MR system and the changes in entrance and exit skin doses in a transverse RBP linac-MR system. These changes are quantified for various field sizes, for various air gaps between the phantom and the magnet pole (longitudinal case), and for various angles of obliquity of the incident beam relative to the patient surface (transverse case). All dosimetric simulations were performed using EGSnrc Monte Carlo codes that were modified to include vector magnetic fields.

II. MATERIALS AND METHODS

II.A. Realistic 3D magnetic fields

Commercially available finite element method software packages, Comsol Multiphysics,\textsuperscript{16} and Opera-3D,\textsuperscript{17} were used to calculate the realistic 3D magnetic fields for the different MRI systems considered below. Corresponding measurements of the magnetic fields were not performed since the systems modeled are not yet ready for experimentation. However, these FEM packages have been benchmarked extensively, previously. Various groups worldwide have compared calculated magnetic field vectors for other magnet designs from these software packages with experimental measurements.\textsuperscript{18–20}

II.A.1. Generic yoked “Helmholtz-pair” MRI system

To investigate the effects of realistic magnetic fields on patient skin dose, we first simulated a generic Helmholtz coil assembly. The magnetic fields from the “Helmholtz-pair” (Helmholtz) coils were calculated by using Comsol Multiphysics.\textsuperscript{16} Superconducting coils with inner radii of 40 and a $10 \times 5$ cm$^2$ rectangular cross section was modeled. Current densities of 5.975 and 2.988 kA/cm$^2$, which can be achieved by using currently available superconducting materials such as (NbTi) and (MgB$_2$) (Ref. 21) were used to produce magnetic field strengths of 1.0 and 0.5 T, respectively. The 0.5 T Helmholtz MRI is representative of a commercially available system. Although a 1.0 T Helmholtz MRI is currently not commercially available, it is used to facilitate comparison with Oborn et al.’s magnetic field models that were based on a 1.0 T main field. For each system modeled, a Helmholtz configuration was used by placing one coil on each side of the magnet. The pair of coils was held by a simple yoke consisting of a pair of AISI 1020 carbon steel\textsuperscript{22} disks (20 cm thick, 75 cm radius). The two sides of the magnet were separated by four steel posts (radius of 7.5 cm). In order to simulate a longitudinal RBP geometry, a 32.5 cm radius hole is made in each disk to ensure an unobstructed path for the linac x-ray beam. To further investigate the effect of a yoke structure on the MRI fringe fields, a 1.0 T Helmholtz-pair system without a yoke was also simulated. The FEM simulations used quadratic vector basis functions with tetrahedral finite elements that were optimized through Delaunay triangulation. A Neumann or natural boundary condition was specified at the model’s external boundaries. The magnetic fields were solved through a direct solver. Further details regarding this simulation can be found in Ref. 16.

II.A.2. Midfield (0.56 T) yoked biplanar superconducting (CC) magnet assembly

Opera-3D (Ref. 17) was used to calculate the complete 3D magnetic field generated by the realistic linac-MRI systems illustrated in Fig. 1. Details of the modeling procedure and magnetic field analysis of this system have been reported by Tadic and Fallone,\textsuperscript{23} the important aspects of which are reviewed below.

The linac-MR system consists of three principal magnetic components: a biplanar superconducting MRI magnet assembly, a treatment machine assembly including the linac and associated components in the linac head, and a mechanical gantry link. The magnet assembly consists of a C-shaped yoke structure constructed from AISI 1020 plain carbon steel and a pair of magnetic pole pieces made of Armco magnetic steel. The magnetic pole pieces act to enhance the magnetic field strength in the imaging volume and can be appropriately shaped to improve the field homogeneity.\textsuperscript{24,25} This magnet assembly has a 60 cm pole-to-pole separation, and a 24 cm diameter hole bored through the yoke and pole structures. A single large coil constructed from MgB$_2$ high-temperature superconducting material surrounds each of the pole pieces and acts as a magnetic source. These coils are of rectangular cross section and possess a current density of 1.764 kA/cm$^2$.

The magnetic assembly surrounding the linac and the associated components is illustrated in Fig. 2. It is comprised of the passive magnetic linac shielding, the electron gun casing, the linac base, secondary collimator base, and the multileaf collimator (MLC) base, all of which were taken to be constructed from AISI 1020 steel.\textsuperscript{23} Although the actual treatment assembly consists of many relatively small magnetic elements, the structures modeled for this study possess equivalent magnetic masses and approximate the true distribution of magnetic material, making simulation of the entire system practical. Magnetic field simulations were performed with the treatment assembly in both parallel and perpendicular
linac-MRI configurations, as depicted in Fig. 1. The treatment assemblies were positioned along the beam axis such that linac target to isocenter distances of 126 and 146 cm were obtained for the parallel and perpendicular configurations, respectively. The latter distance was necessarily increased to avoid physical interference of the MLC subassembly with the cryostat and pole structures of the MRI magnet.

The mechanical gantry link, as shown in Fig. 1, was also modeled as being composed of AISI 1020 steel. This structure acts to provide increased mechanical support of the linac-MRI system and provides the necessary components for mating the integrated system with a rotating gantry. Due to symmetry in

II.B. Monte Carlo simulations

All Monte Carlo simulations were performed using EGSnrc and BEAMnrc radiation transport codes with algorithms implemented to account for the magnetic field deflection of charged particles.4, 5, 9, 26–29 The simulations were run on the Western Canadian Research Grid (Westgrid) high performance computing cluster employing 100 or more processors. For the generation of particle phase spaces, the BEAMnrc simulations included models of a Varian 600C 6 MV linac, the magnet poles, the yoke, and the magnetic shields of the MRIs. The 6 MV photon beam source has been modeled and benchmarked in a previous work.30 Figure 3(a) displays a schematic diagram of the simulated longitudinal RBP linac-MR assembly, with an isocenter at 126 cm distance from the linac target. The air gap is defined as the distance between the phantom surface and the magnet pole, and the air column refers to the hole in the yoke and pole plate. The transverse RBP linac-MR configuration was modeled by rotating the entire Linac and shielding structures by 90° with respect to the magnet (see Fig. 1), and then displacing these objects 20 cm further away from isocenter (source to axis distance, SAD = 146 cm) than in the longitudinal configuration as described in Sec. II.A.2. For the longitudinal configuration, phase-space files were generated at the phantom surface for radiation field sizes of $5 \times 5$, $10 \times 10$, $15 \times 15$, and $20 \times 20$ cm$^2$ at an air gap of 21.5 cm. The phase-space calculations were then repeated for air gaps of 6.5, 11.5, 16.5, 21.5, 26.5, and 31.5 cm for
FIG. 3. (a) Schematic diagram of the longitudinal linac-MR system with the isocenter at 126 cm. (b) The CAX magnetic field maps of our realistic 3D models versus a 1D model (Ref. 15). The 1.0 T MRI of the 1D model extends only to the end of the magnet in Fig. 2 of Ref. 15.

a fixed $10 \times 10$ cm$^2$ field size. For the transverse geometry, phase spaces at a distance of 116 cm from the source (20 cm prior to the phantom surface) were simulated for field sizes of $5 \times 5, 10 \times 10, 15 \times 15,$ and $20 \times 20$ cm$^2$. In both geometries, field size was defined at the machine isocenter. The number of particle histories simulated for the longitudinal and transverse systems were $3 \times 10^8$ and $2 \times 10^9$ which resulted in a total of $\sim 2.3 \times 10^8$ (with $\sim 0.4\%$ electrons) and $\sim 1.4 \times 10^9$ (with $\sim 0.3\%$ electrons) particles in the phase-space files, respectively. Directional Bremsstrahlung splitting was used with a splitting number of 1000. The 3D MRI vector magnetic fields were incorporated in the BEAMnrc models in the calculation of the phase-space files (see Sec. II.B.1).

With phase spaces for the different field size/air gap combinations for the two RBP linac-MR configurations as the input, DOSXYZnrc was then used to score dose distributions in a $30 \times 30 \times 20$ cm$^3$ water phantom (20 cm dimension along the photon beam direction). The CAX percent depth-doses were scored using $2 \times 2$ cm$^2 \times 70$ μm (along beam direction) voxels for depths up to 1.0 cm, and $2 \times 2 \times 0.1$ cm$^3$ voxels at greater depths. For simulations of the transverse geometry, the exit skin doses were also scored using 70 μm voxels in the depth direction in the last 1.0 cm depth of the phantom. A 2D dose distribution with lateral scoring resolution of $0.2 \times 0.2$ cm$^2$ was also generated for each of the 70 μm layers to study the entrance and exit doses in more detail. For the transverse geometry, the effect of variation in the angle of incidence, with respect to the patient surface, on entry and exit doses was also investigated. Entry and exit surface angles of $-45^\circ, -30^\circ, -15^\circ, 0^\circ, +15^\circ, 30^\circ,$ and $45^\circ$ were simulated for a $10 \times 10$ cm$^2$ field size by rotating the water phantom about the systems’ rotation axis (see Fig. 1) passing through the middle of the phantom at 10 cm depth. This was achieved by varying the parameter “theta” and setting the parameters “phi” and “phicol” to zero in the DOSXYZnrc input file. These parameters define the rotations of the phase space relative to the DOSXYZnrc coordinate system. For a head-first-supine patient, positive/negative surface angles then correspond to clockwise/counter-clockwise rotation of the linac gantry.
The DOSXYZnrc simulations were run with 2 and 20 billion particle histories for the percent depth-doses (PDDs) and 2D dose distributions, respectively. Statistical uncertainties of less than 1% were achieved for all voxels in the PDD simulations. The 2D dose distributions had uncertainties of less than 1% with ±2 cm of the CAX, and no more than 3.5% elsewhere. Transport cutoff parameters for all EGSnrc simulations were set to AP = PCUT = 0.01 MeV for photons. For electrons, cutoffs of AE = ECUT = 0.521 MeV were used in order to prevent prematurely terminating the transport of contaminant electrons.

II.B.1. Implementation of the 3D magnetic fields

As discussed in previous works, the EGSnrc Monte Carlo simulations in the presence of a uniform magnetic field are performed by modifying the macro packages beammrc_user.macros.mortran and emf_macros.mortran. In this work these macros are further modified to read the discrete 3D B-field from a file and to interpolate for any particle position. These macros are only invoked after the completion of a conventional charged particle step, i.e., in the absence of the electromagnetic field. During a condensed-history transport step the exact trajectory of the particle is not modeled, nor are the exact forms of inelastic and multiple scattering forces known. Therefore, EGSnrc Monte Carlo implements the approximation that the deflections of electrons and positrons from inelastic scattering, multiple scattering, and from the external electromagnetic field can be decoupled. For accurate simulation of charged particle transport in electromagnetic fields, this approximation requires that the step sizes within the condensed-history algorithm be small enough to ensure that: (a) the relative change in the particle’s kinetic energy remains small, (b) the change in the magnitude of the electromagnetic field across a step is small, and (c) the relative change in the particle’s direction of motion is small. Condition (a) is easily satisfied in this work since the electric field is absent and the magnetic field does not change the particle’s energy. Condition (b) is imposed in emf_macros.mortran by a macro that restricts the maximum change in the magnetic field to

\[
\frac{0.02 \times s}{|\vec{B}(\vec{r}_f) - \vec{B}(\vec{r}_0)|/|\vec{B}(\vec{r}_0)|} \leq 2, \tag{1}
\]

where \(s\) is the conventional step size, \(\vec{r}_0\) and \(\vec{r}_f\) are the particle’s positions at the beginning and end of the step, and the value 0.02 is a user-defined parameter that sets a 2% upper limit on the amount of change of the magnetic field over the transport step. Condition (c) is imposed by a macro that restricts the maximum step size to

\[
\frac{0.02 \times m_e c^2 \times \beta(E_0) \gamma(E_0)}{q_e c \times |\vec{B}_{\perp}|} \leq 2, \tag{2}
\]

where \(m_e c^2\) is the electron’s rest energy, \(c\) is the speed of light, \(q_e\) is the charge of an electron, \(\beta(E_0)\) and \(\gamma(E_0)\) are the familiar relativistic factors, \(E_0\) is the electron’s energy at the start of the step, \(|\vec{B}_{\perp}|\) is the magnitude of the magnetic field perpendicular to the electron track. The value 0.02 is again a user-defined parameter corresponding to a maximum 2% change in direction over the transport step. Condition (c) will be violated for electrons undergoing large angle deflections introduced by multiple scattering. However, the error introduced is negligible except in high-Z media subjected to strong magnetic fields.

II.B.2. Benchmarks

The accuracy of the magnetic field implementation in the EGSnrc Monte Carlo simulations presented in this work was verified by benchmarking electron trajectories against results generated using the GEANT4 Monte Carlo package and the FEM package Opera-3D. Raaijmakers et al. have experimentally verified GEANT4 Monte Carlo simulations of MR-linac dose effects. The trajectory simulations were performed in vacuum to eliminate medium interactions, in a volume measuring 60 × 60 × 126 cm³. The trajectories of electrons, initially with a purely longitudinal (z-directed) momentum, were calculated in both uniform and complex 3D magnetic fields. In the first comparative study, the gyration radius \(r_g\) of an electron with a given kinetic energy (varied from 0.1 to 10.0 MeV) in the presence of a uniform magnetic field (varied from 0.005 to 5.0 T) was determined from each simulation and \(r_g\) was compared against the analytical prediction,

\[
r_g = \frac{p_{\perp} (\text{MeV} c)}{3.00 B (\text{T})}, \tag{3}
\]

where \(p_{\perp}\) is the component of the electron’s momentum perpendicular to the magnetic field lines, \(c\) is the speed of light, and \(B\) is the field strength in Tesla.

The second benchmarking study used a hypothetical full 3D magnetic field defined as

\[
B_x = B_y = 0.01 \times \exp(-0.1z), \tag{4a}
\]

\[
B_z = \begin{cases} 0.15 \exp\left[-(80 - z)\right] & \text{if } z \leq 80 \\ 0.15 & \text{elsewhere} \end{cases}. \tag{4b}
\]

In Eqs. (4a) and (4b), \(B_x\), \(B_y\), and \(B_z\) are the x, y, and z components of the magnetic field in Tesla while \(z\) is the z-coordinate (cm) along the initial direction of electron’s momentum. A 0.5 MeV electron initially traveling in the longitudinal (z-) direction was tracked in the presence of this 3D magnetic field. The B-field was calculated externally for a rectilinear grid with a grid spacing of 1 cm in all directions and then was incorporated into the EGSnrc and GEANT4 simulations. The interpolation algorithm used in both EGSnrc and GEANT4 to calculate the magnetic field at an arbitrary point from the rectilinear magnetic field map was also used to interpolate the magnetic fields at each of Opera-3D’s irregularly spaced FEM mesh nodes. The electron trajectories from the various simulations were then extracted and compared.

III. RESULTS AND DISCUSSION

III.A. Realistic 3D magnetic fields

In Fig. 3(b) the extension of the CAX B-fields is plotted with respect to the longitudinal linac-MR geometry. Our
realistic 3D models exhibit fringe fields that rapidly fall off in the air column and drop to very small residual values as they enter the linac collimation system. The 0.56 T rotating biplanar configuration, which is under installation by our group, has a fringe field with the most rapid fall off, dropping to 18 G at the linac MLC. The generic yoked Helmholtz coil systems display fringe fields that drop to below \( \sim 240 \) G at the linac MLC. The 1.0 T Helmholtz magnet without a yoke has a fringe field of \( \sim 0.19 \) T at the linac MLC, nearly eightfold larger than the yoked Helmholtz fringe field, showing the significance of incorporating a yoke into the Monte Carlo simulations. For comparison an unrealistic 1D model\(^{15}\) of a 1.0 T MRI and a fringe field that varies as \( 1/r^2 \) is shown.

The 3D \( B \)-field of our RBP transverse geometry (not shown) has a main field of 0.56 T and a fringe field similar to that of the 0.56 T longitudinal configuration.

III.B. Monte Carlo simulations in the presence of magnetic fields: Benchmarks

For each uniform magnetic field used in the first comparative study described earlier, the radius of curvature of the electron trajectories from both EGSnrc and GEANT4 deviates by less than 1% from the analytical prediction [Eq. (3)] through at least five complete gyration cycles. Figure 4(a) exhibits the electron trajectories obtained in the second comparative study with the full 3D magnetic field defined by Eqs. (4a) and (4b) being present. The electron trajectories from both the EGSnrc and GEANT4 are in very good agreement with the finite element method prediction by Opera-3D. The discrepancy between different electron tracks is displayed in Fig. 4(b). Although there are no interactions in these Monte Carlo simulations done in vacuum EGSnrc and GEANT4 still use different algorithms to determine the size of the next electron step. Thus, the length of individual steps will be different for the two packages, and as a result the deflections are applied at different locations along the trajectories. Since the magnetic field is approximated to be constant over the course of each step and the deflection is calculated based only on the magnetic field at the end of each step the deflections themselves will also differ. These discrepancies propagate throughout the simulation in vacuum resulting in a maximum cumulative difference of only 0.7 mm between each of these electron trajectories after 1.26 m of electron travel in the \( z \)-direction. This small cumulative difference verifies the accuracy of the implementation of the 3D magnetic field and the sufficiency of the step size sampling algorithm in our EGSnrc Monte Carlo simulations.

III.C. RBP longitudinal linac-MR system

To quantify the confinement effect of different MR fringe fields on contaminant electrons, we extracted the electron energy fluence spectra from phase-space files for a \( 10 \times 10 \) cm\(^2\) field scored below the linac MLC, below the air column, and at the phantom surface (21.5 cm air gap). Figure 5 depicts these spectra obtained with (a) no \( B \)-field, (b) in the presence of our 0.56 T superconducting magnet, and (c) for the 1D \( 1/r^2 \) model.\(^{15}\) In the absence of the magnetic field there is a significant reduction of the electrons (scored in the \( 5 \times 5 \) cm\(^2\) central region) from below the MLC to the end of the air column (\( \sim 46 \) cm distance). This is simply due to the lateral scatter of the electrons in the air column that displaces a substantial number of electrons outside the scoring region. Similarly, a further, albeit smaller, reduction is observed within the 21.5 cm air gap. The energy fluence of the electrons below the MLC is essentially the same with and without our realistic 3D magnetic field [Figs. 5(a) and 5(b)], indicating that the confinement effect of the small fringe field penetrating the linac collimation system is insignificant. Interestingly, the realistic \( B \)-field somewhat reduces the energy fluence at the end of the air column, compared to the \( B = 0 \) T case. The exact cause of this reduction was not investigated, but may be explained by the fact that the \( x \)- and \( y \)-components of the 3D fringe field have rather complicated characteristics. Within the air gap, however, the 0.56 T MRI field clearly exhibits a confinement...
FIG. 5. Energy fluence spectra of contaminant electrons for a $10 \times 10$ cm² field scored below the linac head (MLC), below the air column, and at the phantom surface (21.5 cm air gap) with (a) no magnetic field, (b) 0.56 T superconducting (CCI) magnet, and (c) 1D $(1/r^2)$ model (Ref. 15). These data were extracted from the central $5 \times 5$ cm² region of the phase space. The energy fluences are normalized to the initial particle history in the Monte Carlo simulations.

effect resulting in a small increase in the electron energy fluence below $\sim 2.5$ MeV. In the 1D magnetic field model\textsuperscript{15} a large fringe field encompasses the Linac collimation system and so most contaminant electrons generated anywhere in the Linac head, air column, and air gap are trapped and directed to the phantom surface with minimal lateral spread, giving rise to a “drastic” focusing of electrons in the center [Fig. 5(c)]. It should be noted that the 1D model focuses the electrons into a region smaller than $5 \times 5$ cm² and averaging over a smaller central area (e.g., $2 \times 2$ cm²) would result in an even larger energy fluence (not shown here).

III.C.1. Skin dose in the entry region

Figure 6(a) displays the first 2 cm of the CAX percent depth-doses of a $10 \times 10$ cm² field delivered by a longitudinal linac-MR system with the magnetic fields of Fig. 3, and the phantom surface placed normal to the beam at a 21.5 cm air gap. With the 1D $B$-field model\textsuperscript{15} an extreme skin dose of $\sim 450\%$ is observed, whereas with the realistic 3D models the skin doses are only slightly higher than the no $B$-field case [Fig. 6(b)]. In the presence of the 0.5 and 1.0 T yoked Helmholtz fields the skin dose is approximately 1.5% higher than that of the 0.56 T superconducting systems, which itself only $\sim 8\%$ larger that the skin dose when there is no magnetic field. However, the 1.0 T Helmholtz system without the yoke results in a $\sim 65\%$ skin dose, compared to $\sim 21\%$ for the yoked system, which clearly indicates the necessity of incorporating the yoke in the magnetic field model in order to simulate a realistic field map.

The effect of the field size on the entry skin dose for our 0.56 T superconducting magnet linac-MR is compared to the same linac with zero magnetic field in Fig. 7. Line profiles through the middle of square fields of sizes from $5 \times 5$ to $20 \times 20$ cm² are shown in Fig. 7(a). The observed reduction of penumbra when the magnetic field is present is due to the Lorentz force confining the scattered electrons and focusing them parallel to the CAX. It is obvious that the increase in the skin dose due to the longitudinal $B$-field becomes smaller as the field size increases. The CAX entry skin dose as a function of field size is displayed in Fig. 7(b). In both zero and 0.56 T $B$-field the phantom surface is positioned with a 21.5 cm air gap, and the skin dose is calculated as the dose deposited in the first 70 $\mu$m of the phantom. The increase in skin dose due
FIG. 7. (a) Entry skin dose line profiles through the CAX for our longitudinal linac-MR system in the presence/absence of the realistic 3D B-field model. (b) Comparison of the entry skin dose of the longitudinal linac-MR system with the 0.56 T superconducting magnet to that without a magnetic field as a function of the field size (air gap of 21.5 cm). Lateral voxel sizes of 0.2 and 2 cm were used in (a) and (b) respectively.

to the magnetic field is greater for smaller field sizes, being 11% at $5 \times 5$ cm$^2$ and only 3% for a $20 \times 20$ cm$^2$ field. This is mostly due to the fact that with a small collimator there is a relatively large amount of contaminant electrons from the linac head (mostly flattening filter) that pass through or scatter off the jaws and are subsequently lost in the absence of the magnetic field, but are instead refocused in the presence of the longitudinal B-field. With a larger collimator the skin dose increase due to the longitudinal B-field is less pronounced, as a large portion of the contaminant electrons already contribute to the skin dose in the zero field case.

In Fig. 8, the dependence of the skin dose increase on the size of the air gap is illustrated. At the smallest air gap of 6.5 cm in this study, the effect of the magnetic field is to increase the skin dose by only 1% from 11% (no B-field) to 12% (with B-field). The fact that the skin doses are nearly identical close to the magnet pole (zero air gap) confirm that the fringe fields of our realistic MRI systems have minimal effect on the skin dose since they do not significantly penetrate the Linac collimation system. With increasing air gap, there is very little change in the skin dose in the no B-field case, while in the presence of the B-field there is an approximately linear increase with air gap, up to 25% at the largest air gap of 31.5 cm. This linear increase with gap size is consistent with the majority of additional contaminants being produced in the air gap. Nevertheless, even up to the largest air gap, the increase in skin dose due to the magnetic field is less than 13%.

III.D. RBP transverse linac-MR system

III.D.1. Skin dose in the entry region

III.D.1.a. Perpendicular beams. Figure 9(a) displays the CAX PDDs of a perpendicular (zero surface angle) $10 \times 10$ cm$^2$ photon beam incident on a $30 \times 30 \times 20$ cm$^3$ phantom with and without the 0.56 T transverse magnetic field. On the entry side both the skin dose and the first 4 mm of the depth-dose are slightly lower than the no B-field case [Fig. 9(b)]. This is due to the magnetic field purging electron contamination from the beam path. As discussed previously by Oborn et al., a field strength of 0.6 T sweeps out most of the contaminant electrons and the resultant skin dose is very close to that of a contaminant-free simulation, where only dose from photons incident on the phantom surface is scored. Also, the Lorentz force perturbation of the secondary electrons produced in the phantom entry region has minimal effect on the skin dose. Figure 9(a) further shows that there is no shift in $d_{\text{max}}$ to a shallower depth for this field strength.

The effect of the field size on the entry skin dose is shown in Fig. 10. Line profiles through the middle of square fields of sizes from $5 \times 5$ to $20 \times 20$ cm$^2$ are shown in Fig. 10(a).
Fig. 9. (a) The CAX PDD of our transverse linac-MR system in the presence of the realistic 3D $B$-field model. (b) The magnification of the first 5 mm of (a) is displayed. The data shown are for a $10 \times 10$ cm$^2$ photon beam and with the phantom surface at 136 cm from the linac source.

for the no field and 0.56 T transverse field cases, illustrating the reduction of penumbra when the magnetic field is present. The CAX entry skin dose values are plotted as a function of field size in Fig. 10(b). For the smallest $5 \times 5$ field size, the CAX skin dose is nearly the same with and without the magnetic field. As the field size is increased, both the zero field and 0.56 T field cases show an approximately linear increase in skin dose, but the 0.56 T case does so much more slowly. At a $20 \times 20$ field size, the skin dose with the magnetic field is 7.5% less than that without magnetic field. This is consistent with the fact that a larger photon beam contains more contaminant electrons scattering off the linac head that contribute to skin dose. Head-scattered electrons directed toward the patient surface generally do so at an angle with respect to the CAX, and under the influence of a transverse $B$-field they can be swept out of the central region on a helical path. Those electrons that would reach the surface parallel to the beam CAX will be deflected by the $B$-field on a circular path.

In the main $B$-field of 0.56 T, the gyration radius for the circular/helical path is $\sim 0.9$ cm for an average electron energy of $\sim 1$ MeV, and no more than 3.9 cm, corresponding to electrons with the maximum electron energy of 6 MeV. Since the extension of the uniform main $B$-field above the patient surface will in general be significantly larger than these gyration diameters (as required for imaging), the electrons will remain on these helical paths that for a flat patient surface (“zero surface angle”) will never intersect the patient surface. Hence, the purged electrons will not lead to an increased skin dose away from the field borders for this scenario (“zero surface angle”). This is supported by the lateral entry dose profiles shown in Fig. 10(a), where there is no evidence of hot spots outside the field border. (Note that the profiles are shown with tails that extend a significant distance from the field borders in comparison to the estimated gyration radii, particularly for the smaller $5 \times 5$ and $10 \times 10$ cm$^2$ field sizes.)

Fig. 10. (a) Entry skin dose line profiles through the CAX and (b) entry skin dose as a function of the field size for our transverse linac-MR system in the presence of the realistic 3D $B$-field model. The phantom surface is at 136 cm from the linac source.
geometrical difference between the fixed cylindrical (FC) and rotating biplanar transverse configurations. Figure 1 of the work by Kirkby et al.4 shows the fundamental differences between the two systems. In the FC geometry, the Lorentz force acts perpendicular to the patient axis and its direction changes as the surface orientation is varied. While, in the RBP geometry the Lorentz force is along the superior–inferior patient axis for both positive and negative surface angles, leading to the observed symmetric behavior.

### III.D.2. Skin dose in the exit region

#### III.D.2.a. Perpendicular beams. In Fig. 9(a) the CAX PDDs of the perpendicular $10 \times 10 \text{cm}^2$ photon beam show an increase in the last 1 cm depth dose and in the exit skin dose (defined in the last 70 $\mu$m voxel) in the presence of the 0.56 T transverse magnetic field. This is due to the ERE occurring on the exit side.7 The effect of the field size on the exit skin dose is shown in Fig. 12. Unlike for the entry skin dose, where the center of the fields did not shift in the presence of the transverse $B$-field, the line profiles of the exit skin dose shown in Fig. 12(a) are no longer symmetric about the CAX. This is due to the ERE shift,6, 14 which causes a displacement of the dose deposition. The spatial shifts in the center of the line profiles are 9, 8.5, 7.5, and 5 mm for $5 \times 5$, $10 \times 10$, $15 \times 15$, and $20 \times 20 \text{cm}^2$ field sizes, respectively. It is observed, however, that for fields larger than $10 \times 10 \text{cm}^2$ the increase in the exit skin dose is fairly uniform within the shifted high dose region of the field. This is not necessarily the case for a smaller field size, for example, the exit skin dose for, the $5 \times 5 \text{cm}^2$ field size at the CAX is smaller by 2.7% of $D_{\text{max}}$ than the maximum exit skin dose that occurs off CAX. It has been previously reported6, 14 that the movement of the maximum exit skin dose is primarily evident at small field sizes and low magnetic field strengths, where the average gyration radius of the returning electrons becomes comparable to half of the field size. For comparison, the gyration radii for 0.1, 0.5, 1.0, 2.0, and 6.0 MeV electrons traveling in air and perpendicular to a 0.56 T $B$-field are approximately 0.2, 0.5, 0.85, 1.5, and 3.9 cm respectively [Eq. (3)].

The magnitude of the maximum exit skin dose as a function of the field size is summarized in Fig. 12(b). In the absence of the $B$-field, the exit skin dose ranges from $\sim 28\%$ to $34\%$ of $D_{\text{max}}$, showing a slow linear increase with field size due to the larger contributions of lateral phantom scatter. With the 0.56 T $B$-field the exit skin dose is significantly higher, ranging from $\sim 56\%$ to $73\%$ of $D_{\text{max}}$. In this case the increase is not linear, with the skin dose reaching a maximum at the $15 \times 15 \text{cm}^2$ field size, and then leveling off. Oborn et al.14 suggested that a significant reduction in the exit skin dose can be achieved by adding 1 cm thick exit bolus. We would expect that the use of exit bolus would result in a similar reduction in our rotating biplanar transverse linac-MR system, though this is not investigated in this work.

#### III.D.2.b. Oblique beams. Figure 13 displays the maximum exit skin dose versus surface angle for the $10 \times 10 \text{cm}^2$ field size, with and without the 0.56 T transverse $B$-field. In the absence of the magnetic field the exit skin dose decreases

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**Medical Physics, Vol. 39, No. 10, October 2012**
FIG. 12. (a) Exit skin dose line profiles through the CAX and (b) exit skin dose as a function of the field size for our transverse linac-MR system in the presence of the realistic 3D $B$-field model. The phantom surface is at 136 cm from the linac source.

for more oblique exit angles due to a decreased contribution of the secondary electrons in the phantom to the central region. This observed trend is reversed from what is observed at the entry side (Fig. 11). When the transverse $B$-field is introduced the exit skin dose also decreases for more oblique angles, as with the 0 T case, but remains larger than the 0 T case for all surface angles simulated. Within $\pm 20^\circ$ of perpendicularity, the reduction in the skin dose is quite gradual, but at larger oblique angles an abrupt decrease is observed. Whereas at perpendicular incidences, the exit skin dose is $\sim 40\%$ (of $D_{max}$) larger with the magnetic field, at oblique angles of 45$^\circ$ the 0.56 T exit dose is only 8% larger than the exit dose at 0 T. Similar to the entry skin dose a symmetric behavior is observed on the exit side for positive and negative surface angles. On the exit side the two competing processes are: (1) the ERE which tends to increase the skin dose, and (2) a reduction in the secondary electrons generated in the phantom and above the exit layer, for more oblique surfaces. For the surface angles simulated the first effect always dominates, but to a lesser extent as the magnitude of the angle of obliquity increases (particularly above angles of 20$^\circ$).

Similar to the entry side, our exit skin dose results for positive surface angles are consistent with previous works by Raaijmakers et al.\textsuperscript{6,12} and Oborn et al.,\textsuperscript{14} but differ from their results for negative surface angles. As opposed to the symmetric behavior we observed for our system the exit skin dose in those works tends to increase for negative surface angles up to about $-30^\circ$ and gradually drops beyond that. This difference is again due to the Lorentz force acting perpendicular to the patient axis in the fixed cylindrical geometry and acting in parallel with the patient axis in our rotating biplanar system.

FIG. 13. The CAX exit skin dose of the transverse linac-MR system as a function of the surface angle (nonperpendicular beams), for a 10 $\times$ 10 cm$^2$ photon beam. The surface of a 20 cm thick phantom was placed at a 136 cm distance from the linac target (i.e., isocenter at 10 cm depth).

IV. CONCLUSION

In this work we used Monte Carlo simulations to accurately predict skin dose changes in 6 MV RBP longitudinal and transverse linac-MR systems. By using FEM we accurately modeled realistic 3D magnetic field maps of linac-MR systems that correctly predict the rapid decay of the fringe fields.

For the longitudinal geometry, simulations of the skin dose in the presence of generic Helmholtz coil systems prove the necessity of incorporating a yoke into the linac-MR model. The EGSnrc Monte Carlo simulations predict that realistic 3D longitudinal fields mostly confine contaminant electrons that are generated in the air gap in the central region, where the main magnetic field exists. The fringe fields are too small to cause appreciable confinement of electrons produced in either the air column or the linac head. Using realistic 3D MRI fields, longitudinal linac-MR systems will result in only a small increase in entrance skin dose. This is contrary to a previous report,\textsuperscript{15} which relied on the use of a variety of hypothetical 1D forms for the magnetic field maps. The increase
in the entrance skin dose of longitudinal linac-MRIs remains below 15% of $D_{\text{max}}$ for all the simulated scenarios.

For the transverse geometry, the CAX entry skin dose is equal or smaller than that of the zero $B$-field case for perpendicular beams. When a $10 \times 10 \text{ cm}^2$ field is obliquely incident on a phantom, the presence of the magnetic field increases the CAX entry skin dose by no more than 10% of $D_{\text{max}}$ when the magnitude of the angle of obliquity is $\leq 45^\circ$, and in fact decreases the entry skin dose for angles $\leq 20^\circ$. On the exit side, the simulations suggest that the magnetic field does increase the skin dose by up to 42% of $D_{\text{max}}$ compared to the zero $B$-field case. The increase in exit skin dose is largest for perpendicular beams, but appreciably drops and approaches the zero $B$-field case for large oblique angles of incidence. The magnitude of the exit skin dose for our transverse RBP system is comparable to what has been previously reported by Oborn et al. for a cylindrical transverse geometry. Since in their work a 1 cm thick exit bolus was found to be a potential solution to significantly reduce the exit dose, we expect that a similar approach will be suitable for the transverse RBP linac-MR system.

Based on our results, we do not expect the changes in skin doses caused by the magnetic field of our RBP linac-MR to be a limitation in the use of the system in either the longitudinal or transverse geometric configurations.

ACKNOWLEDGMENTS

A. Keyvanloo is supported by the Endowed Graduate Scholarship in Oncology at the University of Alberta. T. Tadic is supported by the Alberta Innovates-Health Solutions. The authors gratefully acknowledge the Alberta Cancer Foundation for financial support and computing resources provided by WestGrid and Compute/Calcul Canada.

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Medical Physics, Vol. 39, No. 10, October 2012