

Compact dose delivery of laser-accelerated high-energy electron beams toward radiotherapy applications

Bing Zhou^{1,2}, Zhiyuan Guo², Yang Wan^{1,3,*}, Shuang Liu,²
Bo Peng,¹ Jianfei Hua², and Wei Lu^{4,2,†}

¹Laboratory of Zhongyuan Light, School of Physics, Zhengzhou University, Zhengzhou 450001, China

²Department of Engineering Physics, Tsinghua University, Beijing, China

³Beijing Academy of Artificial Intelligence, Beijing, China

⁴Institute of High Energy Physics, Chinese Academy of Sciences, Beijing 100049, China



(Received 3 March 2025; accepted 3 September 2025; published 27 October 2025)

The use of very high-energy electron (VHEE) beams for radiotherapy has been actively studied for over two decades due to their advantageous dose distribution, deep penetration depth, and great potential of ultrahigh dose-rate irradiation. Recently, laser-plasma wakefield accelerator (LWFA) has emerged as a promising method for the compact generation of VHEE beams, due to its substantially higher accelerating gradients compared to traditional radio-frequency accelerators. However, how to compactly deliver the LWFA-based VHEE beams of relatively large energy spread and create a maximum dose deeply inside the body remains very challenging. In this article, we present a simple dose delivery scheme utilizing only two dipole magnets for LWFA-based VHEE treatment. By adjusting the magnet strengths, the electron beams can be guided along different angular trajectories toward a precise position as deep as 20 cm within a water phantom, creating a maximum dose over the target region and significantly reducing the entrance dose. Supported by Monte Carlo simulations, such a beam delivery approach is demonstrated to be insensitive to the beam energy spread and meanwhile capable of controlling precisely the dose-peak position in both lateral and longitudinal directions. As such, a uniform dose peak can be generated by the weighted sum of VHEE beams that reach different dose-peak depths. These results demonstrate that LWFA-based VHEE beams can be compactly delivered into a deep-seated tumor region in a controllable manner, thus advancing the development of the VHEE radiotherapy toward the practical clinical applications in the near future.

DOI: [10.1103/xdsm-7xmf](https://doi.org/10.1103/xdsm-7xmf)

I. INTRODUCTION

Radiotherapy is widely applied in oncology by employing ionizing radiation to either suppress or control the growth of tumor tissue. Currently, three radiation sources are mainly used in radiotherapy: photons, ions, and low-energy electrons (less than 20 MeV). Among them, photon therapy offers significant penetration depth but results in high dose deposition in normal tissues. To reduce this risk, high conformal techniques such as intensity-modulated radiation therapy [1] and volumetric modulated arc therapy [2,3] were introduced. Hadron therapy, on the other hand, leverages the Bragg peak effect of protons and heavy ions, which allows for substantial energy deposition at the range's end while reducing the entrance dose. However,

hadron treatment is quite sensitive to biological matter inhomogeneities, with minor disturbances causing shifts in Bragg peak localization [4] and also requires substantial infrastructure and high costs. Conventional electron beam therapy primarily employs low-energy electron beams, which are suitable for treating surface or shallow tumors because of their high entrance dose and the rapid decrease in dose with depth.

Recently, very high-energy electron beams (50–300 MeV) have been proposed as an alternative candidate for radiotherapy [5,6]. Compared to low-energy electrons, VHEE can penetrate much deeper region with sharper lateral penumbra profiles [5]. Meanwhile, several researches have shown that they exhibit superior therapeutic efficacy to photon modalities for several different tumors [7–11] and less sensitive to inhomogeneities than ion beams [5,12]. However, the accelerating gradient of conventional radio-frequency (rf) accelerators is generally much less than 100 MeV/m, which necessitates a linear accelerator of at least tens of meters in length to produce such very high-energy electron beams, making them difficult to be adapted into a common treatment room [13].

*Contact author: yangwan23@zzu.edu.cn

†Contact author: weilu@ihp.ac.cn

Published by the American Physical Society under the terms of the [Creative Commons Attribution 4.0 International license](https://creativecommons.org/licenses/by/4.0/). Further distribution of this work must maintain attribution to the author(s) and the published article's title, journal citation, and DOI.

Over the past two decades, strong plasma wave driven by ultrashort, ultraintense lasers for accelerating electron beams, known as laser wakefield acceleration (LWFA) [14,15], has attracted great interest worldwide, due to its ultrahigh accelerating gradient of above 100 GeV/m, more than 1000 times stronger than the rf accelerators. With the rapid development of LWFA technology, this new kind of accelerator is now capable of stably delivering VHEE beams of several hundreds of MeV [16–19] with small source sizes (few microns scale) [20,21], low divergences (few milliradians scale) [22], and ultrashort durations (few femtoseconds level) [23,24] within only a few millimeters distance, thereby making VHEE radiotherapy platforms much more compact and economically viable [25,26]. Recently, several research groups have focused on the studies of LWFA-based VHEE radiotherapy, including the measurements of dose distribution [27–29], treatment planning [8], and irradiating mice tumors with a prototype LWFA machine [30]. However, how to transport such a VHEE beam generated from LWFA to target the deeply-seated tumor precisely remains challenging due to the following two aspects. First, VHEEs normally cannot create a dose peak at the tumor target similar as ion beams, but exhibit high dose depositions at both entry and exit points, which could possibly damage the surface skin and other healthy issues [5]. Additionally, VHEEs from laser wakefield accelerators normally have a notable rms energy spread of over 5% if ensuring high charge (more than 100 pC) [15], significantly complicating beam transport modalities.

Previous studies have explored the use of quadrupole magnets to focus the VHEE beams at a specific depth within a water phantom [27,29,31–34]. However, quadrupole magnets are not suitable for beams with large energy dispersion, such as LWFA-based ones. Moreover, current experimental findings do not support the deposition of peak dose beyond a depth of 6 cm in water phantoms [32,34]. In this article, we propose a simple approach to effectively deliver the peak dose of VHEEs into the tumor target by

employing only two dipole magnets. These magnets can guide the electron beams from different directions to achieve a maximum dose at a depth of up to 20 cm inside the water phantom. Monte Carlo simulations also demonstrate that this technique exhibits exceptional resistance to beam energy spread, particularly attractive for LWFA-based applications. Additionally, it allows precise control of the dose-peak position both laterally and longitudinally. These advancements afford promises for the development of a compact VHEE radiotherapy machine with high precision.

II. RESULTS AND DISCUSSIONS

A. The basic concept of the proposed VHEE dose delivery scheme

The basic layout of the proposed scheme is shown in Fig. 1(a), where two rectangular dipoles with strengths B_1 and B_2 (having opposite signs) and lengths S_1 and S_2 , respectively, are separated by a distance d , with the condition $|B_1 S_1| < |B_2 S_2|$. We consider an electron beam with initial longitudinal momentum p_z and zero divergence. For simplicity, we define the beam's initial propagation axis as the z axis, the direction of the magnetic fields as the x axis, and the other transverse direction as the y axis. In this configuration, the beam dynamics are mainly confined to the $y-z$ plane. Since the two magnets have opposite signs, the beam is first deflected by B_1 away from the longitudinal axis and then bent back by B_2 toward the axis, leading to the following geometrical relations:

$$r_1 \sin(\theta_1) = S_1, \quad (1)$$

$$r_2 [\sin(\theta_2 - \theta_1) + \sin(\theta_1)] = S_2, \quad (2)$$

where r_1 and r_2 are the Larmor radii, and θ_1 and θ_2 are the bending angles in the magnetic fields B_1 and B_2 , respectively. The distance between the point where the beam

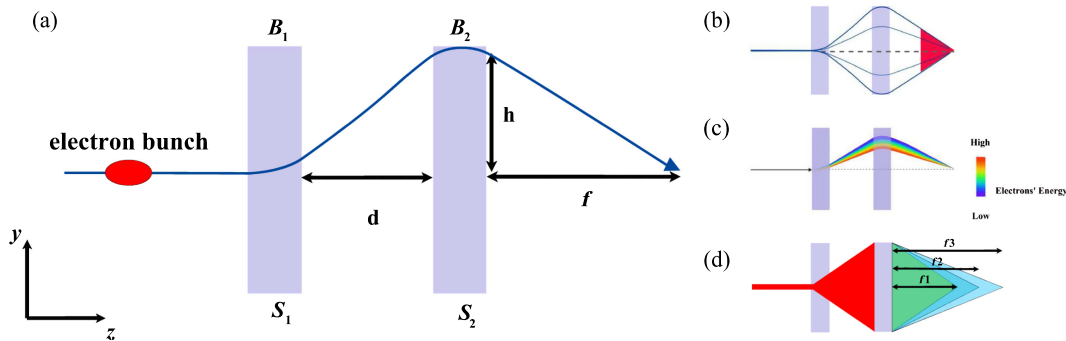


FIG. 1. Schematic illustration of the proposed scheme. (a) The basic two-dipole layout and a typical electron beam trajectory. (b) A maximum dose at the target position is created by guiding electron beams through different trajectories via scanning field strengths of both dipoles. (c) An electron beam with notable energy spread can be guided to target the same position. (d) The dose-peak depth is controlled by the strength ratio k .

crosses the longitudinal axis and the rear side of the second magnet is denoted as f and can be expressed as

$$f = \frac{1}{\tan(\theta_2 - \theta_1)} \cdot [r_1(1 - \cos(\theta_1)) - r_2(\cos(\theta_2 - \theta_1) - \cos(\theta_1)) + d \tan(\theta_1)] \quad (3)$$

In our proposed scheme, the maximum bending angles are approximately 220 mrad for θ_1 and 400 mrad for θ_2 in dipoles B_1 and B_2 , respectively. Since these angles are small, $\cos(\theta_1)$, $\cos(\theta_2)$, and $\cos(\theta_2 - \theta_1)$ are very close to 1, allowing us to simplify Eq. (3) as

$$f \approx d \cdot \frac{\sin \theta_1}{\sin(\theta_2 - \theta_1)}. \quad (4)$$

Then, dividing Eq. (1) by Eq. (2), one can get $\sin(\theta_2 - \theta_1)/\sin \theta_1 = (S_2 r_1)/(S_1 r_2) - 1$. Using the relation $r_{1,2} = -p_z/eB_{1,2}$, Eq. (4) can be rewritten as

$$f \approx \frac{d}{(S_2 B_2)/(S_1 B_1) - 1}. \quad (5)$$

Through Eq. (5), it is evident that for fixed magnetic lengths, the value of parameter f is uniquely determined by the ratio k of the strengths of two magnets, represented as $k = \frac{B_2}{B_1}$. This proposed scheme offers the following three advantages in dose delivery.

First, by scanning B_1 (B_2) from the positive (negative) maximum to the negative (positive) maximum while keeping their ratio k constant, an electron bunch with equivalent energy can be guided toward the same position but along different trajectories. By aggregating all these cases, a concentrated maximum dose can be achieved near the converging position where the tumor is located, as shown in Fig. 1(b).

Second, Eq. (5) is independent of the beam energy. Consequently, electrons with various energies passing through the system will ultimately converge at the same position. This characteristic implies that the system is resilient to beam's energy spread, which distinguishes it from other proposed schemes [27,31,34] [see Fig. 1(c)].

Finally, the beam converging position can be readily adjusted by modifying the strength relations of the two dipoles, providing exceptional flexibility for targeting tumors at different locations. Figure 1(d) shows three cases of different dose-peak depths by varying the ratio k .

Note that the above analytical model is still quite simplified. Considering a realistic and finite magnetic field distribution, VHEE beams will not be guided to converge to a single point but spread into a spot of finite size. However, this effect is significantly smaller than the multiple Coulomb scattering caused by the water medium when propagating deeply inside the phantom (i.e., 10–20 cm).

Therefore, the three advantages previously mentioned remain valid. A detailed analysis of this point can be found in Sec. II and Fig. S2 of the Supplementary Materials [35].

B. Validation using Monte Carlo simulations

In order to validate the proposed scheme, we have performed Monte Carlo simulations using the code TOPAS [36]. The simulation geometry was the same as the above analytical model. The lengths of the two dipoles were set at $S_1 = 10$ cm and $S_2 = 22$ cm, with a gap size of 2 cm. The two magnets were separated by 30 cm. The magnetic field distributions, including the fringe fields, for both dipoles were calculated using three-dimensional (3D) electromagnetic simulation software CST (more detailed field distributions can be found in Sec. I and Figs. S0–S1 of the Supplementary Materials [35]). The maximum strengths of the two magnets were chosen as 1.35 T, with B_1 and B_2 having opposite signs. A water phantom was positioned 30 cm behind the second magnet, with dimensions of 20 cm (x) $\times$ 20 cm (y) $\times$ 40 cm (z), and a voxel size of 1 mm in all directions. The entrance of the water phantom was defined as $z = 0$. Considering a typical electron bunch originating from a standard LWFA, we set its initial distribution in both space and divergence space as Gaussian distribution, with rms sizes as $\sigma_x = \sigma_y = 4$ μ m, rms divergences as $\theta_x = \theta_y = 4$ mrad, and normalized emittance ϵ_n of around 6.3 mm mrad, respectively. The bunch contained a peak energy of 200 MeV and a 5% rms energy spread and was launched 15 cm ahead of the front surface of the first dipole. Based on this configuration, the total length of the beam delivery system from the source to the water phantom entrance was around 1 m.

To achieve a maximum dose at a specific depth deeply inside the water phantom, we launched 11 simulations. In each simulation, an electron bunch containing 10^6 particles was used and only the value of B_1 (B_2) was scanned from the negative (positive) maximum to the positive (negative) maximum with a constant incremental step. After summing up all above cases, the resulting dose distribution is presented in Fig. 2.

Figures 2(a) and 2(b) present the central slice dose distribution in the $z-x$ and $z-y$ planes, respectively, where one can see the appearance of a maximum dose deposited deeply inside the water phantom. To further verify this finding, we present the longitudinal on-axis dose lineout in Fig. 2(c). The dose peak is observed located at a depth of 15 cm, with the entrance dose only accounting for around 33% of the peak dose. Furthermore, Fig. 2(d) displays the transverse ($x-y$) dose distribution at the dose-peak depth, where the full width at half maximums (FWHMs) along the two dimensions were measured as $\text{FWHM}_y = 1.8$ cm and $\text{FWHM}_x = 1.4$ cm, respectively. These values are comparable to or even smaller than the dose size of a typical proton pencil beam at the same depth, suggesting that the combination of VHEE and this dose delivery system can greatly

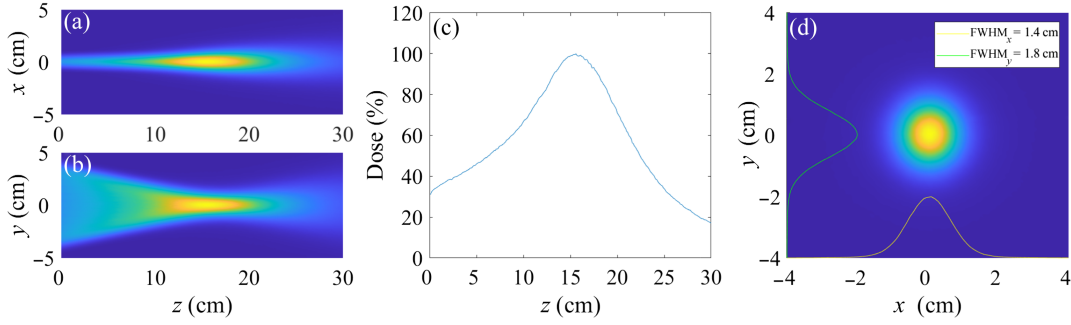


FIG. 2. Monte Carlo simulation results of VHEE dose deposition in a water phantom using the proposed scheme. (a), (b) 2D dose distribution along $(x - z)/(y - z)$ plane, respectively. (c) On-axis dose distribution along the z axis, where an in-depth dose peak at 15 cm can be observed. (d) Transverse dose distribution at the $z = 15$ cm plane, with FWHM sizes of 1.4 and 1.8 cm in x and y directions, respectively.

benefit accurate tumor targeting. In our case, eleven is the minimal number of the above scanning fractions for guaranteeing a relatively uniform dose coverage before the dose peak, minimizing healthy tissue exposure. By increasing the scanning numbers, the dose coverage could be more uniform while the lateral sizes of dose peak are almost unaffected, as verified by a series of Monte Carlo simulations.

To evaluate the system's tolerance to the bunch energy spread, we conducted four simulations with the same setup as previously described, but applying electron beams with different energy distributions. The results are presented in Figs. 3(a)–3(d). For simulations (a)–(c), the energy spectra adhered to a Gaussian distribution with a central energy of 200 MeV and a rms spread of 0.1%, 5%, and 20%, respectively. In simulation (d), the energy spectrum was set uniform, ranging from 100 to 300 MeV. Despite significant disparities in energy spread, the system is shown to be capable of concentrating the dose peak with similar longitudinal and lateral dose distributions. Figure 3(e) provides a more detailed on-axis dose distribution along the z axis for these four cases. The first three cases exhibit negligible differences, with only the fourth case demonstrating minor variation of less than 5%, which confirms the

robustness of the proposed scheme. We note that a dose peak may also be achieved by the asymmetric focusing using a series of quadrupoles [31]. However, to achieve a similar dose-peak depth (e.g., ~ 15 cm) with small entrance-to-peak dose ratio (e.g., $\sim 30\%$), the required parameters for quadrupoles are much more challenging, and they cannot transport electrons with large energy dispersion as effectively as the presented scheme.

C. Controlling the dose-peak position

To ensure a precise and flexible dose delivery, a reliable system is necessary to enable the control over the position of dose deposition for targeting different types of tumors. Here we show that by simply modifying the two dipole field strengths, this requirement can be fulfilled. We consider a simple relation between them as $B_2 = kB_1 + b$, where k and b are controllable variables. This formula allows for the longitudinal (transverse) adjustment of the dose-peak position by varying k (b). A series of Monte Carlo simulations have been carried out to confirm this approach with the same setup as Fig. 2 but different magnet strengths, and the results are presented in Fig. 4. As shown in Figs. 4(a) and 4(b), the

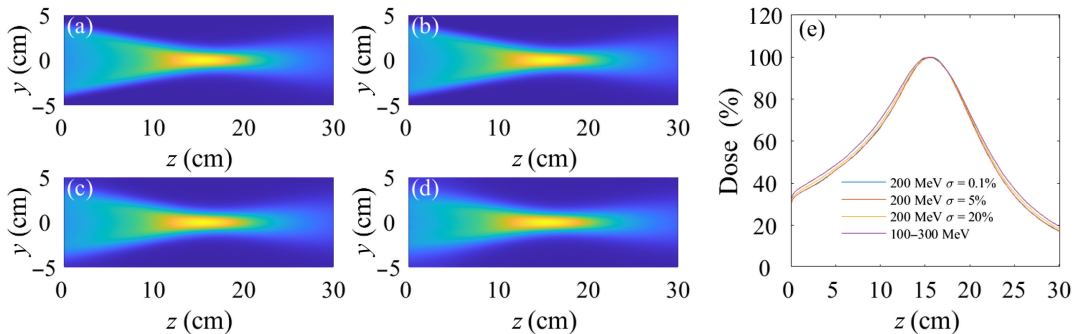


FIG. 3. Relative dose depositions in a water phantom of VHEE beams with different energy spreads. (a)–(d) 2D dose distribution in the $y - z$ plane by an VHEE beam with 0.1%, 5%, 20% rms relative energy spread. (d) 2D dose distribution by an electron beam with a flat energy spectrum ranging from 100 to 300 MeV. (e) On-axis dose distribution of all four cases. The mean energy of the electron beams for (a)–(d) is 200 MeV.

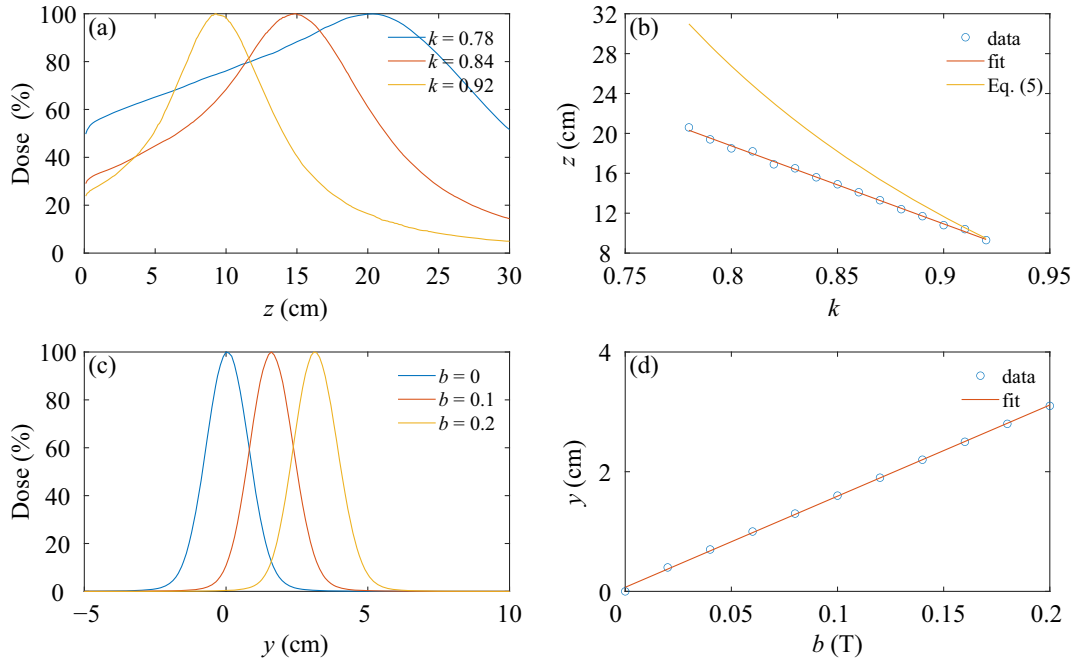


FIG. 4. Dose-peak position control along longitudinal and transverse directions using the proposed scheme. (a) Relative longitudinal on-axis dose distribution with different parameter k . (b) Relationship between the depth of the dose-peak position and parameter k , where the solid yellow curve represents the z position predicted by Eq. (5). (c) Transverse dose deposition lineout at $z = 15$ cm in the $x - z$ plane with $b = 0$ T (blue), $b = 0.1$ T (red), and $b = 0.2$ T (yellow). (d) Dependence of lateral displacement of the dose-peak position on parameter b . For (b), (d), the blue circles and red curve represent simulation results and linear regression, respectively.

dose-peak depth inside the water increases from 7 to 20 cm as k decreases from 0.95 to 0.78, and the correlation between them is almost linear, with the fitted equation expressed as: $z(\text{cm}) = -78.2(\pm 1.06)k + 81.3(\pm 0.9)$ yielding $R^2 = 0.996$ and $\chi^2 = 0.260$. Two things need to be noted here. First, this increase in dose-peak depth may inevitably result in higher entrance doses owing to enhanced scattering of electron beams when propagating into deeper water [see Fig. 4(a)]. Second, by comparing the blue circles and yellow curve in Fig. 4(b), it is clear that the strong water scattering also causes a discrepancy between Monte Carlo simulations and predicted values by Eq. (5), indicating that the simplified analytical model becomes less accurate in deeper regions.

Figure 4(c) illustrates the transverse dose lineouts along the x axis at the dose-peak depth (15 cm) for different b values. By setting b from 0 to 0.2 T, the dose-peak position can be laterally displaced by 3 cm in the corresponding directions along the x axis. As shown in Fig. 4(d), the transverse displacement is also in relatively linear dependence of the b values with a fitting equation expressed as: $y(\text{cm}) = 15.22(\pm 0.13)b(\text{T}) + 0.06(\pm 0.01)$ yielding $R^2 = 0.999$ and $\chi^2 = 0.066$. It is noted that during the transverse scanning (only varying b), the longitudinal dose-peak depth can also be shifted slightly. In our case, a lateral displacement of 3 cm results in an approximate 1 cm backward shift of the dose-peak depth. Nevertheless, this shift can be compensated by adjusting the parameter k .

We also note that chromatic aberration due to large energy spread could potentially affect the lateral size of the dose-peak area during the transverse adjustment of the dose-peak position. To address this concern, we conducted two sets of Monte Carlo simulations with the dose-peak transversely displaced by around 3 cm. Our findings indicate that the impact of chromatic aberration on the enlargement of the dose-peak size is insignificant compared to the contributions from the original divergence of LWFA beams (a few mrad) and multiple Coulomb scattering in water or human tissue. Detailed results are presented in Sec. III and Figs. S3 and S4 of the Supplementary Materials [35].

To enable the dose-peak control in both transverse directions, the two dipole magnets (weighing less than 200 kg in total) can be mounted together onto a gantry and rotated around the axis of the incoming beam. This configuration allows for a full targeting range covering a circular area with a diameter of 6 cm, sufficient for treating small to medium-sized tumors. Additionally, there are several ways to further increase the cover range. For instance, since the LWFA itself is only a few millimeters in size, we can easily displace the entire LWFA module (including the laser focusing optics) transversely by several centimeters. Throughout the dose-peak control process, the VHEE beam can be monitored in real time by placing a very thin scintillator screen at the treatment head (downstream of

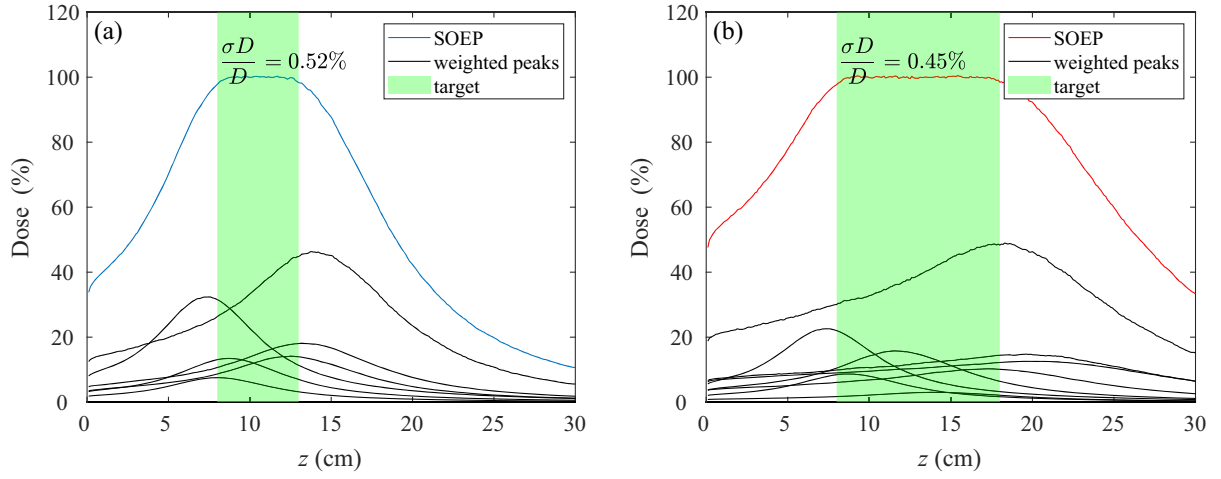


FIG. 5. Simulation results of achieving spread-out electron peak (SOEP) with the proposed scheme. (a) Accumulating six weighted electron dose profiles results in a flat region with a length of 5 cm at a depth ranging between 8 and 13 cm and (b) accumulating eight weighted electron dose profiles results in a flat region with a length of 10 cm at a depth ranging between 8 and 18 cm.

the second dipole). This allows for luminescence monitoring without significantly affecting the beam dynamics.

D. Flattop dose-peak formation

Similar to the commonly used spread-out Bragg peak technique in proton therapy, which broadens the Bragg peak by linearly superposing different proton energies, we can also generate a spread-out electron peak (SOEP) by summing up doses of different dose-peak depths assigned with specific weights. This concept was introduced in [31], and in our case, it can be accomplished by simply modifying the value of the parameter k . Figure 5 illustrates two simulation examples of the SOEP for a 200 MeV VHEE beam. In Fig. 5(a), a dose plateau extending from a depth of 8 to 13 cm is presented, and Fig. 5(b) depicts a plateau spanning from a depth of 8 to 18 cm. The shaded green region within each figure denotes the desired depth range, where the achieved dose uniformity is quite good, with the relative standard deviations of $\frac{\sigma D_{\text{on axis}}}{D_{\text{on axis}}} = 0.52\%$ in Fig. 5(a) and $\frac{\sigma D_{\text{on axis}}}{D_{\text{on axis}}} = 0.45\%$ in Fig. 5(b). Additionally, the gray curves in each figure represent the on-axis doses of different dose-peak depths after applying different weights.

III. CONCLUSION

In this article, we propose a simple dose delivery scheme that utilizes only two dipole magnets to effectively target deep-seated tumors with a concentrated dose peak in a highly compact setup. By adjusting the correlations between these two magnets, we can precisely position the dose peak along both longitudinal and lateral directions. An additional advantage of this system is its resilience to broadband energy spectra, which makes it particularly appealing for the use in LWFA-based VHEE radiotherapy.

To fulfill the therapeutic requirements, the scanning speed of the dipoles has to be precisely controlled and synchronized with the LWFA module. Data from the electromagnet manufacturer indicates that the scanning frequency can typically attain up to 200 Hz, far exceeding the current LWFA repetition rate of below 10 Hz. Furthermore, as the dose per shot is quite small, magnet strengths can be adjusted after multiple shots, reducing the required scanning frequency to around 1 Hz.

We also note that it is crucial to conduct a thorough analysis of how variations in the VHEE beam's energy, divergence, incidence angle, and magnetic strengths affect the final dose deposition. Such an analysis is essential to further validate the robustness and feasibility of the proposed scheme. For instance, we have assessed the effect of an angular deviation in the beam's incidence angle relative to the first magnet's entrance. Through two Monte Carlo simulations, each with an initial angular deviation of 10 mrad in the x and y directions, respectively, we observed only a minor shift in the dose distribution along the corresponding axes. This deviation could be corrected using the real-time beam monitoring. Nevertheless, a comprehensive error analysis will be conducted during the subsequent engineering phase and documented in a separate publication.

In the future, we plan to incorporate the laser system, LWFA module, and the proposed dose delivery system into a small rotational gantry, equipped with advanced diagnostic tools such as femtosecond high-energy electron probe [21,37,38], advanced plasma shaping tools [39,40], and machine learning functions [41,42], which will optimize the LWFA performance and enable a compact dose delivery with large incidence angle, precise tumor conformity, and high dose-peak control flexibility, crucial for the development of LWFA-based VHEE radiotherapy devices toward clinical applications.

ACKNOWLEDGMENTS

This work was supported by the Strategic Priority Research Program of the Chinese Academy of Sciences (Grant No. XDB0530000), National Natural Science Foundation of China (No. 12574380 and No. 12405169), Science Fund Program for Distinguished Young Scholars of the National Natural Science Foundation of China (Overseas), Discipline Construction Foundation of “Double World-class Project,” Key Scientific Research Projects of Henan Provincial Colleges and Universities No. 25ZX002, and Natural Science Foundation of Henan Province No. 252300421300. The simulations were performed at National Supercomputing Center in Zhengzhou and Center of High performance computing, Tsinghua University.

B. Z., Z. G., and Y. W. contributed equally to this work.

DATA AVAILABILITY

The data that support the findings of this article are openly available [43], embargo periods may apply.

- [1] S. M. Bentzen, Radiation therapy: Intensity modulated, image guided, biologically optimized and evidence based, *Radiother. Oncol.* **77**, 227 (2005).
- [2] C. Elith, S. E. Dempsey, N. Findlay, and H. M. Warren-Forward, An introduction to the intensity-modulated radiation therapy (IMRT) techniques, tomotherapy, and VMAT, *J. Med. Imaging Radiat. Sci.* **42**, 37 (2011).
- [3] K. Otto, Volumetric modulated arc therapy: IMRT in a single gantry arc, *Med. Phys.* **35**, 310 (2008).
- [4] M. Urie, M. Goitein, and M. Wagner, Compensating for heterogeneities in proton radiation therapy, *Phys. Med. Biol.* **29**, 553 (1984).
- [5] C. Desrosiers, V. Moskvina, A. F. Bielajew, and L. Papiez, 150-250 MeV electron beams in radiation therapy, *Phys. Med. Biol.* **45**, 1781 (2000).
- [6] M. G. Ronga, M. Cavallone, A. Patriarca, A. M. Leite, P. Loap, V. Favaudon, G. Créange, and L. De Marzi, Back to the future: Very high-energy electrons (VHEES) and their potential application in radiation therapy, *Cancers* **13**, 4942 (2021).
- [7] C. Yeboah and G. Sandison, Optimized treatment planning for prostate cancer comparing IMPT, VHEET and 15 MV IMXT, *Phys. Med. Biol.* **47**, 2247 (2002).
- [8] T. Fuchs, H. Szymanowski, U. Oelfke, Y. Glinec, C. Rechatin, J. Faure, and V. Malka, Treatment planning for laser-accelerated very-high energy electrons, *Phys. Med. Biol.* **54**, 3315 (2009).
- [9] M. Bazalova-Carter, B. Qu, B. Palma, B. Hårdemark, E. Hynning, C. Jensen, P. G. Maxim, and B. W. Loo Jr, Treatment planning for radiotherapy with very high-energy electron beams and comparison of VHEE and VMAT plans, *Med. Phys.* **42**, 2615 (2015).
- [10] E. Schüller, K. Eriksson, E. Hynning, S. L. Hancock, S. M. Hiniker, M. Bazalova-Carter, T. Wong, Q.-T. Le, B. W. Loo, Jr., and P. G. Maxim, Very high-energy electron (VHEE) beams in radiation therapy; treatment plan comparison between VHEE, VMAT, and PPBS, *Med. Phys.* **44**, 2544 (2017).
- [11] G. Zhang, Z. Zhang, W. Gao, and H. Quan, Treatment planning consideration for very high-energy electron flash radiotherapy, *Phys. Med.* **107**, 102539 (2023).
- [12] A. Lagzda, D. Angal-Kalinin, J. Jones, A. Aitkenhead, K. J. Kirkby, R. MacKay, M. Van Herk, W. Farabolini, S. Zeeshan, and R. M. Jones, Influence of heterogeneous media on very high energy electron (VHEE) dose penetration and a Monte Carlo-based comparison with existing radiotherapy modalities, *Nucl. Instrum. Methods Phys. Res., Sect. B* **482**, 70 (2020).
- [13] The CHUV-CERN Collaboration to design and construct a high-energy electron flash therapy facility, in *CLIC Project Meeting 40* (2021), <https://indico.cern.ch/event/1071466/contributions/4538013/attachments/2322311/3954858/CLIC%20project%20FLASH%202021-10-5.pdf#:~:text=As%20part%20of%20this%20effort%2C%20CHUV%20and%20CERN,high-energy%2C%20100%20MeV-range%2C%20electrons%20accelerated%20with%20CLIC-developed%20technology>.
- [14] T. Tajima and J. M. Dawson, Laser electron accelerator, *Phys. Rev. Lett.* **43**, 267 (1979).
- [15] E. Esarey, C. B. Schroeder, and W. P. Leemans, Physics of laser-driven plasma-based electron accelerators, *Rev. Mod. Phys.* **81**, 1229 (2009).
- [16] C. G. R. Geddes, C. Toth, J. van Tilborg, E. Esarey, C. B. Schroeder, D. Bruhwiler, C. Nieter, J. Cary, and W. P. Leemans, High-quality electron beams from a laser wake-field accelerator using plasma-channel guiding, *Nature (London)* **431**, 538 (2004).
- [17] S. P. Mangles, C. Murphy, Z. Najmudin, A. G. R. Thomas, J. Collier, A. E. Dangor, E. Divall, P. Foster, J. Gallacher, C. Hooker *et al.*, Monoenergetic beams of relativistic electrons from intense laser-plasma interactions, *Nature (London)* **431**, 535 (2004).
- [18] J. Faure, Y. Glinec, A. Pukhov, S. Kiselev, S. Gordienko, E. Lefebvre, J.-P. Rousseau, F. Burgy, and V. Malka, A laser-plasma accelerator producing monoenergetic electron beams, *Nature (London)* **431**, 541 (2004).
- [19] A. R. Maier, N. M. Delbos, T. Eichner, L. Hübner, S. Jalas, L. Jeppe, S. W. Jolly, M. Kirchen, V. Leroux, P. Messner, M. Schnepf, M. Trunk, P. A. Walker, C. Werle, and P. Winkler, Decoding sources of energy variability in a laser-plasma accelerator, *Phys. Rev. X* **10**, 031039 (2020).
- [20] R. Weingartner, S. Raith, A. Popp, S. Chou, J. Wenz, K. Khrennikov, M. Heigoldt, A. R. Maier, N. Kajumba, M. Fuchs *et al.*, Ultralow emittance electron beams from a laser-wakefield accelerator, *Phys. Rev. ST Accel. Beams* **15**, 111302 (2012).
- [21] Y. Wan, S. Tata, O. Seemann, E. Y. Levine, S. Smartsev, E. Kroupp, and V. Malka, Femtosecond electron microscopy of relativistic electron bunches, *Light Sci. Appl.* **12**, 116 (2023).
- [22] V. Malka, S. Fritzler, E. Lefebvre, M.-M. Aleonard, F. Burgy, J.-P. Chambaret, J.-F. Chemin, K. Krushelnick,

- G. Malka, S. Mangles *et al.*, Electron acceleration by a wake field forced by an intense ultrashort laser pulse, *Science* **298**, 1596 (2002).
- [23] O. Lundh, J. Lim, C. Rechatin, L. Ammoura, A. Ben-Ismaïl, X. Davoine, G. Gallot, J.-P. Goddet, E. Lefebvre, V. Malka *et al.*, Few femtosecond, few kiloampere electron bunch produced by a laser-plasma accelerator, *Nat. Phys.* **7**, 219 (2011).
- [24] C. Zhang, J. Hua, Y. Wan, B. Guo, C.-H. Pai, Y. Wu, F. Li, H.-H. Chu, Y. Gu, W. Mori *et al.*, Temporal characterization of ultrashort linearly chirped electron bunches generated from a laser wakefield accelerator, *Phys. Rev. Accel. Beams* **19**, 062802 (2016).
- [25] V. Malka, J. Faure, Y. A. Gauduel, E. Lefebvre, A. Rousse, and K. T. Phuoc, Principles and applications of compact laser-plasma accelerators, *Nat. Phys.* **4**, 447 (2008).
- [26] C. Joshi, S. Corde, and W. Mori, Perspectives on the generation of electron beams from plasma-based accelerators and their near and long term applications, *Phys. Plasmas* **27**, 070602 (2020).
- [27] Y. Glinec, J. Faure, V. Malka, T. Fuchs, H. Szymanowski, and U. Oelfke, Radiotherapy with laser-plasma accelerators: Monte Carlo simulation of dose deposited by an experimental quasimonoenergetic electron beam, *Med. Phys.* **33**, 155 (2006).
- [28] O. Lundh, C. Rechatin, J. Faure, A. Ben-Ismaïl, J. Lim, C. De Wagter, W. De Neve, and V. Malka, Comparison of measured with calculated dose distribution from a 120-MeV electron beam from a laser-plasma accelerator, *Med. Phys.* **39**, 3501 (2012).
- [29] K. Svendsen, D. Guénot, J. B. Svensson, K. Petersson, A. Persson, and O. Lundh, A focused very high energy electron beam for fractionated stereotactic radiotherapy, *Sci. Rep.* **11**, 5844 (2021).
- [30] Z. Guo, S. Liu, B. Zhou, J. Liu, H. Wang, Y. Pi, X. Wang, Y. Mo, B. Guo, J. Hua *et al.*, Preclinical tumor control with a laser-accelerated high-energy electron radiotherapy prototype, *Nat. Commun.* **16**, 1895 (2025).
- [31] L. Whitmore, R. I. Mackay, M. Van Herk, J. Jones, and R. Jones, Focused VHEE (very high energy electron) beams and dose delivery for radiotherapy applications, *Sci. Rep.* **11**, 14013 (2021).
- [32] L. Whitmore, R. I. Mackay, M. V. Herk, P. Korysko, W. Farabolini, A. Malyzhenkov, R. Corsini, and R. M. Jones, CERN-based experiments and Monte-Carlo studies on focused dose delivery with very high energy electron (VHEE) beams for radiotherapy applications, *Sci. Rep.* **14**, 11120 (2024).
- [33] F. Reaz, K. N. Sjobak, E. Malinen, N. F. J. Edin, and E. Adli, Sharp dose profiles for high precision proton therapy using strongly focused proton beams, *Sci. Rep.* **12**, 18919 (2022).
- [34] K. Kokurewicz, E. Brunetti, A. Curcio, D. Gamba, L. Garolfi, A. Gilardi, E. Senes, K. N. Sjobak, W. Farabolini, R. Corsini *et al.*, An experimental study of focused very high energy electron beams for radiotherapy, *Commun. Phys.* **4**, 33 (2021).
- [35] See Supplemental Material at <http://link.aps.org/supplemental/10.1103/xdsm-7xmf> for the CST-simulated magnetic field distribution, trajectories of the VHEE particles along the dose delivery system and the effects of VHEE energy spread on the dose peak position and size.
- [36] J. Perl, J. Shin, J. Schümann, B. Faddegon, and H. Paganetti, TOPAS: An innovative proton Monte Carlo platform for research and clinical applications, *Med. Phys.* **39**, 6818 (2012).
- [37] Y. Wan, O. Seemann, S. Tata, I. A. Andriyash, S. Smartsev, E. Kroupp, and V. Malka, Direct observation of relativistic broken plasma waves, *Nat. Phys.* **18**, 1186 (2022).
- [38] Y. Wan, S. Tata, O. Seemann, E. Y. Levine, E. Kroupp, and V. Malka, Real-time visualization of the laser-plasma wakefield dynamics, *Sci. Adv.* **10**, eadj3595 (2024).
- [39] O. Seemann, Y. Wan, S. Tata, E. Kroupp, and V. Malka, Refractive plasma optics for relativistic laser beams, *Nat. Commun.* **14**, 3296 (2023).
- [40] O. Seemann, Y. Wan, S. Tata, E. Kroupp, and V. Malka, Laser proton acceleration from a near-critical imploding gas target, *Phys. Rev. Lett.* **133**, 025001 (2024).
- [41] S. Jalas, M. Kirchen, P. Messner, P. Winkler, L. Hübner, J. Dirkwinkel, M. Schnepf, R. Lehe, and A. R. Maier, Bayesian optimization of a laser-plasma accelerator, *Phys. Rev. Lett.* **126**, 104801 (2021).
- [42] F. Irshad, C. Eberle, F. M. Foerster, K. v. Grafenstein, F. Haberstroh, E. Travac, N. Weisse, S. Karsch, and A. Döpp, Pareto optimization and tuning of a laser wakefield accelerator, *Phys. Rev. Lett.* **133**, 085001 (2024).
- [43] B. Zhou, Y. Wan, and W. Lu, Zenodo (2025), [10.5281/zenodo.17148994](https://zenodo.org/record/17148994).