Target tracking using DMLC for volumetric modulated arc therapy: A simulation study

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Purpose: Target tracking using dynamic multileaf collimator (DMLC) is a promising approach for intrafraction motion management in radiation therapy. The purpose of this work is to develop a DMLC tracking algorithm capable of delivering volumetric-modulated arc therapy (VMAT) to the targets that experience two-dimensional (2D) rigid motion in the beam's eye view.

Methods: The problem of VMAT delivery to moving targets is formulated as a control problem with constraints. The relationships between gantry speed, gantry acceleration, MLC leaf-velocity, dose rate, and target motion are derived. An iterative search algorithm is developed to find numerical solutions for efficient delivery of a specific VMAT plan to the moving target using 2D DMLC tracking. The delivery of five VMAT lung plans is simulated. The planned and delivered fluence maps in the target-reference frame are calculated and compared.

Results: The simulation demonstrates that the 2D tracking algorithm is capable of delivering the VMAT plan to a moving target fast and accurately without violating the machine constraints and the integrity of the treatment plan. The average delivery time is only 29 s longer than that of no-tracking delivery, 101 versus 72 s, respectively. The fluence maps are normalized to 200 MU and the average root-mean-square error between the desired and the delivered fluence is 2.1 MU, compared to 14.8 MU for no-tracking and 3.6 MU for one-dimensional tracking.

Conclusions: A locally optimal MLC tracking algorithm for VMAT delivery is proposed, aiming at shortest delivery time while maintaining treatment plan invariant. The inconsequential increase of treatment time due to DMLC tracking is clinically desirable, which makes VMAT with DMLC tracking attractive in treating moving tumors. © 2010 American Association of Physicists in Medicine. [DOI: [10.1118/1.3511516](https://doi.org/10.1118/1.3511516)]

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I. INTRODUCTION

Internal respiratory organ motion is one of the main concerns for lung and upper-abdominal cancer radiation therapy treatments.¹ The classical approach to manage such intrafraction motion is to add a large margin to ensure that the moving target is covered at all times throughout the delivery. However, this approach increases the volume of healthy tissue that receives significant doses, which is undesirable particularly when the target is located close to an organ at risk.² One solution without this burden is to gate the radiation beam within a specified time/phase window so that dose is delivered only during a certain part of the respiratory cycle.^{3,4} However, respiratory gating significantly increases the treatment time, thereby compromising patient comfort and consequently increasing likelihood of patient body movement.^{1,5} The third and a distinctive approach is real time tumor tracking where the radiation beam is automatically repositioned to follow the tumor's changing position.

Recent studies^{6–14} have shown that this technique provides a promising solution to the tumor motion problem by dynamically adapting multileaf collimator (MLC)-shaped beam aperture to track the target displacement. This method can, under ideal conditions, eliminate the need for a large tumor motion margin while maintaining 100% efficiency in delivery. However, the reliable use of this technique in static, fixed field intensity modulated radiation therapy (IMRT) is still a concern due to static IMRT's often extended delivery time.

Recently a new IMRT treatment technique, volumetric-modulated arc therapy (VMAT) has gained widespread interest in the radiation therapy community.^{15–25} VMAT features simultaneous variations in MLC aperture shape, dose rate, and gantry rotation speed with the radiation continuously on. Compared to static IMRT, one of the most significant benefits of VMAT lies in the much shorter delivery time. For instance, a single-arc VMAT treatment can be completed less than 2 min. This feature would make VMAT valuable in

treating moving tumors aided by dynamic multileaf collimator (DMLC) tracking as motion-compensation strategy.

Several DMLC tracking algorithms have been reported for commonly used radiation therapy treatment modalities.^{10,12,13} However, a robust and optimal solution specifically designed for VMAT is yet to be developed. Compared to static IMRT, VMAT delivery in single arc or multiple arcs is more complex due to additional degrees of freedom. VMAT delivery involves continuously rotating gantry and widely varying gantry rotation speed, dose rate, and MLC leaf position. When target motion compensation is incorporated, these delivery parameters require interactive modulations. To ensure that the treatment plan can be delivered without violating limitations of hardware parameters of a linear accelerator (linac), a tracking solution needs to consider a number of machine constraints such as maximal permissible gantry speed, dose rate, and MLC leaf-velocity. In addition, as VMAT often requires the gantry to accelerate and decelerate frequently to deliver a treatment plan, gantry acceleration constraint should also be taken into account.

In this paper, a DMLC tracking algorithm is developed capable of delivering VMAT to targets that experience two-dimensional (2D) motions in the beam's eye view (BEV). The VMAT plan is optimized based on patient's anatomy at one particular phase of the respiratory cycle using a commercial treatment planning software. The interdependence between several linac parameters, such as dose rate, gantry angular speed, gantry acceleration and MLC leaf-velocity, and target motion is investigated to formulate the VMAT delivery problem for moving targets. An iterative search algorithm is designed with hardware constraints included to determine the locally optimal delivery parameters for 2D DMLC tracking. The goal is to achieve optimal delivery efficiency, i.e., the shortest delivery time while minimizing delivery error resulted from target motion.

II. MATERIALS AND METHODS

II.A. VMAT plans

Five RapidArc (Varian Medical Systems, Palo Alto, CA) plans are created to treat a 62 cc spherical tumor (5 cm diameter) in a lung phantom.²⁴ The complexity of each field is controlled by adjusting the monitor unit (MU) (312–900 MU) while maintaining the same daily tumor dose of 200 cGy. The planning machine is a Varian Novalis Tx linear accelerator with 2.5 mm leaf-width high resolution multileaf collimator. The treatment plan is optimized with gantry angle running counterclockwise from 179.9° to 180.1° (IEC convention) in a single arc. The plan information, including MLC leaf sequence, gantry angle and MUs is stored as discrete control points and exported in a DICOM file. Note that gantry speed, dose rate, and MLC leaf-velocity are not specified in the control points and will be determined later at the linac console by the VMAT delivery software once the plan has been transferred.¹⁷ Each plan consists of 177 control points, and the interval between any two consecutive control points is called a segment in this paper, thus totaling 176 segments. To deliver the VMAT plan to a moving target us-

ing DMLC tracking, the planned MUs in each segment are kept as invariant so that the integrity of the planned conformal dose distribution will not be compromised.

II.B. Motion model

The target is assumed to be a rigid body so that its translational movements in three-dimensional space can be defined by target center's trajectory in the BEV plane.¹⁰ A BEV plane at gantry angle g is defined as a plane that is perpendicular to the beam central-axis and passes through the linac isocenter. Target motion relative to the treatment beam is eliminated in a VMAT delivery by moving the apertures at each gantry angle g in synchrony with the target motion so that the beam portals remain spatially unchanged relative to the target in the BEV plane. In this work, the motion trajectory of a target is assumed to be known prior to the treatment. This assumption is made for simplicity but the algorithm is readily adaptable for real time motion tracking.^{9,12}

The 2D BEV projection of the target motion can be divided into two components: One describes the motion along the x axis (parallel to MLC leaf travel direction), $X(t)$, and the other describes the motion along the y axis (perpendicular to MLC leaf travel direction), $Y(t)$. Lujan's respiratory motion model^{1,26} is used, and defined as the following:

$$x(t) = A_x \sin^4(\pi t / \tau + \phi_x), \quad (1)$$

$$y(t) = A_y \sin^4(\pi t / \tau + \phi_y), \quad (2)$$

where A_x and A_y are the motion amplitudes along the x -axis and y -axis, respectively, τ is the period, ϕ_x and ϕ_y are the initial phases. A phase difference between the motion along x - and y -axis is deliberately set to 0.1π , thus producing an elliptical trajectory in the BEV to simulate commonly observed hysteresis. The magnitude of x motion is set to 2 cm, which is along the patient's superior-inferior direction and is therefore gantry angle independent, while the amplitude of y component is gantry angle dependent, given by $A_y = A_{y0} \sin(g)$, where A_{y0} is the magnitude of target motion along the patient's anterior-posterior direction and g is gantry angle. In this work, A_{y0} is set to 0.5 cm and the motion period τ is set to 6 s. The motion trajectory using these parameters in the BEV plane at gantry 90° is illustrated in Fig. 1 (compare to Fig. 4 of Ref. 27). The origin is at the full-exhale phase in a breathing cycle. The shape of the motion trajectory and the range of travel in each direction are chosen so as to encompass the vast majority of observed respiratory motion.^{27–30}

II.C. Machine constraints

In static IMRT the intensity modulation is achieved through leaf motion at several fixed gantry angles. Previously reported DMLC tracking algorithms therefore only consider MLC leaf-velocity constraint.^{10,13} By contrast, VMAT produces intensity modulation through a combination of MLC leaf motion, gantry rotation, and dose rate change. As such, delivery time is constrained by the allowable gantry speed, dose rate, and MLC leaf-velocity. For DMLC target

FIG. 1. Illustration of the motion of the target projected in the beam's eye view plane. The peak-to-peak amplitudes are 2 cm along the x-axis and 0.5 cm along the y-axis, respectively. The phase difference of 0.1π is applied to produce a hysteresis motion.

tracking in VMAT, another machine parameter, gantry acceleration constraint, has to be considered in the algorithm as well. Machine constraints are listed in Table I. They represent the machine specifications of a modern Varian linac with MLC. Note that for a moving target, delivery efficiency is also limited by the target motion speed. The MLC leaves are required to run fast enough to catch the motion of the target.

II.D. Relations between delivery parameters

A VMAT plan is uniquely defined by the shaped apertures A at each control point associated with gantry angle g , denoted by $A(g)$, and by the beam weight prescribing the number of MUs associated with each aperture or segment, denoted by $M(g)$. $A(g)$ is a sequence of $2K$ numbers, $X_{L,k}(g)$ and $X_{R,k}(g)$, that specify the left and right leaf positions of the leaf pairs $k=1, 2, \dots, K$ at a given gantry angle g . During the delivery of a VMAT plan, the gantry rotates around the patient while both beam aperture $A(g)$ and beam weight $M(g)$ change as a function of gantry angle g . The relationships between the delivery parameters in VMAT are described in the following.

- (1) The beam weight $M(g)$ at a given gantry angle g during the delivery can be expressed in terms of gantry speed dg/dt and dose rate dm/dt as

$$M(g) = \frac{dm}{dg}(g) = \frac{dm}{dt}(g) \cdot \frac{dt}{dg}(g), \quad (3)$$

where $m(g)$ is the accumulative beam weight delivered until gantry angle g . Thus, the dose rate dm/dt at g is given by

$$\frac{dm}{dt}(g) = M(g) \cdot \frac{dg}{dt}(g), \quad (4)$$

showing that the dose rate dm/dt is related to the gantry angular speed dg/dt at gantry angle g and the $M(g)$ prescribed by the VMAT plan.

- (2) The change of beam aperture $A(g)$ as a function of gantry angle, $V_k(g) = dX_{L(R),k}(g)/dg$, can be related with gantry speed dg/dt and MLC physical leaf-velocity $V_{L(R),k}^{\text{phys}}(g)$ as follows:

$$V_k(g) = \frac{dX_{L(R),k}(g)}{dg} = \frac{dX_{L(R),k}(g)}{dt} \cdot \frac{dt}{dg} = \frac{V_{L(R),k}^{\text{phys}}(g)}{[dg/dt]}, \quad (5)$$

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$$V_{L(R),k}^{\text{phys}}(g) = V_k(g) \cdot \frac{dg}{dt}. \quad (5')$$

For VMAT delivery to a moving target with a velocity of $V_{\text{target}}(t)$ in the BEV plane, the physical MLC leaf-velocity $V_{L(R),k}^{\text{phys},m}(t)$ in the MLC-reference frame is then given by

$$V_{L(R),k}^{\text{phys},m}(t) = V_k(g) \cdot \frac{dg}{dt} + V_{\text{target}}(t). \quad (6)$$

II.E. Target tracking algorithm

The goal of the DMLC tracking algorithm in VMAT is to achieve optimal delivery efficiency, i.e., the shortest delivery time, with both treatment plan invariance and limitations on

TABLE I. Summary of machine specifications for a Varian linear accelerator.

Machine parameter	Specification
Maximum gantry speed	5.5 deg/s
Maximum MLC leaf-velocity	3.5 cm/s ^a
Maximum dose rate	600 MU/min
Maximum gantry acceleration	1.8°/s ² ^b

^aReference 31.

^bReference 37.

FIG. 2. Logic flow of the DMLC tracking algorithm for VMAT delivery to targets moving in 2D in beam's eye view.

delivery parameters kept in mind. As the delivery time t_m for each segment m is mainly governed by gantry speed, we aim to achieve maximum gantry speed at each segment. The delivery time, gantry speed, MLC leaf-velocity, dose rate, and motion trajectory are strongly correlated in the VMAT delivery, and the hardware parameters are further constrained by their respective limits. Thus, the problem of optimal delivery of VMAT to a moving target using DMLC tracking can be solved through an iterative search process aiming at local minimization of the delivery time for each segment. The overall logic flow of the DMLC tracking algorithm is shown in Fig. 2 and described next in detail.

II.E.1. VMAT plan extraction

The MLC leaf positions and number of monitor units as a function of gantry angle are derived from the DICOM files. The dynamics of an arc delivery to a moving target is equivalent to a continuous evolution of an open area $A(t)$ with time t , which is made of strip openings formed by the edges between opposing left and right leaves of each pair, $A_k(t)$. Index $k=1, 2, \dots, K$ enumerates the leaf pair and $k\Delta$ denotes the leaf position along the y axis (Δ is the leaf-width). For a given t , the leaf aperture $A(t)$ can be defined by the leaf position $X_m^L(k\Delta)$ and $X_m^R(k\Delta)$, where L and R denote left and right leaf banks, respectively.

II.E.2. Iterative search process

- (i) Initialize the values of delivery duration t_m for the segment m , where $m=1, 2, \dots, 176$.
- (ii) Determine the gantry speed for segment m . The gantry speed ω_m is given by $\omega_m = \Delta g / t_m$, where Δg is the gantry angle interval at each segment (approximately 2° for the VMAT plan used in this study).

- (iii) Determine the target position $x(\sum_{i=1}^m t_i)$ and $y(\sum_{i=1}^m t_i)$ from Eqs. (1) and (2).
- (iv) Calculate new leaf positions. The new leaf positions are calculated to account for the changes in target position. For the x motion component, the new leaf positions at t_m are

$$X_m^{L(R)}(k\Delta)|_{\text{new}} = X_m^{L(R)}(k\Delta)|_{\text{plan}} + x\left(\sum_{i=1}^m t_i\right). \quad (7)$$

For the y motion component, an appropriate leaf “shifting” is made. Similar techniques have been recently described by several groups.^{10,12,13} The shifting technique follows the y component in discrete increments of the leaf-width, i.e., 2.5 mm. When the displacement along the y -axis is more than one leaf-width, the leaf pair k ($k=1, 2, \dots, K$) moves rightward to the current position of leaf pair $k+1$ or leftward to the current position of leaf pair $k-1$, depending on the target moving direction. As a result, the spatial position of the aperture is shifted by a leaf-width Δ along the y -axis while the shape is preserved. Note that the leaf shift can be performed whenever the target's y -displacement is more than $\pm k\Delta$ and is not restricted to shifting single leaf-pair if the y -component motion is greater or equal to two leaf-widths.

- (v) Apply machine constraints to each delivery parameter at the segment m . The dose rate dm/dt is calculated by Eq. (4). The planned leaf positions determine $V_{L(R),k}^{\text{phys}}(g)$ using Eq. (5'), which, in turn, gives the physical MLC leaf-velocity for a moving target using Eq. (6). Gantry speed, dose rate, and MLC leaf-velocity all have to fall within respective limits listed in Table I. If any of these three parameters is larger than the specification, the routine loops back to step (i), increases t_m by a small step of 0.02 s, and continues to step (v) until all the machine constraints are satisfied. The delivery parameters for segment m are now temporarily determined.
- (vi) Apply gantry acceleration constraint. In the same manner as described above, an optimal delivery time, t_{m+1} , and gantry speed $\omega(m+1)$ and other delivery parameters for the subsequent segment $m+1$ are temporarily determined by repeating steps (i)–(v). For a smooth and reliable gantry rotation, the gantry acceleration limit a_{max} has to be enforced. The principle of imposing gantry acceleration constraint is illustrated in Fig. 3. The allowed gantry speed $\omega(m+1)$ has to fall within the interval $[\omega(m) - a_{\text{max}} \times t_{m+1}, \omega(m) + a_{\text{max}} \times t_{m+1}]$. If the calculated gantry speed $\omega(m+1)$ is larger than $\omega(m) + a_{\text{max}} \times t_{m+1}$, i.e., the gantry accelerates, $\omega(m+1)$ is required to be decreased to a value equal to $\omega(m) + a_{\text{max}} \times t_{m+1}$. If $\omega(m+1)$ is smaller than $\omega(m) - a_{\text{max}} \times t_{m+1}$, i.e., the gantry decelerates, $\omega(m)$ is required to be decreased by a small step of $0.02^\circ/\text{s}$, and $\omega(m+1)$ is then recalculated until $\omega(m+1)$ is within the interval $[\omega(m) - a_{\text{max}} \times t_{m+1}, \omega(m) + a_{\text{max}} \times t_{m+1}]$. For the latter scenario, delivery parameters for the seg-

FIG. 3. Illustration of imposing gantry acceleration constraint at segment $m+1$. (a) If $\omega(m+1) > \omega(m) + a_{\max} \times t_{m+1}$, $\omega(m+1)$ is reduced to $\omega(m) + a_{\max} \times t_{m+1}$. (b) If $\omega(m+1) < \omega(m) - a_{\max} \times t_{m+1}$, $\omega(m)$ is reduced at a small step, and $\omega(m+1)$ is recalculated until $\omega(m+1)$ is within $[\omega(m) - a_{\max} \times t_{m+1}, \omega(m) + a_{\max} \times t_{m+1}]$.

ment m need to be recalculated. Finally, the delivery parameters for the segment m are determined.

- (vii) Repeat steps (i)–(vi) for all the 176 segments so as to determine the delivery parameters for the entire arc.

II.E.3. Delivery accuracy estimation

The delivery of fluence to the moving target is simulated. Similar to commonly employed composite verification of a

static IMRT plan, the VMAT fluence from each beam angle is projected to a planar dosimeter that lies parallel to the treatment floor. The total fluence delivered to the target for no tracking, 1D tracking (along the x -axis only), and 2D tracking is calculated as the sum of 176 segments in the *target-reference frame*. The pixel size is $2.5 \text{ mm} \times 2.5 \text{ mm}$. A difference map between the planned fluence and the delivered fluence is produced by subtracting the initial desired fluence map. For comparison and statistics purposes, the total monitor units are normalized to 200 MUs for the five VMAT plans used in this study. To avoid biasing the statistics by including excessive zero-valued pixels on the fluence map, only those pixels with a nonzero value on the fluence map are used to calculate delivery accuracy statistics.

III. RESULTS

Figures 4(a)–4(d) show gantry speed, gantry acceleration, MLC leaf-velocity, and dose rate as a function of delivery time for a VMAT 2D tracking delivery. Red solid lines represent the maximal permissible values for each parameter. For this VMAT tracking delivery, a delivery time of 95.6 s can be achieved, which is only 30.2 s longer than that needed for delivery without tracking (Table II). For the optimal delivery, i.e., the shortest delivery time, at least one of the machine parameters reached its maximum at certain beam angles. The delivery time is mainly determined by the gantry speed, which is strongly modulated so that the MLC leaves can have sufficient time to reach the desired positions. The

FIG. 4. Optimal delivery parameters for a VMAT plan with 2D DMLC tracking as a function of time in a 360° single arc. (a) Gantry speed, (b) gantry acceleration, (c) MLC leaf-velocity, and (d) dose rate.

TABLE II. Treatment time for VMAT delivery. Column 3: No tracking, column 4: 2D tracking.

Plan No.	MU	No tracking (s)	2D tracking (s)
1	312	65.4	93.9
2	460	65.4	95.5
3	550	65.4	95.6
4	750	76.1	103.8
5	900	90.8	117.9

dose rate and MLC leaf-velocity are also significantly modulated as all these delivery parameters are correlated.

Figures 5(a) and 5(b) show the trajectories of one leaf pair (leaf pair No. 25) as a function of gantry angle for the same plan in the absence and presence of 2D motion-compensation. Both leaves have to travel larger distance in order to track the moving target. Periodic modulation of the trajectories can also be observed [Fig. 5(b)]. Figure 5(c) shows the area of the MLC aperture as a function of gantry angle for the no-tracking delivery and 2D tracking delivery, respectively. These two curves overlaid each other perfectly, indicating that the VMAT plan integrity remains intact.

While the aperture shape is able to be dynamically adapted to follow the target motion at each control point during DMLC tracking, it should be cautioned that the MLC leaves may not be exactly synchronized with the target motion due to *linear* MLC leaf-motion and *nonlinear* target motion (Fig. 1), particularly for large target motion velocity. The tracking accuracy along the x direction is shown in Fig. 6 with a time resolution of 0.1 s. The root-mean-square (RMS) tracking error is 0.44 mm, indicating a submillimeter track-

FIG. 5. Trajectories of one leaf pair (leaf pair No. 25) as a function of gantry angle for the delivery in Fig. 4 in the absence (a) and presence (b) of 2D motion-compensation. (c) The area of the MLC aperture as a function of gantry angle. These two curves overlaid each other perfectly, indicating that the VMAT plan integrity remains intact.

FIG. 6. Geometric accuracy of DMLC tracking for parallel-to-leaf target motion. The time resolution is 0.1 s. The root-mean-square geometric error is 0.44 mm.

ing accuracy for the x component. Note that both the magnitude and velocity for the y component are much smaller than that along the x -axis. The tracking error along this direction via leaf shifting is discussed in Sec. IV.

Figure 7(a) shows the planned intensity map for above VMAT plan. Figures 7(b)–7(d) show delivered fluence for no-tracking, 1D tracking, and 2D tracking. Note that the total monitor units are normalized to 200 MUs for comparison purpose. The discrepancies between the planned and delivered fluence are displayed in Figs. 7(e)–7(g). When the plan is delivered without any tracking, a clear mismatch can be recognized. The fluence is severely blurred along the path of motion. Delivery error is particularly large at the boundary regions of the target due to the high dose-gradient in these areas. As a result, significant underdose or overdose occurs on the difference map. The blurring is significantly removed by employing 1D tracking along the x direction, and best match is achieved with 2D tracking. The average RMS difference of the five plans is 14.8 MU for no motion tracking, 3.6 MU for 1D tracking, and 2.1 MU for 2D motion tracking. The respective gamma passing rates (1%/1 mm) are 48%, 90%, and 94%. These simulation data indicate that 2D DMLC tracking can achieve high delivery accuracy to a moving target.

IV. DISCUSSION

This paper presents a DMLC tracking algorithm that achieves optimal delivery efficiency, i.e., the shortest delivery time for VMAT delivery to a moving target. This algorithm takes advantages of the capabilities provided by modern medical linear accelerators in simultaneously modulating MLC leaf position and velocity, gantry speed, and dose rate. It is essential that their respective constraints are kept in mind in the design of such a tracking algorithm for VMAT. Our results demonstrate that a VMAT plan with MLC tracking can be efficiently delivered in less than 2 min, making it promising for treatments of moving tumors.

FIG. 7. (a) The planned intensity map for the same VMAT plan in Fig. 4. [(b)–(d)] Delivered fluence for no-tracking, 1D tracking, and 2D tracking. Note that the total monitor units are normalized to 200 MUs. [(e)–(g)] The discrepancies between the planned and delivered fluence.

The delivery time of VMAT delivery using DMLC tracking is mainly determined by the gantry speed, gantry acceleration, dose rate, and MLC leaf-velocity. All these delivery parameters are widely modulated within their specifications (Fig. 4). Although the gantry speed has to slow down at certain control points, the average gantry speed of approximately $4^\circ/\text{s}$ is still large enough to achieve high delivery efficiency. It should be noted that neither MLC leaf-acceleration constraint nor dose rate change constraint is included in our algorithm. High mechanical acceleration capability allows a MLC leaf to respond sufficiently fast to the target motion. The maximum leaf acceleration observed in the Fig. 4 is about 4 cm/s^2 , which is well within the acceleration limit measured on Varian systems ranging from 46 to 69 cm/s^2 (Ref. 31). Modern linac is able to modulate dose rate fast because the durations of both the klystron and

electron-gun signals are of the order of microsecond with a pulse-to-pulse duration of the order of millisecond.

In addition to the high delivery efficiency, our algorithm is capable of delivering VMAT accurately using 2D compensation strategy. If without tracking, a tumor margin that accounts for target motion needs to be at least 2 cm, i.e., the amplitude of the target motion. In contrast, the required margin for 2D tracking delivery reduces to about 2 mm, which is about the same amount as the localization accuracy that is achievable by modern IGRT (image-guided radiation therapy). This relatively small motion-margin is primarily resulted from the finite MLC leaf-width because the target motion along the y-axis is tracked by discrete increments of the leaf-width, i.e., 2.5 mm. It is conceivable that smaller MLC leaf-width would render smaller delivery error. The most significant error between planned and delivered inten-

sities appears in the boundary regions of a moving target. This agrees with an intuitive perception that errors are prevalent in the regions where the fluence gradient is the largest.

A variety of approaches have been proposed to improve DMMLC tracking efficiency and accuracy for static IMRT.^{10,12} These approaches also apply to VMAT delivery improvement. One approach, for example, is to design a MLC with faster moving leaves. This will enable the leaves to reach the desired positions quickly so as to minimize potential underdose or overdose. This will also reduce gantry modulation so that the gantry does not have to slow down frequently. Less gantry modulation and faster leaf-velocity will lead to both faster treatment and less dosimetric uncertainty. Another approach is to include a leaf-position gradient constraint for any two adjacent MLC leaves in the static VMAT plan optimization. For example, we observe that in one segment of a VMAT plan, two neighboring leaves differ by as much as 2.5 cm in spatial position. In order to accomplish leaf-shifting for y component compensation, one leaf is required to reach the other leaf's position. To ensure that this shifting has sufficient time, the gantry has to slow down in a short period of time from maximum gantry speed of 5.5°/s to about 1°/s. An alternative approach would be to develop a MLC assembly with movable carriages that can move along the y -axis (perpendicular to MLC leaf travel direction), thereby being able to compensate for the y -axis motion. This solution will not only avoid the need of shifting MLC leaves but also reduce the delivery error caused by finite MLC leaf-width.

Current algorithm assumes the known target motion prior to the treatment. The motion model is reasonably assumed for the purpose of future real time clinical implementation.¹² A modification of the motion model used in this study may result in different delivery efficiencies and accuracies. For example, smaller amplitudes in both directions will result in improved delivery efficiency and accuracy. If the motion along the y -axis is less than the leaf-width, it will become unnecessary to shift the MLC leaves, which avoids large gantry deceleration. As a result, higher efficiency is expected. Delivery efficiency and accuracy also depend on target-motion period. For example, if the period decreases from 6 to 3 s, the delivery time will be reduced from 96 to 87 s for the VMAT delivery presented in Fig. 4. In addition, the reference phase in our algorithm is assumed to be at the full-exhale position in a breathing cycle. For different reference phase positions, the delivery time and RMS difference in the fluence map are found to vary less than 10 s and 1.5 MU, respectively. Note that the combination of the planned leaf-velocity and the moving-tumor's velocity rarely exceeds maximal permissible MLC leaf-velocity of 3.5 cm/s. If it does occur, the VMAT plan with DMMLC tracking may not be deliverable without violating the integrity of a VMAT plan. Analysis of this issue is beyond the scope of this work.

It should be noted that the VMAT plans used in this study are created based on patient's anatomy at one particular phase of the respiratory cycle. Relative motion is possible between the target and surrounding normal tissues from phase to phase. A future refinement would be the design of a "deliverable" optimized VMAT tracking plan based on a

planning 4D CT that accounts for the phase-to-phase positional changes of critical organs with respect to the tumor and deformation of targets.^{32,33}

Finally, the purpose of this study is to develop a DMMLC tracking method for VMAT delivery and to illustrate the complex interdependence of a number of delivery parameters. It is not intended as a report on the development of a system ready for clinical implementation. Although the work presented in this paper is merely an early study and preliminary in nature, it serves our purpose of deriving a novel solution for managing the intrafraction motion in the VMAT delivery and is the first step toward the realistic clinical study. Future clinical feasibility and verification studies will involve the implementation of this method in the VMAT delivery software on a linac, a phantom that incorporates a real time, high resolution three-dimensional dosimeter, and a real time respiration monitoring device. The realistic system latency,^{12,34} a motion prediction method,^{35,36} and a "beam hold" mechanism for anomalous conditions will also be included for experimental demonstration. The experimental VMAT delivery efficiency and accuracy on a moving target will be compared to the results presented in this paper.

V. CONCLUSION

An algorithm for compensating the 2D target motion in the beam's eye view has been developed for the VMAT delivery aiming at the shortened delivery time. The simulation demonstrates that VMAT treatment to a moving target using 2D DMMLC motion tracking can be accurately and efficiently delivered with only half a minute increase in delivery time compared to delivery without motion-tracking. The delivery errors arise mainly from the finite MLC leaf-width, while delivery efficiency is mainly determined by the maximum allowed MLC leaf-velocity and by gantry acceleration limit.

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