TOPICAL REVIEW

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Intensity-modulated arc therapy: principles, technologies and clinical implementation

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Abstract

Intensity-modulated arc therapy (IMAT) was proposed by Yu (1995 *Phys. Med. Biol.* **40** 1435–49) as an alternative to tomotherapy. Over more than a decade, much progress has been made. The advantages and limitations of the IMAT technique have also been better understood. In recent years, single-arc forms of IMAT have emerged and become commercially adopted. The leading example is the volumetric-modulated arc therapy (VMAT), a single-arc form of IMAT that delivers apertures of varying weights with a single-arc rotation that uses dose-rate variation of the treatment machine. With commercial implementation of VMAT, wide clinical adoption has quickly taken root. However, there remains a lack of general understanding for the planning of such arc treatments, as well as what delivery limitations and compromises are made. Commercial promotion and competition add further confusion for the end users. It is therefore necessary to provide a summary of this technology and some guidelines on its clinical implementation. The purpose of this review is to provide a summary of the works from the radiotherapy community that led to wide clinical adoption, and point out the issues that still remain, providing some perspective on its further developments. Because there has been vast experience in IMRT using multiple intensity-modulated fields, comparisons between IMAT and IMRT are also made in the review within the areas of planning, delivery and quality assurance.

1. Historical review

1.1. Early development that led to IMAT

Although arc therapy can be traced back to the dawn of the 20th century (Johns *et al* 1953), arcs involving dynamic field shaping using a multileaf collimator were first described by Takahashi (1965). He described a method of rotational therapy, which we now refer to as conformal arc therapy, where the beam aperture shaped by a multiple leaf collimator (MLC)

dynamically varies to match the beam's-eye-view (BEV) of the target. In 1982, Brahme *et al* (1982) solved an integral equation for a hypothetical target wrapped around a critical structure to be treated with arc therapy. They demonstrated that in order to deliver a uniform dose to the target while sparing the critical structure, the beam intensities have to be modulated. In 1983, Chin *et al* proposed and demonstrated that with computer optimization and the freedom afforded by computer-controlled gantry rotation, collimator motion and dose-rate variation, a highly conformal dose distribution can be achieved (Chin *et al* 1983).

These initial developments on arc therapy were accompanied and followed by the development and wide adoption of three-dimensional conformal radiation therapy (3DCRT) in the 1980s (Perez *et al* 1995). The need for more convenient field shaping brought MLC to radiotherapy practice. Brahme *et al* (1988) published a paper showing that if the intensities of radiation can be modulated across a radiation field, the increased freedom would afford a greater ability to shape the volume of high doses, to better conform to the target than 3DCRT. The motorized field shaping capabilities of MLC were quickly explored to modulate the intensities within a radiation field. Intensity-modulated radiation therapy (IMRT) aims to deliver a highly conformal dose to a tumor, while sparing the surrounding normal tissues and sensitive structures. Convery and Rosenbloom (1992) derived a mathematical formula for realizing intensity modulation with the dynamic movement of a collimator. During 1994–95, more works were published to demonstrate the feasibility of using MLC for intensity modulation in either the dynamic mode or static mode (Bortfeld *et al* 1994a, 1994b, Yu and Wong 1994, Spirou and Chui 1994, Stein *et al* 1994, Yu *et al* 1995a). The amount of work on this emerging technology quickly mushroomed, and clinical implementations of the IMRT technique immediately followed (Ling *et al* 1996, Burman *et al* 1997, Chui *et al* 2001).

Mackie *et al* (1993) proposed another form of IMRT using rotational fan beams, called tomotherapy. At the same time, commercial development of tomotherapy was also rendered by NOMOS Corporation (Carol *et al* 1993, Carol 1995a, 1995b). Intensity modulation was achieved with a binary collimator, which opens and closes under computer control. As the fan beam continuously rotates around the patient, the exposure time of a small width of the fan beam, or a beamlet, can be adjusted with the opening and closing of the binary collimator, allowing the radiation to be delivered to the tumor through the most preferable directions and locations of the patient. The initial commercial system by NOMOS Corporation added the binary collimator onto a linear accelerator and delivered radiation treatments one slice at a time with the slice thickness equal to the width of two beamlets. The treatment table had to be precisely indexed from one slice to the next. Helical tomotherapy was then developed by Tomotherapy, Inc. as a dedicated rotational IMRT system with a slip-ring rotating gantry achieving more efficient delivery by continuous gantry rotation and treatment couch translation.

The dosimetric advantages of rotational treatments are illustrated by Shepard *et al* (1999), which summarizes results from an optimization series performed for a C-shaped target with a sensitive structure in the concavity of the C. For these simulations, all planning parameters, such as percent dose constraints, were held constant except for the number of beam angles. The results are summarized in table 1. It was shown that each increase in the number of beam angles led to a more homogeneous dose in the tumor and a lower dose to the sensitive structure. Significant dosimetric improvements continued well beyond the number of beam angles typically used for fixed-field IMRT. It is also noteworthy that the total integral dose is nearly independent of the number of beam angles.

Intensity-modulated arc therapy (IMAT) was introduced by Yu (1995b). Like tomotherapy, IMAT delivers photon radiation treatment in an arc manner. Instead of using rotating fan beams as in tomotherapy, IMAT uses rotational cone beams of varying shapes and

Table 1. Impact of a number of beam angles on plan quality. Reproduced from Shepard *et al* (1999).

No of angles	Obj. funct. value	Std. dev. in target dose	d_{95}	Mean dose to RAR	Total integral dose
3	0.665	0.124	0.747	0.488	2732.5
5	0.318	0.090	0.814	0.215	2563.3
7	0.242	0.064	0.867	0.206	2596.8
9	0.222	0.064	0.855	0.192	2598.3
11	0.202	0.058	0.879	0.186	2570.2
15	0.187	0.053	0.908	0.180	2542.9
21	0.176	0.049	0.912	0.171	2545.1
33	0.151	0.038	0.933	0.155	2543.5

varying dose weightings to achieve intensity modulation. Starting from the same tomotherapy plan, which approximates the full arc with evenly spaced fixed fields, the strategy was to convert the intensity patterns into multiple segments and deliver with overlapping arcs. Based on the fact that numerous segment configurations can yield the same intensity pattern, it is possible to find a segment configuration at each beam angle such that segments at successive angles are connected geometrically. The stacks of overlapping beam apertures can then be delivered with multiple overlapping arcs.

Through the initial proof-of-principle study, it was shown that IMAT could be a valid alternative to tomotherapy in terms of treatment delivery. However, unlike tomotherapy, IMAT must account for restrictions on MLC movement as the gantry moves from one beam angle to the next. Because deliverability must take priority, an optimal field shape may have to be altered in order to produce smooth delivery. As a result, plan quality would be adversely affected for some cases. This restriction does not apply to tomotherapy due to the use of a binary MLC. Therefore, tomotherapy should theoretically have the best plan quality (Bortfeld and Webb 2009). As compared with tomotherapy, IMAT also has some advantages: (1) IMAT does not need to move the patient during treatment and avoids abutment issues as seen with serial tomotherapy; (2) IMAT retains the ability of using non-coplanar beams and arcs, which has great value for brain and head/neck tumors; (3) IMAT uses a conventional linac, thus complex rotational IMRT treatments and simple palliative treatments can be delivered with the same treatment unit.

1.2. Continued efforts in planning and clinical implementation

In spite of the demonstrated advantages, there were limited research activities on IMAT between 1995 and 2006. The potential reasons may lie in the lack of an efficient planning method for IMAT and the lack of commercial interest. This section summarizes the major efforts in the planning and clinical implementation of IMAT during this relatively dormant period.

In 2000, a phase I clinical trial of IMAT using forward planning was conducted at the University of Maryland (Yu *et al* 2002), to test the safety and feasibility of changing the field shape during gantry rotation. Fifty patients with cancers of various sites were treated using IMAT. Due to the lack of an IMAT inverse planning system, forward planning was used to determine the arc range and aperture shapes. Arcs were approximated as multiple shaped fields spaced every 5° to 10° around the patient. Multiple coplanar or non-coplanar arcs

were allowed. At each beam angle, irregular field shapes were defined based on the BEV of the planning target volume and normal critical structures. Typically, at a given angle, one field shape conformed to the BEV of the target and additional field shapes were set to shield individual critical structures that overlapped the target in the BEV. Because beam rotation and irradiation are concurrent, the delivery is very efficient. A typical treatment with three to five arcs takes less than 10 min from start to finish, which is comparable to conventional techniques. As is the case with most forward planning techniques, these approaches rely heavily on the experience of the planner and could result in sub-optimal plans as well as prolonged planning times. Nonetheless, standard solutions can be developed for less challenging treatment sites, such as the prostate (Ma *et al* 2001). IMAT could also be used to treat a target wrapped around a critical structure, as demonstrated by Cotrutz *et al* (2000).

Another clinical study was conducted at Ghent University Hospital. The aperture shapes were first determined based on the BEV of the target and critical structures, similar to the approach adopted by University of Maryland. However, the anatomy-based apertures were further refined by allowing the leaves to move slightly using a greedy search optimization scheme (DeGersem 2004). Treatment planning studies were published for rectal cancer and whole abdominopelvic radiation therapy (Duthoy *et al* 2003, 2004).

Wong *et al* (2002) formulated the forward planning into a practical approach which they termed simplified IMAT (SIMAT). SIMAT starts by creating multiple arcs based on the BEV of the anatomy, with each arc serving a distinct planning goal such as covering the whole target or protecting one critical structure. The weightings of the arcs were subsequently optimized, assuming a constant dose rate delivery. The SIMAT strategy was applied to various sites including prostate and high-risk endometrial cancer (Bauman *et al* 2004, Wong *et al* 2005).

Although these early works kept the rotational IMRT alive, they suffered from the lack of efficient inverse treatment planning methods. On the other hand, most of these early developments on IMAT were rendered with equally spaced beams under the technical limitation that the machine could not vary the dose rate dynamically during gantry rotation. Under the assumption that the machine dose rate has to be constant during arc rotation, Yu *et al* (2006) proposed a hybrid of IMRT and IMAT to increase delivery efficiency. Using an angular cost function to define the angular weights, an arc delivered with a constant dose rate can be supplemented with fixed intensity-modulated fields at a few most important angles.

1.3. Work leading to the commercialization of single-arc IMAT

In proposing IMAT as an alternative to tomotherapy, Yu (1995b) predicted that with the increase in the number of gantry angles, the number of intensity levels at each gantry angle can be reduced without degrading plan quality. It was argued that the plan quality is a function of the total number of strata, defined as the product of the number of beam angles and the number of intensity levels. In other words, it is the total number of aperture shape variations that determine the plan quality. Assuming this is true, a single arc with a sufficient number of aperture shape variations would be able to create optimal treatment plans. Many subsequent works have attempted to use a single arc for IMAT. As illustrated by Jiang *et al* (2005), a single arc with 36 beam aperture variations under a constant dose rate cannot realize the optimal plan quality. To reach the desired plan quality, one must either increase the number of field segments or apertures, or allow the dose rate to vary during gantry rotation, or both.

MacKenzie and Robinson (2002) proposed a technique whereby 24 equally spaced beam orientations are optimized for sliding window IMRT and arc delivery is performed by allowing the gantry of the linear accelerator to rotate to static gantry orientations and deliver the optimized sliding window IMRT deliveries. Crooks *et al* (2003) developed a single-arc IMAT

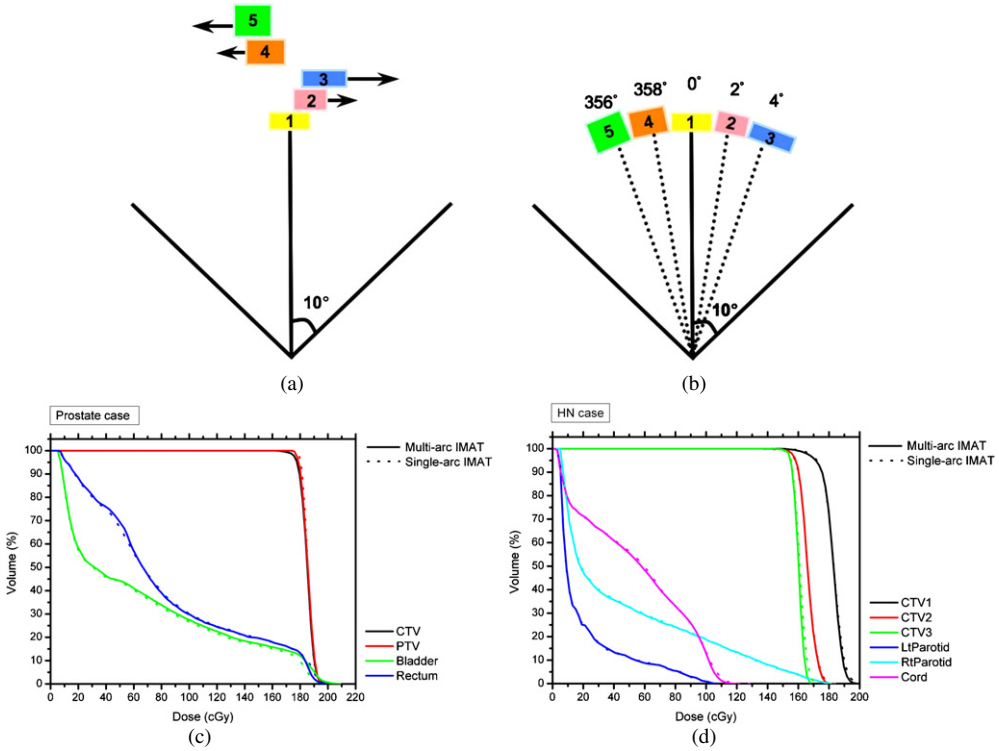

Figure 1. Converting multi-arc IMAT to single-arc delivery by (a) rearranging the stacked apertures at the planning beams in a multi-arc IMAT plan and (b) relocating the apertures into the planning angular interval resulting in a series of neighboring apertures. The resultant single-arc plans in (c) and (d) show minimal dosimetric degradation (from Tang *et al* (2007)).

planning algorithm that is based on the observation that the dose error resulting from beam apertures being delivered at angles a few degrees away from the planned angles is very small. In their algorithm, referred to as aperture-modulated arc therapy (AMAT), IMRT fields were created approximately 30° apart with 56–74 segments per beam direction. The segments were spread out based on the observation of the dosimetric insensitivity to angular deviations, and the plan was simulated and delivered in a single arc. Although the dose distributions from AMAT deviated from the original IMRT plan by over 10% at some locations, the overall dose patterns were similar.

Cameron (2005) developed a sweeping window arc therapy (SWAT) technique to deliver an IMRT treatment in one arc rotation. The collimator angle was initialized to 90° so that the leaf positions are normal to the axial plane of the patient. Shapes of the MLC apertures prior to optimization are initialized so that the MLC leaf positions sweep across the PTV as the gantry rotates around the patient. Optimization of MLC leaf positions is then performed by simulated annealing and arc weight optimization, which can be performed for a constant or variable angular dose rate.

Tang *et al* (2007) showed that a multi-arc IMAT could be converted into a single arc by spreading the stacked apertures to neighboring angles with a minimal effect on the plan quality. Figure 1 shows the method and part of their results. A five-arc IMAT plan was created by optimizing the aperture shapes and weights on 36 beam angles. The resulting

plan had five apertures stacked at each of the beams spaced every 10°. A new plan was then created by simply rearranging the stacked apertures into neighboring angles by minimizing the movement of the geometric center of the apertures as schematically shown in figures 1(a) and (b). Dose calculations for the original plan with stacked apertures and for the new plan with spaced apertures showed almost identical results for different plans, as shown in figures 1(c) and (d). This simple exercise elucidated that given the same number of aperture shape variations, single-arc IMAT and multi-arc IMAT theoretically have the same degree of freedom for optimizing the dose distributions, if the apertures in the single-arc arrangement could be geometrically connected. It also demonstrated that in rotational delivery, the dose distribution is insensitive to small angular deviations. Therefore, although there is no intensity modulation within each beam in single-arc IMAT, the needed intensity variation at the target region to take advantage of the geometric arrangement between the target and its surrounding critical structures is achieved with apertures from neighboring angles. For example, if the optimal intensity distribution at a given angle contains two peaks, it is not necessary to shape two disjointed apertures at this beam angle, which is not possible with MLCs from some vendors (Webb 2010). The two desired high intensity regions can be delivered from two or more beam angles. That is, the inability to modulate beam intensity in a beam is made up by the use of more beams. This simple fact is the key reason why single-arc IMAT works.

Ulrich *et al* (2007) developed an optimization technique whereby arc therapy plans are optimized for a single-arc delivery. In their algorithm, the aperture shapes are optimized by a tabu search optimization algorithm and the aperture weights are optimized by a gradient search. The algorithm demonstrates better treatment plans than an in-house IMRT optimization technique and requires a variable dose rate delivery with gantry rotation.

By assuming that the machine dose rate can vary as needed, Otto (2008) developed a single-arc IMAT algorithm that he referred to as volumetric-modulated arc therapy (VMAT). In addition to allowing dose-rate and gantry speed variation, the VMAT algorithm uses progressive beam angle sampling to optimize a large number (>100) of apertures using direct aperture optimization. The aperture shapes and weights are optimized initially for a number of coarsely spaced gantry angles with little consideration of aperture connectivity. Once the solution converges, additional gantry angles are inserted. As the angular spacing becomes smaller, the optimizer considers aperture shape connectivity both in the initialization of aperture shapes and during the optimization. The initial shapes of the newly inserted apertures are linearly interpolated from their angular neighbors. Such coarse-to-fine sampling is termed progressive sampling, and allows the optimization to converge faster. Because the aperture shape connectivity is ignored initially, the optimizer is given the freedom to aim for an optimal dose distribution. Since the final plan ensures aperture connectivity, the optimized single arc can be delivered within 2 min.

Luan *et al* (2008) developed an arc sequencing algorithm for converting continuous intensity maps, using a k-link shortest path algorithm, into multiple arcs. The algorithm was tested for prostate, breast, head and neck, and lung and it was demonstrated that the plans rivaled helical tomotherapy plans. Based on the method developed by Luan *et al* (2008), Wang *et al* (2008) sequenced the intensity patterns optimized for 36 beams into a single-arc delivery.

Bzdusek *et al* (2009) first optimizes the fluence maps based on static gantry angles that are evenly spaced at every 24° within the user-defined arc length. The optimized intensity maps are then converted into MLC segments and are evenly distributed within the arc, resulting in a single-arc MLC sequence. Compared to IMRT, this algorithm can achieve similar or better plan quality in prostate, head-and-neck, brain and lung cases. In a similar approach, Bedford (2009a) also optimized intensity maps for uniformly spaced beams over one or more arcs. The intensity maps are then sequenced into MLC apertures that approximate the fluence profiles.

A direct-aperture optimization is then used to improve the solution, taking into account the allowed range of leaf motion of the MLC.

These many contributions point to the same principle that because rotational delivery is not sensitive to small angular deviations, in-field intensity modulation can be traded with the use of more beam angles. As long as there are similar independent aperture variations in the optimized plans, single-arc IMAT and multi-beam IMRT can achieve similar plan qualities. The demonstration of the superior delivery efficiency of single-arc IMAT without sacrificing quality led to the present different commercial offerings of single-arc IMAT.

1.4. Commercial development and nomenclatures

Although IMAT has been proposed since 1995 and many researchers have developed different planning methods to demonstrate that IMAT is capable of creating highly conformal treatment plans that can also be efficiently delivered, large-scale clinical implementation did not start until Varian adopted Otto's VMAT algorithm (Otto 2008) and marketed it with the trade name, RapidArc™, in 2007. The linac control was also updated to allow dose rate variation during gantry rotation. Not long after Varian's announcement, Elekta started to market their IMAT solution with the trade name VMAT™. Bzdusek *et al* (2009) have introduced a rotational IMRT solution, which is marketed by Philips Medical Systems, Inc. with the trade name, SmartArc™. In describing their two-step planning method for single-arc IMAT, Wang *et al* (2008) named their method arc-modulated radiation therapy (AMRT). Other names, such as aperture-modulated arc therapy (Crooks *et al* 2003) and arc-modulated cone beam therapy (Ulrich *et al* 2007) were also used. VMAT is widely recognized as a single-arc technique that utilizes dose rate variation, although both single-arc IMAT (Yu 1995, Earl *et al* 2003) and dose rate variation had been employed for IMAT plan optimization before the nomenclature, VMAT, was proposed (Cao *et al* 2007). Since all of these variations subscribe to the same principle of IMAT, no trade names or other acronyms will be used except the original acronym, IMAT, in this review.

2. Planning

The general concept and process for IMAT planning are not very different from IMRT planning. The inverse planning principles are almost identical. However, due to the many degrees of freedom in IMAT planning, optimizing an IMAT plan is computationally more difficult. The differences are in the number of beams used to approximate an arc and the consideration of aperture connectivity. The difficulties in planning IMAT treatment, despite its many advantages, have been the main obstacle in the clinical implementation of IMAT. Effective planning tools for IMAT have only been developed recently (MacKenzie and Robinson 2002, Crooks *et al* 2003, Cameron 2005, Ulrich *et al* 2007, Shepard *et al* 2007, Cao *et al* 2007, Otto 2008, Wang *et al* 2008, Luan *et al* 2008, Oliver *et al* 2008, Bzdusek *et al* 2009, Bedford 2009a).

Because the speed of rotation cannot have frequent and drastic variations due to the weight of the linear accelerator's gantry, the variations in aperture weights are primarily achieved through varying the machine dose weight. The transition between the aperture shapes from one beam angle to the next is accomplished through dynamic motion of the MLC leaves. As the gantry rotates around the patient and the radiation beam is on, it is important that the subfields of adjacent beam angles do not require the MLC leaves to travel long distances. Ensuring such connectedness of adjacent subfields for smooth leaf motion is of great concern in the leaf-sequencing algorithm for IMAT. In addition, depending on the hardware capability,

dynamic collimator rotation can also be considered during IMAT optimization. The ability of rotating the collimator angle during delivery may increase the optimization freedom and produce better plan quality (Zhang *et al* 2010). It has also been shown that unwanted dose arising from the parked MLC leaf gaps can be minimized if the collimator angle is allowed to vary (Webb 2010).

As with conventional IMRT plan optimization, different methods for IMAT plan optimization can be grouped into two categories: two-step IMAT planning and one-step IMAT planning. Details of these two planning methods are provided in the following sections.

2.1. Two-step process

The two-step IMAT planning process starts by optimizing the intensity distributions for all beams used for approximating an arc. After the intensity optimization, a leaf-sequencing step is used to convert the optimized beam intensities into deliverable MLC segments to form an arc or arcs. During intensity optimization, no constraint related to delivery is imposed. Both the conversion of the intensities into segments and the connection of segments into deliverable arcs are considered in the leaf-sequencing step. In the initial work proving the feasibility of IMAT by Yu (1995), the two-step process was used. Recent works utilizing two-step planning include Cao *et al* (2006), Shepard *et al* (2007), Luan *et al* (2008), Wang *et al* (2008) and Bedford (2009a). The following summarizes the different approaches.

Optimized intensity distributions on tightly and uniformly spaced beams are first translated into a stack of superimposed irregular fields of uniform beam intensities. Different algorithms can be used for converting the intensity distributions into field segments of different shapes and weights. A leaf-sequencing algorithm attempts to define a sequence of MLC field shapes in order to create a deliverable intensity distribution that is as close as possible to the distributions obtained from the optimization. The stacks of field segments at all the beam angles must be linked together to form deliverable arcs. These two steps, approximating the intensity distribution using multiple uniform apertures and connecting the apertures from neighboring angles to form arcs, are normally not separated. In connecting apertures of adjacent beam angles, it may be necessary to alter the shape to force geometric connectivity. The corresponding errors created by such alteration can be compensated by both optimizing new weightings for these apertures and changing the shapes of the remaining apertures at the same beam angle.

The simplest leaf-sequencing algorithm assumes an ideal flat beam with no head scatter, and an ideal MLC with no transmission or leakage. In order to deliver a predictable dose distribution, a number of other refinements are often added in an accurate dynamic MLC sequence to account for effects such as field flatness, head scatter, penumbra, leaf leakage, rounded leaf ends and back-scatter into the transmission ion chamber. The under-dosing effects of the tongue-and-groove design of the MLC can also be included.

Yu used this two-step process in his initial illustration of using overlapping cone beam arcs for delivering tomotherapy plans (1995). Gladwish *et al* (2007) developed another work that converted tomotherapy plans for IMAT delivery. By using a 'bottom up' segmentation approach and clustering beamlets with similar weightings, the algorithm was able to convert tomotherapy plans to IMAT plans with only minor plan quality degradation. This method would have the potential to improve the plan quality if variable dose rates were allowed within the algorithm.

Cao *et al* (2006) developed a leaf-sequencing method called continuous intensity map optimization (CIMO) for converting the intensity distributions into deliverable segments for step-and-shoot IMRT delivery. They quickly applied the same technique to convert continuous

intensity maps optimized for 36 beam angles into deliverable arcs by finding aperture shapes and weights, so that the differences between the intensity created by the overlapping segments and the continuous intensity maps optimized in the plan optimization step are minimized (Cao *et al* 2007). The algorithm was tested for prostate, brain, head-and-neck and pancreas cases, and the results showed the overall superior plan quality as compared with IMRT using fixed beams.

Luan *et al* (2008) modeled the interconnectedness of the IMAT beam shapes and MUs using an aperture-based graph algorithm and sequenced continuous intensity maps using the k-link shortest path algorithm. This algorithm was tested for prostate, breast, head-and-neck and lung, demonstrating that the plans had rivaled helical tomotherapy plans. Based on the method developed by Luan *et al*, Wang *et al* (2008) sequenced the intensity patterns optimized for 36 beams into a single-arc delivery. In their approach, the geometric connectivity of all the apertures, designed for approximating the intensity distribution at a given angle, is ensured by using the coupled path planning algorithm. The geometric connectivity among the apertures derived from intensity distributions from neighboring angles is guaranteed by using the shortest path algorithm over the entire arc. They tested their two-step IMAT planning algorithm for brain, lung, prostate and head-and-neck cases and showed that the resulting single-arc IMAT combines the dosimetric advantages of rotational IMRT with speedy deliveries.

Other two-step approaches are reported by Bzdusek *et al* (2009) and Bedford (2009a). Although the works are performed independently, their approaches are very similar. The intensity profiles are first optimized on beams at static gantry angles evenly spaced over the range of one or more arcs. Initial aperture shapes are generated to approximate the optimized intensities. These apertures are then spaced evenly over the angular range to form a single arc, and their weights and shapes are further optimized using a direct aperture optimization algorithm, taking into account the allowed range of leaf motion of the MLC.

2.2. *One-step planning*

Realizing that the two-step process may produce a large number of complex field shapes and lead to inefficient treatment delivery and increased collimator artifacts, one-step planning was investigated by the research groups in the Ghent University Hospital and University of Maryland (De Gersem *et al* 2001, Shepard *et al* 2002). De Gersem *et al* developed a one-step planning method for step-and-shoot IMRT called leaf position optimization (LPO) (De Gersem *et al* 2001, Claus *et al* 2001). Instead of optimizing the fluence distribution with a subsequent MLC leaf segmentation in the two-step planning process, LPO begins an optimization with a set of MLC apertures first determined by the BEV of the target and its neighbouring critical structures. Using the simulated annealing approach, the leaf positions are optimized against a dose distribution. LPO also incorporates a segment weight optimization once the leaf sequence is optimized and finalized. This LPO algorithm was later adapted by Oliver *et al* for IMAT treatment planning (Oliver *et al* 2008, 2009).

In a similar approach, Shepard *et al* also developed a one-step planning method for step-and-shoot IMRT delivery called direct aperture optimization (DAO) (Shepard *et al* 2002). The aperture shapes and weights are simultaneously optimized using a simulated annealing algorithm. Physical constraints of the MLC, such as leaf movement limits, inability to interdigitate, and the minimal gap between opposing leaves and opposing adjacent leaves, can be conveniently considered in the optimization process. Only deliverable MLC shapes are considered and the need for leaf sequencing is eliminated. Without leaf sequencing, the number of apertures can be significantly reduced while maintaining the conformal capabilities of IMRT, considerably reducing the complexity of IMRT.

The efficiency advantage of DAO makes it ideal for planning IMAT. Following their successful development of DAO for step-and-shoot IMRT delivery, Earl *et al* (2003) applied this one-step plan optimization approach to IMAT treatment planning. Beams were equally spaced within the range of an arc. The task of DAO algorithm would be to optimize the beam aperture shapes at all the beam angles. In order to deliver the IMAT plan using constant dose rate and constant gantry speed, the common limitations of treatment machines at the time, all apertures were kept at the same weight. To ensure that the resulting IMAT arcs were deliverable, constraints were placed on the aperture shapes to make sure they did not differ significantly from one beam angle to the next. As it turned out, these constraints for both the geometry and aperture weights not only reduced the efficiency of optimization, but also limited its potential in terms of plan quality.

The possibility of varying dose rates during gantry rotation and irradiation affords greater freedom for direct aperture optimization. Using a DAO scheme as Earl *et al* (2003) but employing a more efficient way selecting the initial aperture shapes, Ulrich *et al* (2007) showed that IMRT-like dose distributions could be achieved with a single arc. When optimizing a large number of beam apertures from a large number of beam angles, the scheme by Ulrich *et al* (2007) and Earl *et al* (2003) can take a long time for the optimization to converge. Otto (2008) devised a coarse-to-fine DAO optimization scheme that starts with a small number of beams and large angular spacing, and gradually inserts new beam angles to be optimized. Geometric connectivity was facilitated by initializing the shapes of new apertures with shape interpolation between its neighbors and by constraining maximum leaf travel near the end of the optimization process.

Both the two-step and one-step optimization methods were successfully applied to planning IMAT treatments employing either multi-arcs or a single arc. The obvious efficiency advantages of single-arc delivery have encouraged linear accelerator vendors to offer different single-arc IMAT solutions. The clinical implementations of single-arc IMAT has shown that beam intensity modulation is not a fundamental requirement for achieving optimal treatment plans, as long as the optimizer is given enough freedom to take advantage of the angular and location preferences intrinsic to a given case (Webb and McQuaid 2009).

2.3. Dose calculation

Since the doses are calculated beam by beam in IMAT as with other techniques, there is nothing special about dose calculation in terms of statistical accuracy. On the other hand, due to the large number of beams used to approximate an arc, dose calculation poses new challenges both during the optimization process and in the final dose calculation, offering new opportunities. Tang *et al* (2008) compared a Monte-Carlo-based dose calculation with a collapsed-cone convolution/superposition-based dose calculation for arc deliveries. It was found that the calculation time for the Monte-Carlo-based algorithm is largely independent of the number of beams used, while the calculation time using all the empirical methods linearly increases with the number of beams. By comparing the calculation times needed for their home-grown stochastic kernel-based superposition approaches (Naqvi *et al* 2003) with the collapsed-cone convolution/superposition method, they concluded that if the number of beams is greater than 43, the Monte-Carlo-based superposition is faster than the collapsed-cone convolution algorithm. Depending on the plan properties such as aperture shape and weight variations, final dose calculation for IMAT plans may require a large number of interpolated beams. Thus, stochastic methods could offer the opportunity to calculate the doses as with the actual delivery, thereby eliminating the discrepancy between the calculation using the planned static segments and the dynamic delivery, as described in section 3.

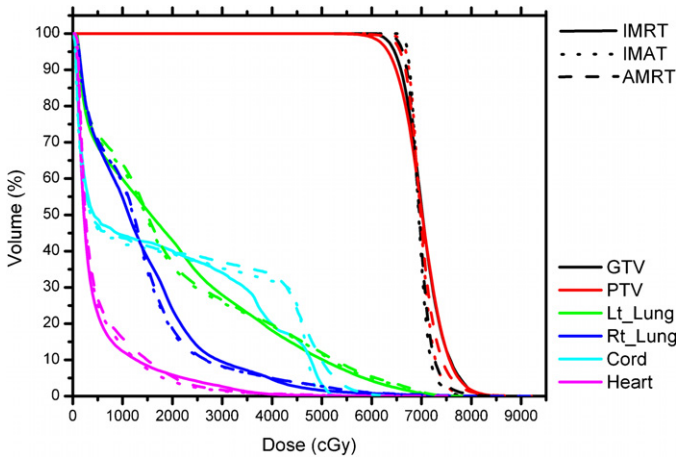

Figure 2. DVH comparison for a lung cancer treatment using IMRT with seven fields (solid lines), IMAT with multiple arcs (dotted lines), IMAT with one arc (AMRT, dashed lines).

2.4. Plan comparisons

Single-arc and multi-arc IMAT treatment plans using either the two-step optimization process or the one-step optimization have been compared with other IMRT methods, including tomotherapy, IMRT using multiple intensity-modulated fields and among other IMAT planning methods. Different investigators have conducted many such comparisons. This section proceeds to offer a few of the examples.

In a comparison of tomotherapy and MLC delivery, Mavroidis *et al* (2009) found that linear accelerator delivery with MLC has a slight advantage over tomotherapy for most sites other than the head-and-neck. Similar results have been found by Muzik *et al* (2008). Cao *et al* (2007) compared the treatment plan quality of IMAT plans and tomotherapy plans for ten cases including head-and-neck, lung, brain, and prostate. It was found that these two kinds of rotational delivery methods are equivalent for most cases. For cases where non-coplanar beams are desirable, such as for intracranial tumors and some head-and-neck cases, the use of partial non-coplanar arcs in IMAT was found to be more advantageous. Shepard *et al* (2007) compared IMAT plans with IMRT and found that the employment of rotational IMRT was advantageous for most of the cases. This finding was also supported by Cozzi *et al* 2008 where they observed clinically significant improvements to the normal tissue sparing in cervix uteri cases. Another planning study was done for prostate cancer patients that compared constant dose-rate VMAT, variable dose-rate VMAT and five-field IMRT plans (Palma *et al* 2008). The results demonstrated that the variable dose-rate VMAT plans produced the best results in terms of dose distributions and overall efficiency of (monitor units) MU usage.

Tang *et al* (2009a) conducted a plan quality comparison among IMRT, multi-arc IMAT and single-arc IMAT, which they referred to as AMRT. To maximally eliminate bias, they used the same optimization engine with the same objectives and the same dose calculation engine. A total of 12 clinical cases for four different disease sites including brain, head-and-neck, lung and prostate was used in the study. The largest difference among the three techniques was shown in one of the lung cases, for which the DVH comparison is shown in figure 2. It shows that multi-arc IMAT (dotted lines) is capable of creating better dose coverage of the targets and the best dose uniformity while giving lower dose to all four organs at risk. This advantage, although most noticeable for this case, did represent the general trend seen in all cases for all sites. The study demonstrated that when multiple arcs were allowed,

the aperture shape connectivity placed fewer constraints on the optimization, and the plan quality was the best among these three methods. The single-arc version of IMAT, AMRT, was able to generate plan quality in between multi-arc IMAT and IMRT with multiple fixed fields. However, the differences in plan quality among these three methods were generally smaller than that shown in figure 2 and the advantages of one method over the others were not clinically meaningful. The only meaningful advantage was that AMRT, the single-arc version of IMAT, could be delivered much more efficiently than multi-arc IMAT and IMRT using multiple intensity-modulated fields.

One may assume that the multi-arc IMAT optimized with 36 beam angles each with five to seven apertures in Tang's work should approach the ultimate plan quality for the given photon beams. Such assumption is also supported by the work of Cao *et al* (2007) comparing a non-commercial multi-arc IMAT with tomotherapy. Then, how could IMAT with a single arc and IMRT with seven fields create treatment plans not noticeably inferior in plan quality to the ultimate? The fundamental reason lies in the fact that there are many ways to take advantage of the intrinsic angular and location preferences in irradiating a target. This fact was referred to by Mohan as 'degeneracy' (Mohan 2009). With a reasonable optimization algorithm, the plan quality is often limited by the physics of photon dose deposition and little improvement can be gained with more complex intensity modulation or more aperture shape variations. On the other hand, greater freedom given to the optimizer always helps in pushing the limit of plan quality. Because the intensity modulation in tomotherapy is not subject to the geometric constraints as with IMAT, it should theoretically provide the best plan quality for the same coplanar angular arrangements. Indeed, in most comparative studies involving tomotherapy, tomotherapy plan quality rivals or exceeds others. Cases where tomotherapy plans showed inferior quality are mostly due to suboptimal implementation, such as the use of the same helical pitch and beamlet size at both ends along the patient axis, or its inability to use non-coplanar arcs when needed.

It is important to note that there are many other issues in addition to plan quality that are associated with different delivery techniques. These include the efficiency of planning, delivery, quality assurance (QA), the complexity and reliability of delivery, and the total MUs required to deliver the prescribed doses and the total leakage radiation received by the patient outside the target region.

3. Delivery

3.1. Control points

To understand the dynamic delivery of IMRT and IMAT, we must first understand how planning systems handle dynamic delivery and communicate with the linac. In a planning system, a dynamic delivery sequence is approximated with multiple 'segments', or 'sub-fields', each defined as the delivery of a fixed number of MUs with a fixed aperture shape at a fixed gantry angle. The use of these static sub-fields is also what the user observes on the planning system. In communicating with the linac for dynamic delivery, such segments are translated into a set of control points. In the simplest translation, the first control point always has a cumulative MU of zero, with all other variables, i.e. field shape, gantry, and collimator angles, set to that of the first planned segment. For dynamic delivery, the second control point has the cumulative MU of the first planned segment but with the aperture shape, collimator and gantry angles of the second planned segment. In essence, the first planned static segment is converted into the dynamic transition from the first and the second control points. The transition from the second to the third control point then delivers the MUs of the second planned segment, and

so on. Every pair of neighboring control points defines a dynamic delivery interval and the corresponding planned segment is just a static sample of this delivery interval. Note that small variations to this segment to control point conversion scheme exist among different planning systems. For example, the planned segment shape and angles can be placed in the middle of two control points such that the planned segment samples at the mid-point of a delivery interval.

In all dynamic IMRT deliveries including IMAT, the delivered MU serves as the independent variable. In 'sliding window' IMRT delivery, aperture shape is the only dependent variable that varies with the delivery of MUs. All the other variables, such as the collimator angle, the couch angle and the gantry angle, are set as constants for each radiation beam. In IMAT delivery, the gantry angle also varies from one control point to the next, so that the field shape formed by the MLC varies during simultaneous gantry rotation and irradiation.

By convenience, an arc is commonly approximated by evenly spaced static beams in a planning system. To give the planning system the maximum freedom to take advantage of the angular and positional preferences intrinsic to the planning geometry, the segment weights or the MUs assigned to different segments, are allowed to vary. The uneven segment weightings necessitate dose-rate variation during dynamic delivery. If the required dose rate is too high or too low for the linac, gantry rotation must be slowed down or sped up. The dose rate and the gantry speed are not specified by the planning system, but rather figured out by the delivery control system of the linac.

Existing linacs from different vendors have different capabilities and control mechanisms in coordinating the delivery. First, the MLC designs from different vendors are very different. Varian's 120 leaf MLC is mounted on a carriage and acts as a tertiary collimator below the conventional collimator jaws. During rotational delivery, the jaws as well as the carriage do not move. The range of motion for the leaves is also limited by the carriage design and the long distance from the source. For large targets, greater than 15 cm in the leaf motion direction, it often requires two arcs to provide full dose coverage. Because the leaves from opposing leaf banks can interdigitate, the Varian MLC is capable of creating multiple island apertures in the same beam angle, providing greater freedom to the plan optimizer for driving up the plan quality. The MLC on the Elekta linac is a secondary collimator with x- and y-back-up jaws placed under the MLCs. Each leaf can travel 12.5 cm across the center and 20 cm from the center. The back-up collimators always dynamically follow the MLC-shaped fields and lower the leakage radiation especially through the parked gaps. The MLC on Siemens' Artiste linac has 160 leaves without backup jaws. Each leaf can travel 20 cm across the center and 20 cm from the center thereby is capable of shaping any aperture within a 40 cm $\times$ 40 cm area. The maximum speeds of leaf motion are also different for MLCs from different vendors. They are 2, 2.5 and 4.0 cm s^{-1} for MLCs from Elekta, Varian and Siemens, respectively.

How the linacs from different vendors deliver IMAT plans is also different. For Varian machines before the latest TrueBeam™, the linac and the MLC are controlled by separate computers that interact with each other; the control points designed for treatment delivery are decomposed into two groups of control parameters. (1) The MLC positions, as a function of delivered MUs, are sent to the MLC controller. (2) The gantry angle, as a function of cumulative MU, is sent as a segmented treatment table to the linac control system, which translates the segmented treatment table into commands that control the dose rate and gantry speed during dynamic arc delivery. The latest machines (the TrueBeam™) are more integrated with both the MLC and the beam generation system controlled from the same 'supervisor', the computer that interpolates the control points of the plan and coordinated the entire delivery. For Elekta machines, the coordination between the MLC and the linac delivery is performed by the 'RT Desktop'. The methods of dose rate control are also different. For Varian linacs using

a gridded gun, the instantaneous dose rate change in their C-series of machines is performed by the dose rate servo using pulse dropping. These older systems do not vary pulse width. TrueBeam uses a much more sophisticated control means which includes all four mechanisms including varying the radiation pulse width, height, and repetition frequency, and dropping radiation pulses. For Elekta machines employing a non-gridded gun, the dose rate is achieved by varying the rate of the pulses and the dose from each pulse is fixed. As a result, dose rates can be varied among discrete levels. The dose rate is automatically calculated to be the maximum that can be used without exceeding any of the movement speeds of leaves, jaws and the gantry. If the dose to be delivered in a given interval is too high, the gantry and the leaves slow down accordingly.

3.2. The disconnect between planning and delivery

In all rotational deliveries including tomotherapy, the treatment plans are developed by approximating the continuous arc rotation with tightly spaced static beams. Therefore, there is an intrinsic disconnection between the treatment plan and the treatment delivery. For IMAT delivery, not only the beam aperture shape, but also the machine dose rate varies dynamically. Because the field shapes are changing dynamically, the optimized aperture shape only appears for a short instant during delivery. The beam aperture takes interpolated shapes for the majority of the time. This also means that the MUs optimized for a fixed aperture shape are actually delivered with different shapes at different angles from the planned ones.

The effects of such disconnection between planning and delivery have not been thoroughly studied. In a recent work by Tang *et al* (2008), the coarsely defined static beams were interpolated (into 720 beams) with a fine angular spacing of 0.5°, to more accurately approximate continuous rotation. A Monte-Carlo-based kernel superposition algorithm was used to compute the radiation doses for both the originally 36 planned beams and the 720 beams used as the surrogate of continuous delivery. It was found that for most of the treatment plans, the difference between the plans created with static beams and continuous delivery is minimal. However, for some plans, large differences were noted. Figure 3 shows the discrepancies between the planned DVHs and the delivered DVHs for two head-and-neck plans generated by two different methods for the same case. Plan 1, as in figure 3(a), shows large differences between the calculated dose with 36 fixed fields and the delivered dose with continuous arc delivery, simulated with interpolated field shapes and MUs to 720 beam angles. Plan 2, as in figure 3(b), however, shows little difference between the calculated and the delivered doses. Further studies of both the leaf travel histogram (figure 3(c)) and the MU distribution (figure 3(d)) show that the plan exhibiting large differences between planned and delivered doses is the one with large leaf travels and large dose-rate fluctuations.

This is essentially the same as the digital sampling problem. Both large variations in MLC aperture shapes and large dose-rate variations represent high spatial and temporal frequencies. To accurately calculate such treatment plans, tighter samples (i.e. more bean angles) are required in dose calculation to faithfully represent the treatment delivery. Therefore, treatment planning parameters not only strongly affect deliverability, but also the accuracy of delivery. Understanding the physical limitations of the linac and the MLC, and the intrinsic disconnection between planning and delivery, is essential for establishing effective QA programs.

3.3. Dose rate variation is unnecessary

Since the planned segments are evenly spaced and have different MU weightings, dose-rate variation is required during dynamic delivery. Therefore, variable dose rate should not be

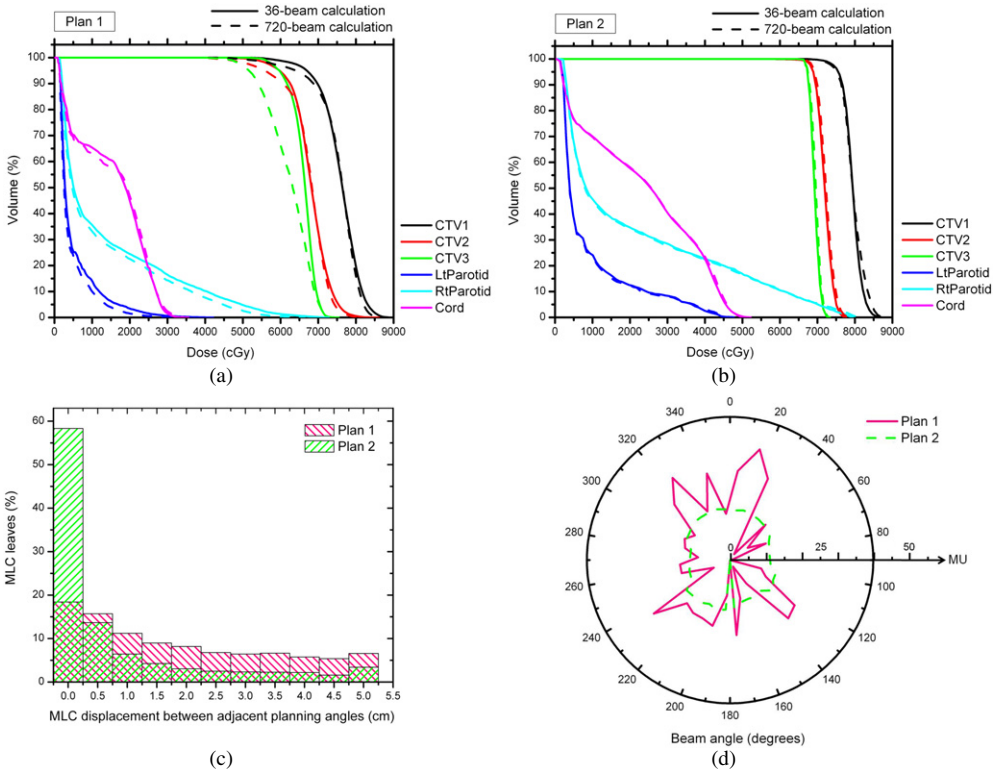

Figure 3. Effects of large MLC leaf travel and dose-rate variation on the accuracy of delivery. For the same head-and-neck case, large discrepancies between the planned and delivered doses are exhibited in (a) plan 1, but not in (b) plan 2. As compared with plan 2, plan 1 was found to have longer leaf travel as shown by the leaf travel histogram comparison in (c), and larger dose rate fluctuations as shown in (d) (from Tang *et al* (2008)).

viewed narrowly as the variation of machine dose rate but rather as the delivery of a different number of MUs within different evenly spaced angular intervals. This can be achieved by either (i) varying the machine dose rate at constant gantry velocity or, (ii) keeping the machine dose rate constant and varying the gantry velocity or (iii) varying both machine dose rate and gantry velocity. Although gantry speed variation is theoretically feasible, it is least desirable due to the large inertia of the linac gantry. Therefore, most of the aperture rate variations demanded by the treatment plan are achieved through varying the machine dose rate. Variable dose rate (VDR) delivery not only complicates delivery and QA, but also limits clinical adoption because most of the existing linacs are not equipped with VDR capability. However, restricting the segment weights to be constant limits the freedom of the plan optimizer and can lead to suboptimal plans. This is demonstrated by Palma *et al* (2008), as they found that VDR-optimized single-arc plans produced superior dose distributions to those optimized with constant dose rate (CDR). In their study, the treatment plans were generated using a series of evenly spaced static beams, which is the general approach for IMAT planning.

Based on the observation that the dosimetric error introduced by displacing the beam apertures from the planned angle to a slightly different angular position is minimal (Crooks *et al* 2003, Tang *et al* 2007, Wang *et al* 2008), Tang *et al* (2009b) hypothesized that varying

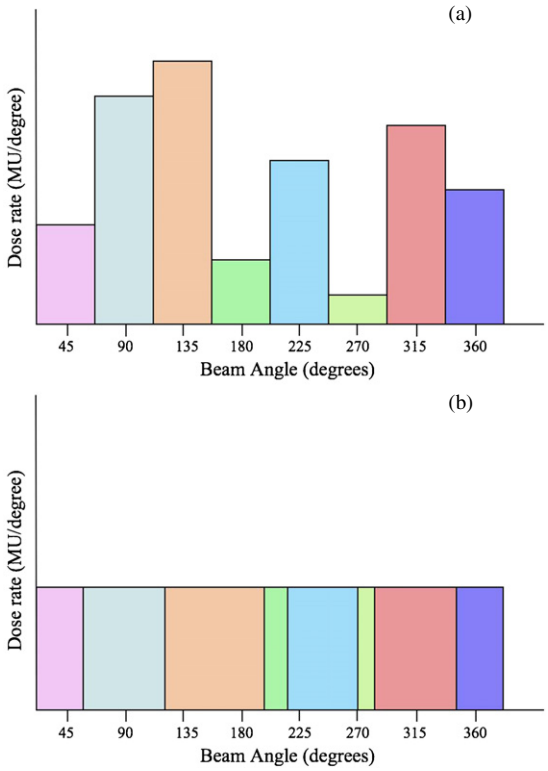

Figure 4. Variable dose rate with even angular spacing (a) is converted to constant dose rate with variable angular spacing (b) by conserving segment MUs.

the angular spacing of the apertures using CDR delivery is equivalent to varying the weights of evenly spaced beams. A similé can be drawn from radio broadcasting, where VDR delivery of evenly spaced beams resembles amplitude modulation while CDR delivery of unevenly spaced beams resembles frequency modulation. They proved such equivalence by converting RapidArc plans, which require VDR delivery, into CDR plans by assigning larger angular intervals to segments with larger MUs, and vice versa. The conversion scheme is schematically illustrated in figure 4, where the vertical axis represents the dose rate in MU/degree and the horizontal axis depicts the beam angles. Each vertical bar represents a planned segment. VDR and CDR counterparts of the same segment have the same area indicating the same segment MUs. The completed CDR plans were delivered and dosimetrically verified using a conventional linac without the capability of dose-rate variation. They found that the plan qualities and the delivery times of the CDR and VDR plans were comparable, which proves that single-arc IMAT can be delivered using either VDR with even angular spacing, or CDR with variable angular spacing.

4. Clinical implementation

IMAT is a rotational form of IMRT. With nearly 15 years clinical experience in the use of IMRT including tomotherapy for different sites, clinical implementation of IMAT is much

less challenging than starting a new IMRT program for the first time. Nevertheless, there are differences between IMRT using limited fixed beam angles and IMAT. Understanding these differences is important to ensure a smooth and successful clinical implementation.

4.1. Acceptance and commissioning

There have been extensive guidelines on the commissioning and QA of treatment planning systems used for radiation therapy (AAPM TG 53, 1998) but limited publications on the commissioning and QA of IMAT (Ling *et al* 2008, Bedford and Warrington 2009c). The same basic requirements for dosimetric accuracy hold: the calculated doses must match the measured ones for both homogeneous and heterogeneous geometries. The processes used to ensure such accuracies are also similar.

IMAT delivery often requires more advanced linac control capabilities, including variable dose rate, variable gantry speed and dynamic MLC movement. When all the components required for implementing IMAT are acquired and a technical chain is established, the first step is to verify the reliability and accuracy of the entire chain from planning to delivery and to assess the technical limits. This step ensures that the intended machine is capable of precisely making the planned variations. Ling *et al* (2008) have demonstrated a method of verifying control accuracies in a stepwise manner. All delivery systems have some provisions for handling treatment interruptions and abnormal terminations. It should be examined to see if such interruptions or terminations cause dosimetric errors.

As with IMRT, the geometric and dosimetric characteristics of the linac must be put into the planning system and modeled using the commissioning tools of the treatment planning system. Although single-arc IMAT has no intensity modulation at each beam direction, beams from neighboring angles overlap at the target. The doses of an arc plan are calculated by summing the calculations of individual beams. Therefore, geometric errors in MLC positioning can translate to large dosimetric errors by the same mechanism as with fixed-field IMRT (Budgell *et al* 2000, Mu *et al* 2008, Rangel and Dunscombe 2009). The under-dose effect of the tongue and groove design of the leaf sides should also be accurately modeled. In many cases, the IMAT planning capability is an added software module to an existing planning system. The dose calculation algorithms used for IMAT planning and for the existing IMRT module are identical. In such cases, there is typically no need to repeat the machine data modeling. However, it is prudent to carefully verify that the planned doses are delivered accurately. The dose calculation model for the fixed beams may not accurately reflect rotational delivery due to the lack of adequate sampling as discussed previously. Another culprit is how the planning system handles the treatment couch. Vanetti *et al* (2009) found that for the 'IGRT couch' made of radio-transparent foam encased by a carbon fiber shell, ignoring it in planning can cause a dose overestimation of 1.5 Gy in a prostate treatment with a 70 Gy prescription.

It should be emphasized that the planning and delivery of IMAT is more tightly integrated than IMRT due to the additional requirements on the linac control. As with IMRT, the planning and delivery systems can be from different vendors. Therefore, it is important that the delivery systems, i.e. the linac and the MLC, have the capabilities required by the planning system. For IMAT planning systems designed specifically for machines from a particular manufacturer, considerations are already made on the maximum gantry speed and the maximum speed of MLC leaf motion. For IMAT planning systems intended for linacs from different vendors, different mechanical and dosimetric constraints must be entered. A caution is that the maximum speed of MLC motion given by the linac vendors is typically a dynamic value. In reality, MLC leaves may need to change direction and accelerate, resulting in the actual achievable speed lower than such a maximum. The ability and speed of dose-rate

variation is another consideration. An effective planning system should be able to utilize all the freedoms provided by the delivery system but without stretching the machine's capabilities, and to optimize the plans under the physical constraints of the delivery system.

As a final proof that the technical chain is ready for the clinic, a set of planning and delivery exercises must be conducted to show dosimetric accuracy under different delivery conditions and for different clinical sites. This process forces the entire technical chain to be used and verified. This is also a time to practice and learn how to use the different components. This step is achieved normally with phantom studies, where different clinically challenging arrangements of target and sensitive structures will be planned and delivered on the phantom.

For beam collimation systems in which the back-up jaws cannot follow the MLC-shaped apertures, the gap between the closed opposing leaves outside the aperture gives additional leakage radiation. If the collimator angle is fixed at zero, leaving the gap parallel with the rotational axis, the leakage radiation from the gap will be focused on the rotational axis or form a cylindrical high dose shell around the rotational axis outside the target. One way to spread the leakage dose is to set the collimator angle to 45°. Webb (2010) demonstrated that if the collimator angle is also included as an optimization parameter, additional freedom can be gained for improved deliverability, leading to a reduction in the number of parked gaps and unwanted leakage dose. Providing such a freedom to the IMAT plan optimizer may also alleviate some physical constraints on aperture shape connectivity.

4.2. Quality assurance

In addition to the complexities in the planning and delivery of IMRT, IMAT introduces continuous gantry rotation and dose-rate variations, both of which cause the optimized MUs to be delivered at unintended beam angles. Because the optimized MUs are never intended to be delivered at the planned beam angle in IMAT, the dose deposition is fairly tolerant to small errors in the gantry angle. As long as the intended MUs assigned to each of the segments are delivered accurately—a proven reliable functionality of all modern linacs—IMAT delivery is less susceptible to the errors in dose rate and gantry speed. For all IMRT deliveries including IMAT on machines from different manufacturers, the delivered MUs are treated as an independent variable, upon which the other variations are enslaved. As a result, IMAT deliveries are not necessarily less reliable or more susceptible to delivery errors than IMRT using fixed fields. Therefore, similar procedures used for IMRT QA can be used for IMAT QA.

Like IMRT, the geometric errors of MLC positioning can also lead to large dosimetric errors in IMAT delivery. The emphasis of machine QA for IMAT delivery should also be placed on the geometric uncertainties during dynamic delivery. Most institutions have adopted some form of routine procedures for assessing the MLC positioning accuracy. These include monthly delivery of multiple abutting segments or multiple thin lines (the picket fence pattern) (LoSasso *et al* 2001). Replacing such a routine QA sequence with a new sequence of control points, allowing similar patterns to be delivered during gantry rotation, would be adequate for machine QA of IMAT delivery.

Since IMAT is a form of IMRT, patient-specific QA is an integral part of the clinical practice. It should be emphasized that the planning and delivery of IMAT is more tightly integrated than IMRT due to the additional requirements on the linac control. Patient-specific QA allows the complete technical chain from the IMAT treatment planning system to the specific IMAT delivery system for which the plans are intended to be tested.

Most methods used for patient-specific QA of IMRT have been adapted for IMAT QA. These include applying a treatment plan to a phantom to allow the calculated and the

measured doses to be compared (Vergote *et al* 2004, Van Esch *et al* 2007, Bedford *et al* 2009b, Létourneau *et al* 2009, Haga *et al* 2009), the use of an electronic portal imaging device or other detectors to measure and compare the portal dose distributions (Nicolini *et al* 2008, Iori *et al* 2010), recreating dose distributions using recorded delivery control files (Schreibmann *et al* 2009, Teke *et al* 2010), and perform an independent dose calculation with Monte-Carlo methods (Li *et al* 2004, Bush *et al* 2008). IMRT QA based on field fluence measurements was also proposed for patient-specific QA of IMAT treatments by delivering all fields at a fixed gantry angle and comparing to the calculated cumulative fluences (Iori *et al* 2007). Due to the lack of rotational information, such methods may hide individual errors, making the verification less capable of serving its intended purpose. However, if the cumulative intensities from gantry-mounted detector arrays, such as MatriXX^{TM} detector (IBA Dosimetry Gmbh, Schwarzenbruck, Germany), and the DAVID^{TM} multi-wire chamber array (PTW, Freiburg, Germany), are read out frequently to obtain the beam intensities delivered in different angular intervals, the measured fluences can be compared with the calculations from the planned aperture shapes and weights, over the same angular intervals, to effectively serve the purpose. Such measured mean intensities can also be applied to the original patient images to obtain dosimetric comparisons in the patient, provided that such angular intervals are not too large ($\leqslant 10^{\circ}$), by assuming identical patient positioning between imaging and treatment delivery.

One of the widely used methods for IMRT dosimetric verification is the application of the treatment plan to a phantom, allowing a comparison between the calculated and the measured doses to take place. Since the actual treatment or a variation of the actual treatment is delivered to the phantom, such experiments are also called 'dry-runs'. The dry-runs not only verify the treatment dosimetrically, but it also ensure that the prescribed dynamic MLC delivery can be carried out. The wide clinical adoption of rotational IMRT also prompted new commercial phantom developments. For example, doses can be measured and compared on a cylindrical plane (the ArcCheck^{TM} phantom, Sun Nuclear Corporation, FL, USA) or on two perpendicular planes (the Delta^{4} ^{TM} Phantom, Scandidos, Uppsala, Sweden).

It is important to note that because most of the dose verification phantoms are homogeneous in nature, agreements between the calculated and the measured doses do not mean that the intended doses are actually delivered to the patient. One of the difficulties in IMAT planning is the calculation of a large number of radiation beams. When using empirical dose calculation algorithms, the total dose calculation time is proportional to the number of beams. In single-arc IMAT, hundreds of beam angles are typically used. To complete the dose calculation in a reasonable amount of time, manufacturers are forced to take shortcuts. Therefore, it is important that inhomogeneous phantoms are used, at least in the initial commissioning process.

Independent dose or MU calculation can also be used as a way of patient-specific QA after gaining confidence with a large number of phantom measurements. Such dose verification would have to reconstruct the plan by inputting data from the treatment prescriptions, or from DICOM-RT control points sent to the machine for delivery. In comparison, the use of treatment delivery log files to recreate the dose distribution either on the original patient CT images or on the CT images acquired with an on-board or in-room CT provides a more stringent verification.

5. Discussion

The way IMRT, including IMAT and tomotherapy, spares critical structures is by redistributing the normal tissue dose to less critical regions and reducing the high dose volume to cover just

the target. For a given integral dose to the target, the integral dose to the surrounding structures is roughly constant, as dictated by the physics of dose deposition. IMAT and other rotational treatments deliver lower doses to a greater volume of surrounding normal tissues as compared to treatments employing a limited number of fields. This phenomenon is often inappropriately labeled 'dose dumping', disregarding the fact that, given the same integral dose to the normal tissues, delivering a lower dose to a larger volume is better than delivering a larger dose to a smaller volume for most disease sites. Nevertheless, volume considerations must be carefully given for parallel organs, such as the lung, and for pediatric applications.

A single rotation could theoretically contain hundreds of aperture shape variations, enough to achieve the needed modulation for taking advantage of the angular and positional preferences dictated by the geometry and dosimetric objectives. In reality, however, the apertures cannot be independently shaped because of the physical constraints of MLC leaf motion and gantry rotation. For certain cases, adding a second arc could provide the planning system additional freedom for achieving better treatment plans. This is especially true for MLCs that cannot (geometrically) cover a large target with one dynamic sequence. For tumors that are not centrally located, and for tumors surrounded by parallel organs, the use of partial arcs is often desirable. For tumors in the brain and in the head-and-neck region, it is often advantageous to use multiple non-coplanar arcs for better tumor targeting and critical structure avoidance.

The capability of IMAT in creating highly conformal dose distributions, especially in the single-arc form, has been a subject of debate (Bortfeld and Webb 2009, Mehta *et al* 2009, Ling *et al* 2009, Mohan 2009). In principle, the more freedom we give to the planning system, the better quality of plans can be generated. However, with photon beams, there is a limit to the treatment plan quality. This limit is often not set by the degree of intensity modulation but by the physics of photon dose deposition. For the vast majority of clinical cases, limited intensity modulation using fixed beams or the use of a single-arc rotation is able to take the angular and positional preferences intrinsic to the given geometry into account. IMAT uses a large number of beams and aperture shape variations. Complex intensity modulation is achieved at the level of the target where the beams overlap rather than at the collimator level. The means by which single-arc IMAT achieves the optimal dose distribution is therefore the same as with IMRT. Without understanding these principles, it is easy to draw the wrong conclusions that the use of less modulated beams, such as those IMRT plans created with direct aperture optimization, or the use of a single arc, can only be suitable for simple clinical cases, such as the treatment of prostate cancer. On the other hand, there is also a limit on how simple a plan can be without compromising plan quality. If a single-arc IMAT plan is optimized using less than 50 aperture variations, i.e. with a limited number of strata, it would be unrealistic to achieve high plan quality for complex clinical cases. Comparisons based on single-arc IMAT plans of such simplicity should not represent the true potentials of IMAT.

6. Conclusion

Significant technological developments in treatment planning and linac design have led to the eventual clinical adoption of IMAT, a technique initially proposed in 1995. As these solutions mature, IMAT will reach its full potential in both plan quality and delivery efficiency. This review should provide the readers an understanding of the principles of IMAT as well as the available technologies for its planning and delivery. Clinical implementation and QA mostly mirror that for IMRT using limited fixed fields and the delivery should not be less reliable with the involvement of gantry rotation and dose-rate variations. The efficiency of IMAT will find an increasing role in the practice of radiation therapy.

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