

Review article

IMRT: delivery techniques and quality assurance

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Abstract. Intensity modulated radiotherapy (IMRT) is a major development in the delivery of radiation therapy that has the potential to improve patient outcome by reducing morbidity or increasing local tumour control. Delivery techniques include those based on purpose built devices and treatment machines together with those utilizing the capabilities of computer controlled multileaf collimators which are more widely available. The complexity of IMRT techniques demands a high level of quality control both in the operation of the equipment and in the delivery of treatment to individual patients. The purpose of this paper is therefore to review the techniques available, concentrating on the use of multileaf collimators, and to consider the necessary quality control requirements for clinical application. It demonstrates that the technology is mature and sufficiently well understood so that IMRT can be safely implemented in the general clinical environment rather than being limited to application in the research environment.

The need for intensity modulated radiotherapy (IMRT) arises from the requirement to sculpt precise dose distributions which conform in three dimensions to the shape of planning target volumes (PTVs) and which avoid organs at risk (ORs). In many, maybe the majority, of cases acceptable conformal dose distributions can be produced by directing a small number of beams, each shaped to match the projection of the PTV, and weighting the fluence from each beam so that the dose within the PTV is acceptably uniform and the dose to the ORs is acceptably low. Irradiation of ORs can be avoided first by judicious choice of beam direction, second by the accurate shaping of each beam and third by reducing the fluence of beams which cannot avoid the OR. A treatment plan is considered acceptable if the dose within the PTV is uniform to within predefined limits, typically $\pm 5\%$, and the doses to ORs are lower than those which would cause unacceptable normal tissue complications.

In those cases where the summation of the doses from the selected uniform beams does not result in an acceptable dose distribution, improvements can be achieved by the use of wedges and or compensators, attenuators placed in the beam to modify the dose distribution normal to the beam direction. In the case of a wedge the variation in dose distribution is one dimensional, monotonic and usually designed to produce a constant gradient across the beam. Compensators, often designed for individual patients, are two dimensional attenuators, which can produce arbitrary dose distributions as required for an individual treatment plan. Although wedges and compensators produce intensity modulations, their use has in most cases been limited to the correction of unacceptable dose gradients in the PTV, for which wedges provide a solution, and the correction of unacceptable dose variations caused by variations in the thickness of overlying tissue, for which (missing tissue) compensators provide a solution.

Contemporary developments of IMRT have been predicated on developments of the “inverse problem” in which the analogy is drawn between CT scanning, where

the back projection of a series of intensity profiles can reconstruct an image of structures within a patient, and radiotherapy where the projection of a series of intensity profiles construct an arbitrary distribution of absorbed dose. The attraction of this approach is that it is possible to generate dose distributions to conform to targets which are concave in the plane of the incident beams, a result which is not possible to achieve with uniform coplanar beams of high energy X-rays.

The purpose of this paper is therefore to review the delivery techniques that are available for the generation of intensity modulated beams and consider the requirements for quality assurance for these techniques which are significantly more complex than those used for conventional radiotherapy.

Techniques available

The purpose of IMRT is to produce a three dimensional dose distribution within a patient. In principle beams of radiation can be projected from any direction but initially, to distinguish between two separate classes of techniques, only coplanar beam directions will be considered. It is also necessary, for this discussion, to distinguish between a technique, an arrangement of beams which when added together produce the required three dimensional dose distribution, and a delivery method describing the way in which the radiation from a particular beam direction is modulated.

A technique consists of a series of beam directions each defined by the linear accelerator gantry angle. From each beam direction a modulated beam is projected towards the isocentre. The beams can be narrow fan beams, modulated in one dimension or divergent cone beams modulated in two dimensions.

Tomotherapy

The rotation of a series of fan beams around a patient generates a dose distribution within a slice, analogous to the slice thickness of a tomographic scanner; hence this

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technique has been called tomotherapy. In order to get a three dimensional representation of a patient from a CT scanner is necessary to image multiple slices. Similarly, in order to build up a three dimensional dose distribution in tomotherapy it is necessary to irradiate a series of slices between which the patient is moved by the slice width along the axis of rotation. In CT scanning the axial resolution is limited by the slice thickness, similarly in tomotherapy the axial resolution of the dose distribution is limited by the slice thickness. The parallels between CT scanners and tomotherapy treatment machines can be seen further in the ongoing development of the technology. The first applications of tomotherapy being delivered one slice at a time, requiring accurate indexing of the patient between slices. The more modern development of helical tomotherapy in which the patient is moved continuously though the rotating fan beam can be compared with spiral (strictly helical) CT.

Tomotherapy, delivered by either indexed single slice irradiation or by the more advanced helical method, requires complex equipment which is not widely available. The technical details of these delivery methods will not therefore be discussed any further. However, it has been noted [1] that at the turn of the century more IMRT treatments had been carried out by tomotherapy than by any other technique. While no recent statistics are available to this author it is likely that the use of tomotherapy still represents a significant proportion of IMRT worldwide.

Physical compensation

Physical compensators [2] have been used for many years to modulate the fluence incident on the surface of a patient in order to produce a uniform dose at depth in plane normal to the beam. Compensation allows for the variations in depth due to the surface shape and in some cases variations in effective depth due to the presence of inhomogeneties between the surface and the target plane.

The extension of the use of such two-dimensional attenuator to produce specific non-uniform intensity modulations have been investigated by many groups [3–5] and such devices should perhaps be described as dose optimization filters [6] rather than compensators.

Although conceptually simple, compensators have limitations in both their design and manufacture. Account must be taken of the overall effects of placing the two-dimensional attenuator in the beam. These include hardening of the primary beam and the generation of scattered radiation which is added to the radiation incident on the patient. Simple formulae have been derived to calculate the optimal thickness profile for compensators from the primary fluence profile [7] and it has been shown that the presence of the compensators usually has an insignificant effect on the depth dose characteristics of the modulated beams even though the primary photon spectra can be significantly altered. Limitations in manufacture arise from the practical considerations of the time required to machine a compensator or mould with very high spatial resolution and ultimately on the physical size of the cutters used [8, 9]. These determine the smoothness of machined slopes and the minimum radii of curvature for valleys and ridges in the required profiles. However these physical limitations are in general much smaller than the equivalent

limitations in beam profiles caused by the finite width of multileaf collimator (MLC) leaves. A further limitation in the use of a physical compensator is to the minimum fluence that can be achieved with practical thickness of attenuating material. However the practical minimum of approximately 5% is probably low enough for most applications of IMRT.

These limitations are often cited as arguments against their use, but compensators do have several advantages. A significant advantage is that the irradiation of a patient through a compensator can be accomplished with a single exposure thus avoiding problems of match-lines (including the tongue and groove effect) and as all parts of the field are irradiated simultaneously problems of patient movement are minimized [10]. Furthermore, as long as the minimum thickness of the compensator is zero, corresponding to attenuation at the point requiring the maximum fluence, the use of a compensator is extremely efficient.

Examples of the many applications of physical compensators for IMRT include head and neck treatments designed on the basis of fluence distributions generated by Konrad (Konrad is an inverse planning program developed at DKFZ Heidelberg), the inverse planning program used in conjunction with the Plato (Plato Nucletron BV, Veenendaal, The Netherlands) treatment planning system [11] and a five field IMRT technique in which five compensators were designed and mounted in the correct orientation on a single rotating turret [12]. This mechanical arrangement allowed a degree of automation in the positioning of the appropriate compensator for each beam, reducing the manual handling problems associated with the use of collimators and aiding their precise location relative to the collimator axis. Some more conventional applications of physical compensators, for example compensators designed to improve the uniformity of breast irradiation [13] appear in the IMRT literature. Clearly the line between conventional compensation and IMRT is arbitrary and classification of a particular technique is best made on the basis of the planning method used to determine the fluence. Any inverse calculation starting with a prescribed dose distribution in a patient and resulting in a fluence distribution for which a compensator is designed can reasonably be classified as IMRT.

Finally, although physical compensators can provide intensity modulation of beams used in IMRT, they have to be used in conjunction with other devices to define the bounding shape of those modulated beams. Beam shaping can be achieved with an MLC, using this well proven technology for its primary purpose while avoiding the complexity of using it for intensity modulation. Alternatively beam shaping can be carried out with customized shaping blocks, which can be cast to include the compensation material, in which case IMRT can be carried out on any simple linear accelerator, or even a telecobalt unit [14].

Multileaf collimation

There are two general methods of modulating a beam with a MLC. The first is based on the sequential exposure of sub-beams or segments for which the collimators are (automatically) positioned while the radiation beam is switched off. After irradiation of each segment the

Table 1. The sequence of events necessary to deliver a modulated beam by the segmented multileaf collimator (SMLC) and the dynamic MLC (dMLC) methods

SMLC	dMLC
Move MLC to define 1st segment Irradiate with M_1 monitor units Stop irradiation Move MLC to define 2nd segment	Move MLC to 1st control point Irradiate with M_1 monitor units during movement of MLC to 2nd control point
Irradiate with M_2 monitor units Stop irradiation Move MLC to define 3rd segment ↓ ↓ Move MLC to define n th segment	Irradiate with M_2 monitor units during movement of MLC to 3rd control point ↓
Irradiate with M_n monitor units Stop irradiation ↓ ↓	Irradiate with M_n monitor units during movement of MLC to $(n+1)$ th control point ↓

collimators move to the correct positions for the next segment and so on until the total modulation has been achieved. This method is known colloquially as the “step and shoot” method but is otherwise known as the multiple static field (MSF) or segmented MLC (SMLC) method. The use of “SMLC” has been recommended by the Intensity Modulated Radiation Therapy Collaborative Working Group in the USA [15] and is likely to become the common term.

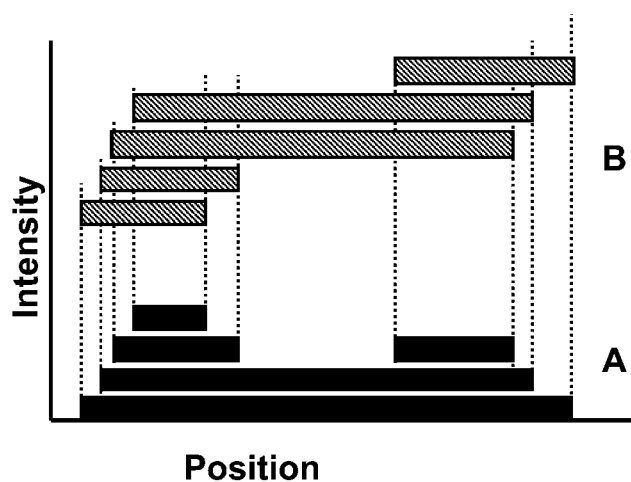
The second method is based on continuous irradiation during which the collimators move according to a predetermined trajectories designed to give the desired modulation. This is the dynamic MLC (dMLC) method. A comparison of the sequence of events needed to deliver a modulated beam is given in Table 1. In each case the collimators move to a series of control points which define the position of each of the collimating elements (leaves and back up collimators) after a particular fraction of the total exposure has been delivered. There are clear similarities between both methods but it is convenient to discuss each of them separately in the following sections.

Multiple static fields – segmented MLC delivery

The use of MSFs from the same beam direction is conceptually and practically the simplest method of intensity modulation with an MLC. From a regulatory point of view segmented delivery is merely an extension of conventional radiotherapy, from the irradiation of a patient with a few beams, to the irradiation with a very much greater number of small beams. However there are many technical challenges in both the design of the segments to reproduce the desired modulation and in the delivery which has to take account of constraints imposed by the MLC and the linear accelerator.

Before discussing the technical issues in detail it may be useful to demonstrate schematically how a one dimensional intensity distribution can be delivered by a series of segments using one pair of leaves of an MLC. The segments, “A”, shown in Figure 1 are chosen to be an acceptable quantised approximation of the required distribution which would in general be continuous.

The simplest way to deliver these segments is to deliver the widest segment first, then the second. On reaching the valley the widest segment in the left peak is delivered

**Figure 1.** Two sets of segments to reconstruct a quantised intensity distribution.

followed by the highest segment. The collimator could then be moved to deliver the segment comprising the right peak. This sequence represents the “close in method” which is simple for uni-modal distributions but for a bi-modal distribution requires the leaves defining the right hand edges of each segment to be driven into the beam for the first peak, reversed so that the second peak can be irradiated and then driven into the beam a second time to shape the right hand edge of the second peak. So in terms of efficiency of leaf movement the close in method is less than ideal.

A second method in which both leaves move continually in the same direction is called the sliding window or leaf sweep method. In this case sequence of segments “B” is generated by defining the positions of the steps in the quantised distribution, shown by the vertical dotted lines in Figure 1, and forming segments bounded by the first rising step and the first falling step then the second rising step and the second falling step and so on, until the last segment which is formed by the last rising step and the last falling step. Inspection of the segments shown in “A” and “B” shows that, at all positions, the total intensity delivered by the five segments in each of sequences is identical.

These two sequences have identical efficiency in terms of

the number of monitor units required for delivery but the sliding window sequence requires less movement of the leaves and hence the total treatment time including the time to move the leaves is less. The intensity can be scaled in terms of the number of monitor units that need to be delivered and so the total number of monitor units are those required to deliver the highest peak plus the difference between the valley and the second peak. "Close in" and "sliding window" are just two of many sequences which can reconstruct a quantised distribution. In a bi-modal distribution with P_1 steps in the first peak, P_2 steps in the second peak and with a local minimum at the intensity level of the M th step it can be shown [16] that the number of possible sequences is: $P_1! \cdot P_2! / M!$. In this simple example $P_1=4$, $P_2=3$ and $M=2$ so the number of sequences is 72, a figure that soon escalates of the numbers of steps increase to those more typically found in practice.

Many leaf sequencing algorithms have been developed some of which are included in commercial planning systems with IMRT capability. The prime requirement for a leaf sequencing algorithm is that it faithfully reproduces the fluence profiles generated by the planning process. A second equally important requirement is that the algorithm takes account of constraints imposed by the design of the MLC which will be used to deliver the sequence. These include constraints on the minimum separation of leaves and the ability (or not) of the leaves to interdigitate, that is for a leaf in one bank to pass between adjacent leaves on the opposing bank. Rigorous mathematical analysis of leaf sequencing algorithms for segmented multileaf collimation [17] shows that unidirectional sequences are always as "monitor unit efficient" as bi-directional sequences and are more "movement efficient". The extension of the sequencing problem to the decomposition of two dimensional dose distributions is straightforward, although the complexity of the algorithms is greater and other effects have to be taken into account.

In the one dimensional case described above the segments were formed by dividing the profile into horizontal strips of variable length and uniform intensity. The profile could have been divided vertically with a view to delivering a series of contiguous narrow beams of variable intensity, indeed the MIMiC device (Nomos Corporation, Sewickley, PA) segments profiles in this way (although the actual delivery of the segments is simultaneous with the intensity modulated temporally by opening and closing the individual attenuating vanes). Delivery of abutting segments sequentially is subject to matchline problems and specifically an effect known as the tongue and groove effect illustrated in Figure 2. When a segment is formed by a MLC the vertical edges will be defined by ends of opposing leaves and the horizontal edges by the sides of leaves. Most MLC designs have either "tongue and groove" or "stepped edges" so that when leaves are together they overlap and interleaf leakage is substantially reduced. Figure 2 shows this overlap and the small restriction in the aperture formed, the projection of this overlap in the isocentric plane is typically 1 mm. If the regions A and B are irradiated as part of the same segment the dose across the junction region, shown by broken lines, is uniform. However, if region A is irradiated as part on one segment and region B is subsequently irradiated as part of a second, then the area between the broken lines will be partially shielded on both occasions. The minimum

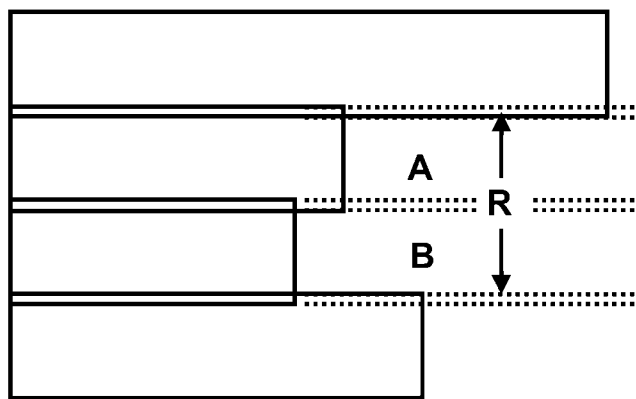


Figure 2. Diagram to explain the tongue and groove effect. A and B are regions that could be irradiated as abutting segments. R is the restricted aperture caused by the leaf overlap.

dose in this overlap region has been measured [18] to be approximately 80% of that intended. Fortunately there are leaf sequencing algorithms [19] which include the possibility of the tongue and groove effect as a constraint and generate synchronised sequences of segments which minimize or avoid the effect.

Clearly the accuracy of reproduction of a quasi-continuous fluence profile depends on the number of levels in which the profile is quantised. Intuitively, 10 equal levels will result in an accuracy of $\pm 5\%$ of the maximum intensity for a single modulated beam but as most treatments are delivered as a superposition of beams from several directions it is often possible to achieve an overall accuracy of $\pm 5\%$ with significantly fewer levels and therefore fewer segments. However, it is not necessary to restrict the intensity of each segment to be the same and several algorithms have been developed to convert two dimensional profiles into the minimum number of variable intensity segments necessary to achieve the required accuracy. The areal segmentation method [20] looks for areas with intensity above approximately 50% of the maximum intensity and creates segments of those shapes. These are subtracted from the target profile and the remainder is examined to find areas with intensity above 25% of the original maximum and so on. This binary decomposition will match the profile to within $\pm 1\%$ with only 6 levels (but often many more segments as some of the levels will have isolated segments which need to be irradiated separately). An alternative approach to variable fluence step segmentation is to use a clustering algorithm [21] to determine the intensity levels at which the profile will be segmented. A cluster is defined as all elements of the two dimensional fluence (bixels) within a certain deviation from the mean intensity of that cluster. Hence a cluster is characterized by its mean level and by its bandwidth so clusters of different bandwidth will give rise to segments of different intensity. Clustering algorithms create segments of low intensity in regions of low gradient and segments of higher intensity in regions of high gradient which produces an effective trade off between efficiency (small number of segments) and accuracy.

All segmentation algorithms need to take into account the "hard" mechanical constraints of the MLC to be used. However other factors have to be taken into account. These include a detailed consideration of the radiation

transport characteristics of the MLC which, ideally, will be accurately modelled by the treatment planning system. This is particularly important if, as is often the case, the optimization of the intensity modulated beams and their segmentation are carried out separately as shown in Figure 3. The final dose calculation, projecting the segments onto the patient, will in general result in a slightly different dose distribution from that calculated during the optimization. For example, if the segmentation results in many small fields the integrated dose delivered to a particular point in the patient includes leakage radiation from all segments which do not directly irradiate that

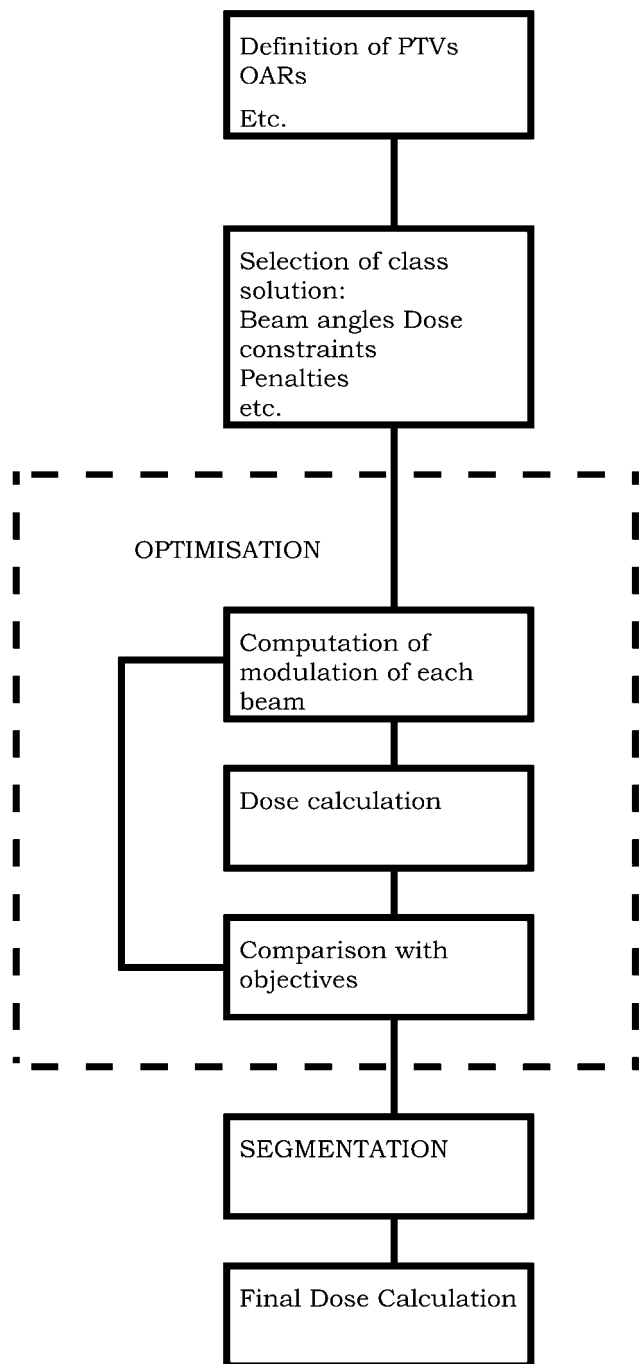


Figure 3. Flow diagram for the planning of segmented intensity modulated radiotherapy. PTV, planning target volume; OAR, organ at risk.

point. If, on the other hand, the segmentation results in fewer fields which do not directly irradiate the point then the integrated leakage will be less and the dose will be lower. Similar discrepancies are caused if segmentation results in differences in head scatter from that accounted for during optimization. This problem is avoided if the segmentation step is included in the iterative optimization loop [22], if not the final dose calculation might indicate that re-optimization, perhaps with slightly different objectives or constraints, is necessary.

Segmentation can result in sub-fields which are small both in terms of their dimensions and the number of monitor units required for delivery. The dosimetry and modelling of geometrically small fields demands particular attention as the variation in output factor (μcGy^{-1}) as a function of fields size and shape is very high. A 2 mm change in a field of approximately $10\text{ mm} \times 10\text{ mm}$ can change the output factor by up to 16% [23], an observation which has implication for the precision of control of the MLC. Ideally the number of small fields should be limited but if this is not possible care should be taken to ensure that the small fields do not contribute to a significant portion of the dose at any point otherwise the uncertainty in output factor will dominate the uncertainty to the dose delivered. For example, a 16% discrepancy on a segment contributing only 10% of the dose to a point will result in an acceptable 1.6% error whereas the same discrepancy on a segment contributing 50% of the dose would be unacceptable. The dosimetry of sub-fields which require small numbers of monitor units depends critically in the start up characteristics of the linear accelerator and the linearity of the dose measuring systems down to very low doses [23, 24]. Figure 4 shows the results of measurement of the dose per monitor unit as a function of segment dose, normalized to unity for a dose of 100 mu. The data measured two Elekta SL (Elekta Oncology Systems, Crawley UK) linear accelerators. The upper curve was measured on an older machine which used a different method of beam control from that for the newer machine on which the lower curve was measured. Although the dose per monitor unit is of prime importance the beam flatness, symmetry and beam quality during start will also influence the accuracy of IMRT delivery. Following the argument set out in relation for geometric accuracy it is important to ensure that segments with low numbers of

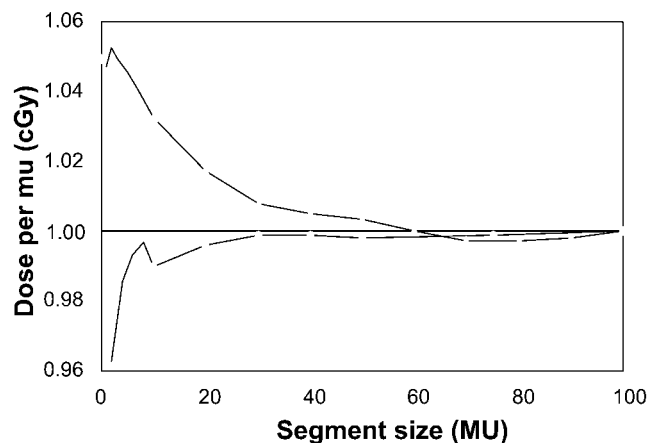


Figure 4. Variation of dose per monitor unit with segment dose (from Hansen et al, 1998 [24]).

monitor units do not contribute such a large fraction of the total dose that the variations become significant.

An increase in the number of segments in pursuit of accuracy also reduces the efficiency of delivery particularly for linear accelerators which require the radiofrequency (RF) power to be turned off between segments. This is a consequence of the finite time needed for the RF system to stabilize before a beam is launched. The use of fast tuning magnetrons on some machines [25] have been shown to improve efficiency by reducing intersegment time while maintaining acceptable beam performance. On machines with grid controlled guns the RF power can be maintained during the intersegment period and so the beam can be re-established as soon as the leaves have moved to the correct position for the next segment. This maximizes efficiency but care has to be taken to avoid any dark current (residual beam current that is accelerated through the wave guide when the electron gun is turned off) which will give rise to unwanted dose during the intersegment period [26]. This effect has not been widely reported and it has only been observed in the "high energy" modes of operation. Careful commissioning and subsequent quality control procedures on linear accelerators is indicated.

Dynamic MLC fields

As already stated, there are many similarities between segmented and dynamic MLC treatments and the differences have perhaps been exaggerated by researchers who need to demonstrate the novelty of their particular approach and manufacturer's who need to establish commercial advantages over their competitors. Dynamic delivery can be seen as an extension of segmented delivery to the limit where the segments are so small that their delivery without interruption becomes possible. Alternatively it can be seen as an extension of the dynamic wedge concept [27] first into two dimensions by making use of the MLC and second to allow both positive and negative gradients by controlling each of the opposing leaves [28].

Many of the issues discussed in the previous section are also relevant to dynamic delivery. These include the need to avoid the tongue and groove effect, the need to include the calculation of leaf trajectories (as opposed to sequences of segments) in the dose optimization procedure and the need to avoid delivering significant portions of the dose while beam apertures are so small that the achievable accuracy in collimator positioning compromises the accuracy of dose delivery (c.f. output factor variation).

The principle of the dynamic sliding window technique for a single pair of opposing leaves is shown in Figure 5, a diagram to explain how the intensities at the three points marked by dotted lines are controlled. Trajectories, defining the positions of the left and right leaf as a function of the number of monitor units delivered are plotted in panel A. Both leaves move from left to right. At the start of irradiation the first point is between the leaves and is exposed and will continue to be exposed until the left leaf passes. The total intensity will therefore be equivalent to the number of monitor units at which the left leaf passes point 1. Point 2 is not exposed at the start of irradiation because it is shielded by the right leaf. It is first exposed as the right leaf passes and the exposure is completed as the left leaf passes. So the total intensity to point 2 is

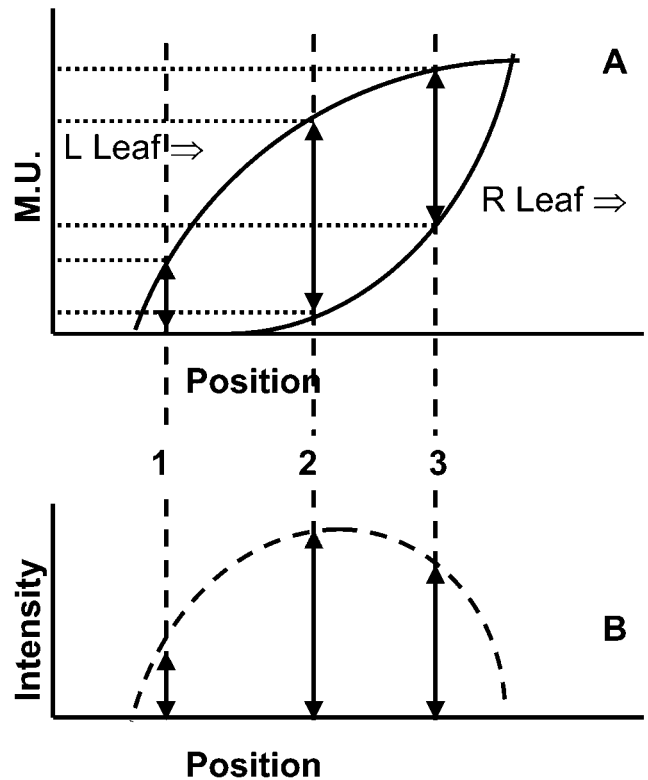


Figure 5. The principle of the dynamic sliding window modulation.

equivalent to the difference in monitor units between the right and leaf trajectories. Point 3 is similar to point 2 but the total intensity is somewhat lower.

In practice the process described above has to be inverted as, rather than determining the intensity modulation from a set of trajectories, it is necessary to determine the trajectories from a pattern of modulation which has been optimized in the treatment planning process. Algorithms to perform this task have been called "interpreters". One such algorithm, which has been elegantly described by Webb [1, 29], is shown schematically in Figure 6. The required intensity distribution A, (chosen to be a continuous version of the quantised distribution in Figure 1) is divided into regions of positive and negative gradient. The profiles in the negative regions are inverted and each is moved upwards so that the continuous trajectories shown in B are formed. These trajectories are a continuous representation of the segmented sequence in Figure 1B, demonstrating the equivalence of the two methods. However the trajectories in Figure 6B are not deliverable as the horizontal sections correspond to infinite leaf velocity. Deliverable trajectories can be created by adding a constant gradient, equivalent to a leaf velocity less than, or equal to, the maximum possible leaf velocity, to each of the trajectories. This operation maintains the vertical difference between each trajectory and therefore maintains the required modulation.

The ability to add a gradient to both trajectories whilst maintaining their vertical separation and hence dose delivered can also be used to control the relative positions of adjacent leaf pairs and eliminate the tongue and groove effect [19]. Although the adjustments to the trajectories are simple for two pairs of leaves they become complex when

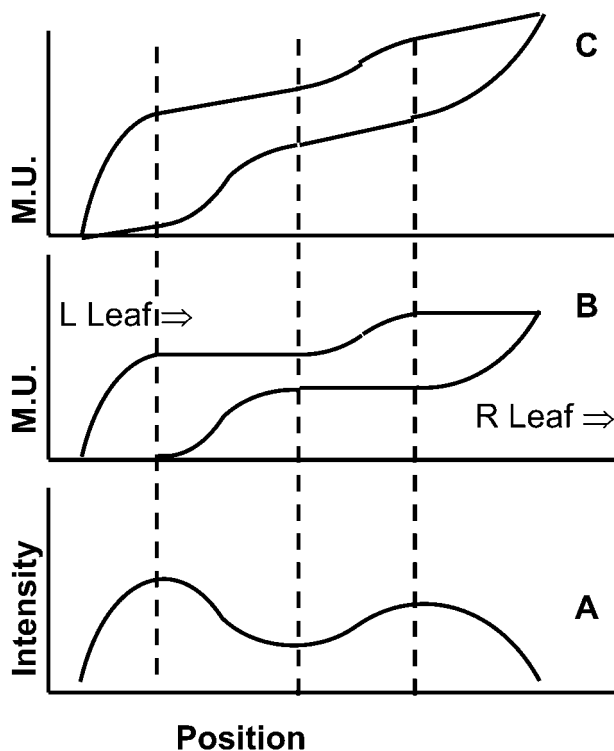


Figure 6. The generation of dynamic leaf trajectories from a modulated profile. A required profile. B trajectories without velocity constraints. C trajectories constrained to maximum leaf velocity.

an entire bank of leaves is considered. Adjustment of the trajectories of the i th pair of leaves in order to avoid the tongue and groove effect with the adjacent $(i-1)$ th pair might introduce or exacerbate the effect between the i th and $(i+1)$ th pair. So, the trajectories have to be adjusted, or synchronised, iteratively to eliminate the effect at all junctions. The cost of synchronisation is to increase the overall number of monitor units that have to be delivered and hence reduce the efficiency of delivery. In turn this increases the proportion of the dose which is delivered by leakage radiation transmitted through the collimators which ultimately puts a lower limit on the minimum dose that can be achieved within a modulated field.

As the dose delivered to any point in a dynamically modulated beam is determined by the difference in monitor units delivered between the tip of each leaf passing over that point the accuracy of positional control effects the accuracy of modulation. Considering the trajectories in Figure 5 again, if the right hand leaf had a calibration offset of, say 1 mm, the trajectory would be displaced by 1 mm to the right and a vertical separation between the two trajectories would increase so increasing the dose at all points to the right of the starting position. It should be noted that although an error in the positioning of either of the opposing leaves will cause dose errors the critical parameter is the relative error between each pair of opposing leaves as errors in the same direction cancel out the effect on dose. This interdependence between dosimetric accuracy and geometric accuracy requires careful consideration as in some circumstances it may be necessary to demand tighter tolerances on collimator position than are normally acceptable for conventional

collimation [30] or to increase the frequency and tolerances of quality control procedures. However it is necessary to distinguish between the absolute accuracy to which a collimator can be positioned for a segmented delivery and dynamic delivery. In the segmented case the time taken to achieve the correct position is, within reason, not important. However, in the dynamic case the accuracy has both a spatial and temporal dimension. Each leaf must be controlled to move along its planned trajectory so that the leaf is not only at the correct position but also arrives there at the correct time. This will be a function both of the calibration of the leaf position measuring system and the performance of the drive control system. The magnitude of the dosimetric effects of positional tolerances depends on the gradient of the leaf trajectory so the concept of a dynamic tolerance envelope [31] has been suggested, this is illustrated in Figure 7.

The modulation and trajectories in Figure 6 are shown as continuous; they are in fact quasi-continuous as the optimization program will have computed the required intensity on a grid of discrete points. In addition the continuous profiles have to be sampled as the MLC control system will only accept a finite number of control points representing the position of each leaf as a function of monitor units delivered. Hence the trajectories are formed by a series of control points and the leaves are controlled to move linearly between them. The accuracy of dynamic delivery depends on the number of control points and on the complexity of the modulation [32], typical modulations have been shown to be reproduced within 2% with 20 control points with only highly modulated beams requiring more than 50 control points. These figures relate to studies of individual beams and it may be possible to reduce the number of control points when several modulated beams are added together.

Dynamic delivery avoids some of the problems associated with segmented delivery, particularly the dosimetric problems of delivering small numbers of monitor units. However the problem of geometrically small segments remains. An interpreter which results in small beam apertures during dynamic delivery will be subject to similar uncertainties to small segments in segmented delivery so, in general, solutions which maintain large apertures for most of the irradiation are preferred. Similarly the superposition of many fields, either continuously or sequentially, places additional demands on both the measurement of beam data and on the subsequent modelling of those beams for dose calculation.

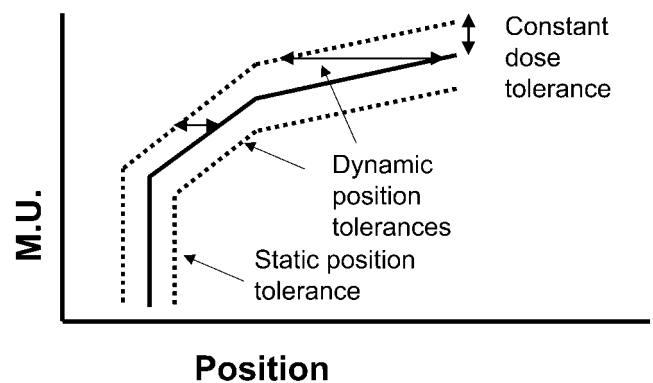


Figure 7. The dynamic tolerance concept.

In an unmodulated beam, dosimetric accuracy is the primary requirement in the central region of the beam, geometric accuracy is the primary requirement in the penumbral regions and accuracy is not critical outside the beam. In a beam created by the integration of sub-fields, dosimetric and geometric accuracy are required for all parts of each sub-field as all will contribute to the dose delivered throughout the modulated beam. This requirement has significant consequences for data collection and subsequent quality assurance.

Finally in relation to dynamic delivery, consideration has to be given to the effect of patient movement on the dose distribution [10]. The effect can be explained qualitatively by imagining a narrow sliding window crossing a structure. If the structure is stationary the dose delivered will be proportional to the time (monitor units) taken between the leading leaf exposing the structure and the closing leaf occluding it. If the structure moves in the same direction as the sliding window the exposure will be increased and conversely if it moves in the opposite direction the exposure and hence dose will be reduced. The effects of movement can be minimized by; choosing the direction of the sliding window and avoiding the predominant directions of anatomical movement, using the widest apertures possible, and using the lowest collimator speeds possible. They can be avoided by more active measures such as anatomical movement control [34] and gating of the radiation [35]. Future developments in image guided radiotherapy will undoubtedly increase the interest in this area of work and relieve restrictions to IMRT caused by uncertainty in patient position and stability.

Intensity modulated arc therapy

Intensity modulated arc therapy (IMAT) [33, 34] can be seen as a cone beam alternative to tomotherapy, avoiding the junction problems of slice based tomotherapy and utilizing a standard, IMRT capable linear accelerator. Figure 8 shows five field treatment with each field individually modulated, for the sake of this discussion it is assumed that each field is split into the same L levels. In

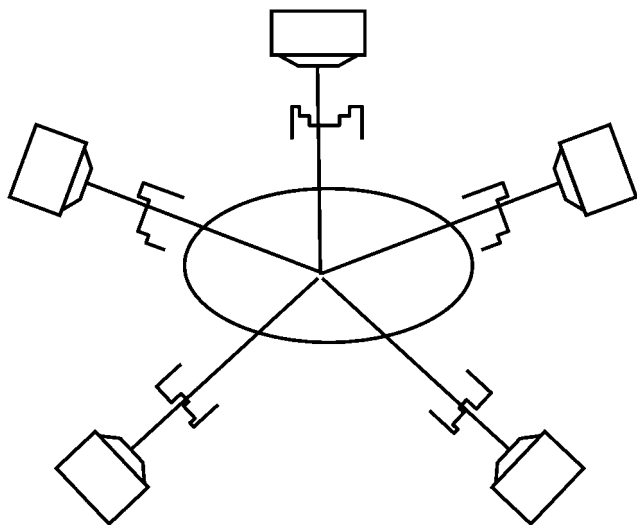


Figure 8. Intensity modulated radiotherapy fields for five gantry angles.

general there could be N fields with gantry angle G_1 – G_N . these fields could be irradiated sequentially by the segmented method with L levels, L_1 – L_L . Segmented delivery would normally involve the irradiation of all the segments from one direction to be delivered consecutively before moving the gantry to the next angle to the next field. However there is no *a priori* reason for this sequence and the same integrated dose distribution would be achieved if the lowest level segment from all gantry angles were delivered in turn followed by the second level and so on.

So the sequences for fixed gantry angles would be:

$$G_1(L_1, L_1, \dots - L_L), G_2(L_1, L_1, \dots - L_L), \dots - G_N(L_1, L_1, \dots - L_L)$$

And sequences for fixed levels would be:

$$L_1(G_1, G_2, \dots - G_N), L_2(G_1, G_2, \dots - G_N), \dots - L_L(G_1, G_2, \dots - G_N)$$

It is now easy to see that, if N is large, each level could be delivered by arc therapy during which irradiation is continuous as the gantry rotated as and as the shape of the Lth segment is continuously adjusted. The number of arcs required to deliver the entire treatment is equal to the number of levels in each distribution.

There are clearly a many common features between SMLC, dMLC and IMAT. Segmented delivery is the simplest and in terms of intuitive understanding of the process treatment planning. IMAT is perhaps the most demanding in terms of the control system requirements for the linear accelerator and MLC.

Quality assurance for the delivery IMRT

Quality assurance for IMRT includes testing at least three distinct phases of the delivery process. Extensive and complex testing is also required for treatment planning and optimization but these are outside the scope of this article. First it is necessary to ensure that the delivery system is capable of delivering modulated beams with acceptable precision, taking into account the performance of the MLC and linear accelerator. This will be established initially during commissioning of the facility and subsequent testing to ensure that the baseline performance has been maintained. Second it is necessary to ensure that the sequences or trajectories together with the monitor unit calculations, which comprise the prescription for each individual patient, will result in the correct dose and dose distribution. This assurance is necessary before a patient is treated. Finally *in vivo* measurements are often indicated to ensure that the prescribed irradiation is delivered with acceptable precision. *In vivo* dosimetry is a requirement the early stages of implementation of a technique to treat a particular class of patients. These three phases of quality control will be discussed in turn.

Linear accelerator and MLC performance

Most of the quality control requirements for the delivery of IMRT with an MLC are the same as for a standard MLC used for conformal therapy. However, consideration has to be given to the tolerances for specific tests and for the addition of tests to cover those aspects of IMRT that distinguish it from conventional delivery methods. Particular attention has to be given to precise control of

leaf positions, including their dynamic characteristics for dMLC delivery, and to the accelerator's performance in the delivery of small segments for SMLC delivery.

Segments with small numbers of monitor units depend critically on the linearity of the dose monitoring system and on the beam start up characteristics. Once the performance has been established it is desirable to include such measurements on a regular quality control programme as an additional IMRT related test.

Because the sensitivity of the final dose distribution to the precision of leaf position is highly dependent on the method of delivery, including the algorithms used for calculating leaf sequences and trajectories, it is not possible to recommend universal tolerances for leaf position accuracy. However requirements for dynamic delivery will usually be more demanding than those for conventional delivery techniques, tolerances of 0.5 mm to 1.0 mm have been suggested. Considerable effort has been put into efficient methods of measuring leaf setting, and it has been shown that the use of electronic portal imaging devices [36, 37] can provide fast and accurate measurements. While accuracy measurements of leaf position provide sufficient quality assurance for segmented delivery it is also necessary to measure the dynamic positioning performance for dynamic delivery. It has been shown [38] that exposure of a film by moving the leaves dynamically across the field, stopping for a few monitor units at predetermined positions, creates a pattern that can be analysed easily and will detect errors of much less than 1 mm.

Pre-treatment prescription verification

The treatment planning process, including optimization and segment sequence or leaf trajectory generation, results in prescriptions which consist of a large volume of data which is not intuitively related to either final 3D dose distribution or the 2D fluence distributions for each beam. Furthermore because the number of monitor units required for each beam is dependent on the particular characteristics of the sequencer or interpreter and subject to the uncertainties of small field dosimetry, there is no unique relationship between the prescribed dose at a reference point (or to a defined volume) to the number of monitor units required.

Monitor unit checking is therefore critical and the most secure method [39] is to take the prescription from each beam and using the treatment planning system, re-project the modulation on to a uniform phantom so that the dose at some reference point can be calculated. Each beam can then be delivered to a uniform phantom and the dose measured at a reference compared with the calculation. The two dimensional dose distribution for each beam can be verified in a similar way, that is by re-planning each beam on to a uniform phantom and calculating the dose distribution in a plane and comparing these calculations with measurements from a film exposed in the same plane. Evaluation of the differences between the between calculations and measurements can use the "gamma method" [40, 41] which recognises that the requirements for dosimetric accuracy are highest in regions of low dose gradient and that the requirements for geometric accuracy are highest in regions of high dose gradient. There are therefore two criteria for accuracy. The dosimetric criteria are that each point in the measured distribution should be

within a set tolerance of the expected dose. The geometric criteria are that for each point in the measured distribution there should be a point of the same dose in the expected distribution within a set tolerance distance. The method identifies areas in the two dimensional distribution which meet at least one of these criteria. For example, if the criteria are set at 3% and 3 mm than all points which are within 3% of the expected dose or are within 3 mm of the nearest point with the expected dose area considered acceptable.

Full 3D verification of the distribution is possible by the projection all the beams on to a suitable anthropomorphic phantom. These 3D distributions can be measured by sampling the doses at a point, using ionization chambers, thermoluminescent dosimeter (TLD) or other detectors, by sampling the doses in planes using photographic film or by 3D gel dosimetry [42], measuring the radiation induced changes to the properties of polymer gels. While the concept of 3D dosimetry is very attractive it is impractical for patient by patient verification but might be very useful in the validation of techniques for classes of patients.

In-treatment verification

As with any form of radiotherapy, *in vivo* dosimetry during IMRT can provide a final verification that the correct dose has been delivered and that excessive doses have not been delivered to critical structures. In addition to the normal precautions, taking account of temperature effects, careful calibration etc., great care has to be taken in positioning *in vivo* dosimeters. Regions with high dose gradients should be avoided if possible so that the results are not too sensitive to precise positioning. If electronic portal imaging is available, it can be used for verification that the fields are correctly placed and preliminary work [43] has shown that reasonable estimates of dose can be made by transit dosimetry, relating the intensity of the transmitted beam to the dose in the patient. If a portal image is required, then either the image has to be acquired over the total irradiation, accepting that the image will be modulated, and maybe rendered difficult to interpret, with the input modulation, or it can be acquired over the first few monitor units if the sequencer or interpreter can be programmed to set the first segment or initial aperture to cover the boundaries of the irradiated area.

Quality control and verification for IMRT are currently very labour intensive, with the time needed for pre-treatment verification being typically 2–6 h [39]. To increase greatly the number of patients receiving IMRT it will be necessary to reduce IMRT verification to a manageable level.

Future perspectives for delivery and quality assurance

The techniques and methods described above have developed rapidly over the last decade. The literature seems to have increased exponentially and this is a measure of the effort that has been devoted to improving dose distributions with the aim of improving patient outcome. As a result of this effort, and the considerable investment and innovation by the commercial sector, MLC based IMRT is now widely available if not

implemented. There is little doubt that these developments will continue, but maybe at a slower pace, as this technology is consolidated and becomes part of normal practice. Higher resolution MLCs will become the norm, but although these will allow better beam shaping and better modulation, the basic principle of MLC based IMRT will be unchanged. Some radically different approaches to 3D conformal therapy have been proposed and these will continue to be explored. Helical tomotherapy [44] including helical imaging is an ongoing development of single slice tomotherapy, but its widespread implementation is likely to be limited by the need for dedicated purpose built machines. More innovative solutions based on linear accelerators mounted on robotic arms [45, 46] guided by real time image guided control systems have been suggested and built but these also require very specialized equipment and if they do become more widely available it is likely it will be for niche treatments that cannot be carried out on more conventional equipment. Finally it is necessary to consider the use of protons and heavy ions which have been advocated as the ultimate tool for conformal therapy [47]. Intensity modulated proton therapy might eventually prove to be a significant improvement over intensity modulated X-ray therapy but not until proton facilities are as accessible as linear accelerators with MLCs.

While competing technologies are being developed the use of linear accelerator based IMRT will be refined. Confidence in the method will grow and the quality control and verification processes eliminated or streamlined. The development of portal imaging devices to provide online verification, integrated seamlessly into the delivery process, will remove one of the major inhibitions to more general use. Finally the integration of kilovoltage imaging systems into the linear accelerator environment [48] will provide the means to measure patient position and movement and then control the accelerator accordingly, thus exploiting the full potential of IMRT in the delivery process.

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