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Determination of beam intensity in single step for IMRT inverse planning

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Abstract

In intensity modulated radiotherapy (IMRT), targets are treated by multiple beams at different orientations each with spatially-modulated beam intensities. This approach spreads the normal tissue dose to a greater volume and produces a higher dose conformation to the target. In general, inverse planning is used for IMRT treatment planning. The inverse planning requires iterative calculation of dose distribution in order to optimize intensity profile for each beam and is very computation intensive. In this paper, we propose a single-step method utilizing a figure of merit (*FoM*) to estimate the beam intensities for IMRT treatment planning. The *FoM* of a ray is defined as the ratio between the delivered tumour dose and normal tissue dose and is a good index for the dose efficacy of the ray. To maximize the beam utility, it is natural to irradiate the tumour with intensity of each ray proportional to the value of *FoM*. The nonuniform beam intensity profiles are then fixed and the weights of the beam are determined iteratively in order to yield a uniform tumour dose. In this study, beams are employed at equispaced angles around the patient. Each beam with its field size that just covers the tumour is divided into a fixed number of beamlets. The *FoM* is calculated for each beamlet and this value is assigned to be the beam intensity. Various weighting factors are incorporated in the *FoM* computation to accommodate different clinical considerations. Two clinical datasets are used to test the feasibility of the algorithm. The resultant dose–volume histograms of this method are presented and compared to that of conformal therapy. Preliminary results indicate that this method reduces the

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critical organ doses at a small expense of uniformity in tumour dose distribution. This method estimates the beam intensity in one single step and the computation time is extremely fast and can be finished in less than one minute using a regular PC computer.

1. Introduction

The goal of radiation therapy is to maximize the dose to the tumour volume whilst keeping the dose to the surrounding normal tissues at acceptable levels. Some optimization techniques have already been developed to aid the radiation therapy treatment planning. Recent techniques include optimizing beam weights, wedge angles in coplanar plans and beam profiles for intensity modulated radiotherapy (IMRT).

The optimization algorithms can be categorized into deterministic (Bortfeld *et al* 1990, Spirou and Chui 1998, Olivera *et al* 1998, Wu and Mohan 2000) and stochastic methods (Webb 1991, Mageras and Mohan 1993). The optimization algorithms iteratively compute a set of non-negative beam intensity profiles that minimize the cost function. A quadratic cost function defined as the sum of costs for target and critical organs is used to quantitize the optimization process. For deterministic methods, the number of iterations in reaching the optimum is typically less than 100. However, they require that the cost function be convex, otherwise the process may get trapped in a local minimum. Stochastic algorithms such as simulated annealing methods have the advantage of escaping the local minimum and finding a global minimum. But these algorithms are slow and need tens of thousands of iteration to reach optimum. Each iteration involves the calculation of radiation dose from all the beams. The dose calculation is very time consuming; it takes several hours to finish just one optimization planning (Pugachev *et al* 2000, 2001). The enormous computation time is due to the large search space when the beam orientation optimization is taken into consideration.

In this paper we report a single-step method to determine the beam profiles for IMRT. Our method utilizes a figure of merit to estimate the radiation efficacy for each beamlet and this value is then used as the beam profile. Since the critical organ dose is already considered in the determination of beam intensity, only the tumour dose uniformity needs to be optimized. There is no need to recalculate the dose distribution in beam weight optimization and the planning process is very fast.

2. Method

In our method, the intensity of each beamlet is determined in one single step with its value to be proportional to its corresponding *FoM* such that the isocentre receives the prescribed dose. These nonuniform beam intensity profiles are then fixed and the weights of all beams are determined iteratively. The beams are distributed at equispaced angles around the patient. Figure 1 is the flow chart illustrating the proposed method.

2.1. Model description

The isocentre of the treatment is set near the centre of the tumour volume. The gantry rotational angle is divided into K equally spaced directions, where K is the number of beam ports to be employed. The size of the beam at each angle is just large enough to cover the tumour volume.

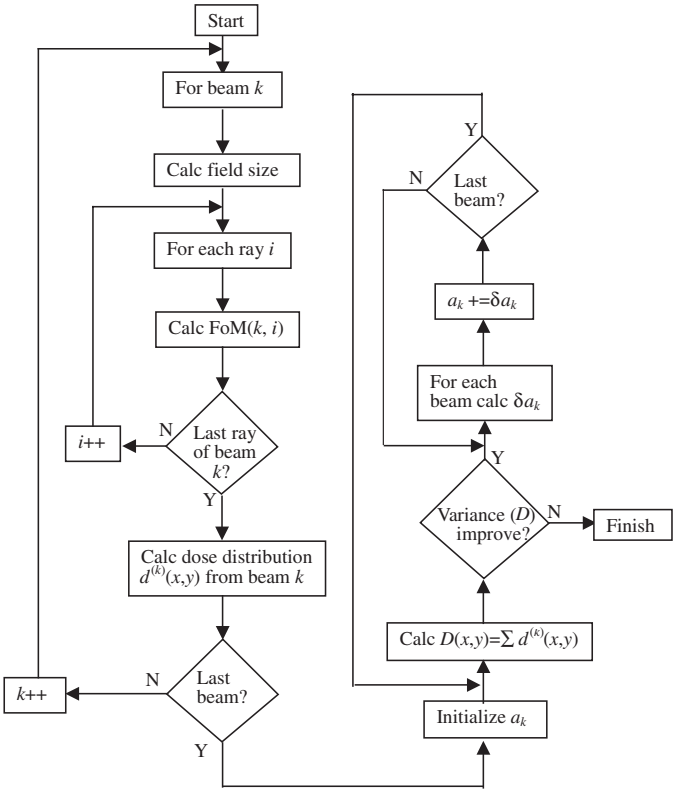

Figure 1. The flow chart illustrates the proposed method.

Depending on the resolution required, each incident beam is divided into I (=20) beamlets or rays. The problem of treatment planning is to determine the $K \times I$ values of beamlet intensity.

2.2. Figure of merit

The figure of merit of a ray is defined as (Johns and Cunningham 1983)

$$FoM = \frac{\text{Total energy imparted to the target volume by the ray}}{\text{Total energy imparted to patient by the ray}}. \tag{1}$$

The *FoM* can be used as an index for the efficiency of dose utilization in radiation therapy. In this study, the *FoM* is used to estimate the beam intensity for IMRT. The *FoM* for the *i*th ray of *k*th beam (figure 2) is defined as

$$FoM(k,i) = \frac{d_{\text{off}} + \sum_{(x,y)\in(\text{PTV}\cap\text{beamlet}(k,i))} [\Delta t(x,y) \times d^{(k)}(x,y)] \times s(x,y)}{\sum_{\substack{(x,y)\in\text{beamlet}(k,i) \\ (x,y)\notin\text{PTV}}} \Delta t(x,y) \times d^{(k)}(x,y) \times w_{\text{T}}(x,y) \times P^{(k)}(x,y)} \tag{2}$$

where d_{off} is the dose offset, Δt is the intersecting length of beamlet (k, i) with pixel (x, y), $d^{(k)}$ denotes the integral radiation dose from the kth beam to the pixel and w_{T}, $P^{(k)}$ and s are the tissue weighting factor, penalty function and star pattern correction matrix, respectively. These factors are included for various clinical considerations. The dose calculation algorithm and the meaning of each factor will be described in the following sections.

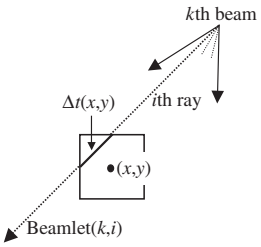

Figure 2. The geometric relation of a beam with a pixel for dose calculation. Each beam with its size just covers the tumour is divided equally into I beamlets. Δt is the length of interception of beamlet (k, i) with pixel (x, y).

2.2.1. Dose calculation. In this study, the doses on fixed locations for each beamlet are pre-calculated and the dose at any pixel is interpolated from them. The dose is expressed as

$$d(x, y) = \frac{SAD^2 \times T(x, y)}{t^2} \tag{3}$$

where *SAD* is the source to axis distance, t is the distance between the pixel and the source and T represents the probability that the ray does not undergo any interaction before reaching the pixel. Only primary radiation and no scattering is taken into consideration. This simplified dose calculation model is used for the purpose of demonstrating the idea and understanding the fundamental aspect of the system. Although the scattered dose is closely related to the beams, its effect on the optimization is small.

2.2.2. Dose offset. The dose offset is used to control the range of the *FoM* value. Higher offset value means smaller dynamic range of *FoM*, and the beam's intensity is closer to uniform profile as of conformal therapy. As a result, a more uniform dose distribution to the tumour will be generated accompanied with a higher dose to the sensitive organ.

2.2.3. Tissue weighting factor. The tissue weighting factor represents the proportionate detriment to a tissue when the whole body is irradiated. Typical values of weighting factors can be found in the literature (Turner 1995). In general, the weighting factor of an organ is proportional to its sensitivity to radiation damage. These tissue weighting factors can be varied by the user for special clinical cases. In this study, unless specified, all the tissues will be treated as muscle.

2.2.4. Penalty function. To limit the normal tissue dose below a tolerance level, a penalty function is also included in the calculation of *FoM*. The penalty function $P(x, y)$ is defined as

$$P^{(k)}(x, y) = \begin{cases} 1 & \text{if} \quad d^{(k)}(x, y) < d_{\mathrm{c}}(x, y) \\ \left[\frac{d^{(k)}(x,y)}{d_{\mathrm{c}}(x,y)}\right]^2 & \text{otherwise} \end{cases} \tag{4}$$

where $d_c(x, y)$ $(=Th(x, y) \times d^{(k)}(\text{isocentre}))$ is the tolerance dose, $d^{(k)}(\text{isocentre})$ is the dose at isocentre delivered by the kth beam and Th, the tolerance factor, is dependent upon the prescribed dose and organ type under consideration. The tolerance factor is small for sensitive organ and higher for noncritical tissues (Pugachev *et al* 2000). In this study, the tolerance factor is set to be equal to $w_{\mathrm{muscle}}/w_{\mathrm{T}}(x, y)$. This definition of penalty function is conformed to the general practice that is set to be proportional to the square of the excess dose beyond the tolerance dose.

2.2.5. Star pattern correction matrix. In this algorithm, every pixel in the target volume receives radiation with intensity proportional to *FoM* from all angles. The *FoM* is a function of the intersecting length between the beamlet and the target. In general, this method is similar to backprojection algorithm (Bortfeld *et al* 1990, Holmes *et al* 1995). Such an irradiation will lead to a dose distribution that is too low near the edges of the target volume. This is analogous to the star pattern produced by simple backprojection in the CT reconstruction. To remedy this, we employ a correction matrix to correct the non-uniformity in the tumour dose distribution (Chuang and Tzeng 2000). The star pattern correction matrix is calculated before treatment planning based on the shape of tumour volume and the port orientations. We first estimate the tumour dose ($d_u(x, y)$) with beam intensity equal to the intersecting path length of each beamlet with the target. Using this dose distribution to calculate the star pattern correction matrix $s(x, y)$ ($= 1/d_u(x, y)$) and assign it as the weight to each pixel of the target, a more uniform result can be expected. The whole ideas can be expressed by the following equations:

$$P_\theta(t) = \iint_{(x,y)\in \mathrm{PTV}} w_{\mathrm{pr}}(x, y) \times \delta(x\cos\theta + y\sin\theta - t)\,\mathrm{d}x\,\mathrm{d}y \tag{5}$$

$$d_u(x, y) = \int P_\theta(x\cos\theta + y\sin\theta)\,\mathrm{d}\theta \tag{6}$$

where w_{pr} is a prescribed weighting factor and is equal to 1 for a uniform tumour dose distribution, $\delta(x)$ is the delta function and θ is the port angle, and $d_u(x, y)$ is the resultant image of simple backprojection and is afflicted with star pattern. Letting w_{pr} be equal to the inverse of $d_u(x, y)$ and performing simple backprojection again will yield a more uniform tumour dose.

2.3. Beam weight optimization

After the intensity of each beamlet is determined (=*FoM*), the final stage of the planning is to determine the weight for each beam such that the resultant tumour dose is uniform. Two techniques, range minimization and variance minimization, are employed and performed in tandem.

2.3.1. Range minimization. The resultant dose distribution $D(x, y)$ from all beams is

$$D(x, y) = \sum_k a_k d^{(k)}(x, y) \tag{7}$$

where a_k is the weighting factor of kth beam with $\sum a_k = 1$. To avoid the hot spot and cold spot in the tumour volume, the goal of optimization is to minimize the dose range in tumour. This can be achieved by iteratively updating a_k by the following increment:

$$\delta a_k = R_c[d^{(k)}(\mathrm{minptr})/d^{(k)}(\mathrm{maxptr})] \tag{8}$$

where R_c (=0.35) is a constant controlling the speed of convergence, 'minptr' and 'maxptr' are the locations of the minimum and maximum dose in $D(x, y)$, respectively. The inverse of the quotient in equation (8) (i.e. $d^{(k)}(\mathrm{maxptr})/d^{(k)}(\mathrm{minptr})$) represents the contribution from the kth beam to the tumour dose nonuniformity. A major inhomogeneity contributor will get a small value in δa_k (>0) and that will cause a decrease in its weighting factor (a_k) after normalization. The iteration is continued until there is no more increase in the value of $D(\mathrm{minptr})/D(\mathrm{maxptr})$.

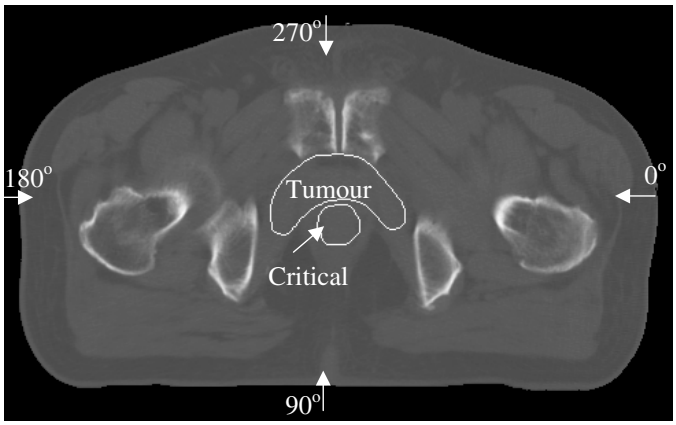

Figure 3. The CT image of prostate cancer with the C-shape tumour and critical organ outlined.

2.3.2. Variance minimization. This method tries to minimize the variance of the tumour dose distribution. The beam weight optimization can be achieved by updating each weighting factor with the amount of δa_k calculated as

$$\delta a_k = R_c \frac{\sum_{(x,y)\in \mathrm{PTV}} [D(x,y) - \overline{D}] \times d^{(k)}(x,y)}{\sum_{(x,y)\in \mathrm{PTV}} d^{(k)}(x,y)^2} \tag{9}$$

where R_c (=1) is a convergent rate constant and $\overline{D}$ is the average of $D(x, y)$. The optimization starts with equal weights for all beams and stops if the variance of the tumour dose distribution shows no more improvement.

3. Experiments and results

A prostate cancer and a nasopharynx cancer (NPC) are used for the study. The isocentre is located at the tumour centre. The photon energy used is 2 MeV which is equivalent to 6 MV spectrum. The source to axis distance (*SAD*) is 100 cm. The beams are equally distributed in clockwise direction around the isocentre.

3.1. Prostate cancer

Figure 3 shows the CT image of the prostate cancer patient. The tumour is of C-shape and beneath it is the sensitive organ (rectum). The tissue weighting factor of the sensitive organ is set to 3 (relative to w_{muscle}).

3.2. FoM of a beamlet

Figure 4 shows the *FoM* calculated for each beamlet with port angles of 0°, 90°, 180° and 270°. Note that in the 90° and 270° curves, the values of *FoM* in the central beamlets are much lower than the rest. It is due to the presence of critical organ. Only a small portion of the critical organ is intersected by the 180° and 0° beams and it causes the decrease (increase) in the *FoM* values on the right-hand side for the 180° (0°) beam. The presence of the critical organ also causes the much larger dynamic ranges in the *FoM* values for the 90° and 270° beams. The *FoM* values are smaller for the 0° and 180° beams than for the 90° and 270°

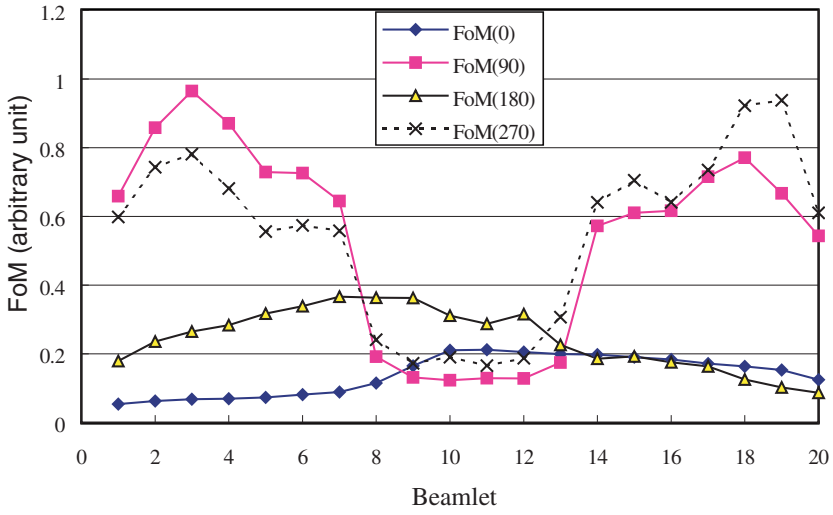

Figure 4. The profiles of *FoM* as calculated at angles 0°, 90°, 180° and 270° for the prostate case ($K = 4$, $d_{\text{off}} = 100$, $w = 3$ for the rectum and $w = 1$ for the rests).

beams. This is because the denominator in the calculation of *FoM* is proportional to the length of the ray intersecting the body, which is larger for the 0° and 180° beams.

3.3. Dose efficacy and dose uniformity

The dose ratio (*DR*) for *k*th beam is defined as

$$DR(k) = \frac{\sum_{(x,y)\in \text{PTV}} d^{(k)}(x, y)/N_{\text{t}}}{\sum_{(x,y)\in \text{OAR}} d^{(k)}(x, y)/N_{\text{c}}} \tag{10}$$

where N_{c} and N_{t} are the pixel numbers of critical organ and tumour, respectively. The *DR* represents the dose efficiency of each beam. Figure 5 illustrates the variation of *DR* with port angle for each beam for the prostate case with beam number equal to 7. The *DR* of conformal technique is less dependent upon port angle and roughly equal to 1. The *DR* of *FoM* technique is varied between 2 and 3 which means the efficiency of dose usage for *FoM* technique is about two to three times better than conformal technique. Both techniques show a maximum efficiency at 0°. Due to the shape and orientation of the tumour, the thickness of the tumour intercepted by the beam from the 0° port angle is larger than that from other angles. Furthermore, part of the critical organ is located outside the irradiation field of the 0° beam and spared. As a result, the beam from the 0° port angle shows the best efficacy.

Figure 5 also shows the coefficient of variation (*CV*) for both techniques. The *CV* is defined as the ratio of standard deviation and mean of a distribution. The *CV* of a tumour dose distribution can be served as an index of dose uniformity. The modulation of beam intensity causes the dose difference in tumour and thus the curve of *CV* follows the shape of *DR*. The *FoM* technique has large dynamic range in beam intensity profile and thus large variation in dose distribution.

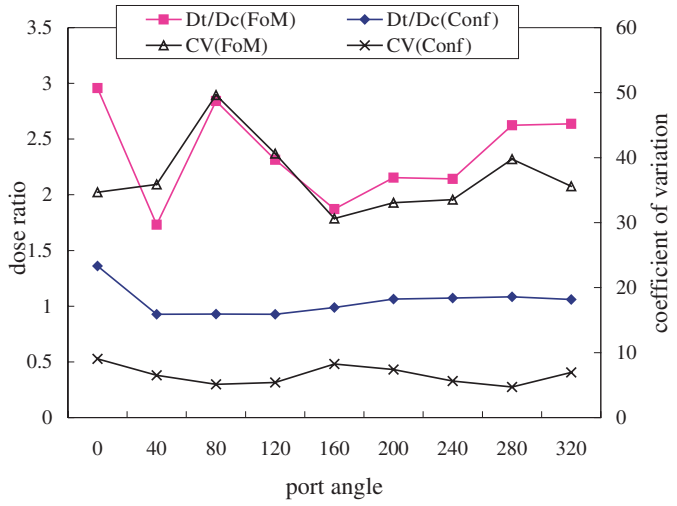

Figure 5. The dose ratio and coefficient of variation (CV) using conformal (Conf) and *FoM* techniques for the prostate case ($K = 9, d_{\text{off}} = 100, w = 3$ for the rectum and $w = 1$ for the rests).

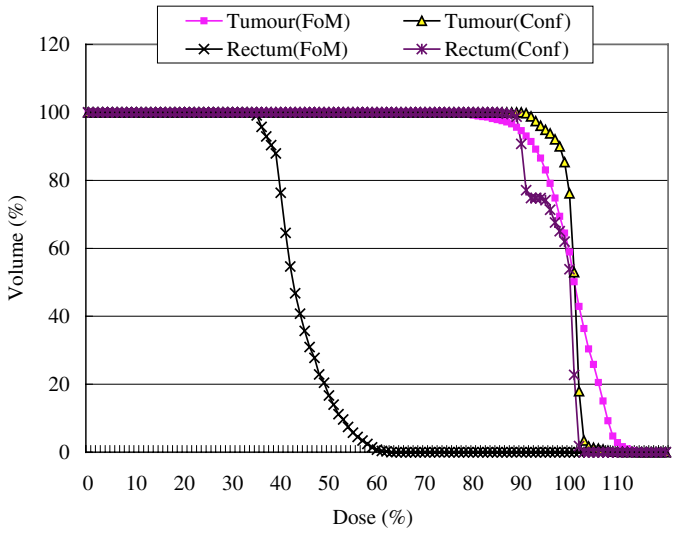

Figure 6. DVH of tumour (D_{t}) and critical organ (D_{c}) for conformal (Conf) and *FoM* techniques ($K = 5, d_{\text{off}} = 100, w = 3$ for the rectum and $w = 1$ for the rests).

3.4. Dose–volume histogram

Figure 6 plots the dose–volume histogram (DVH) for tumour and critical organ for the prostate treatment with beam number equal to 5. Over 90% of the target region receives more than 90% of dose units for both techniques. The conformal technique has slightly better DVH for tumour. Because the separation between the tumour and the critical organ is quite close, treatment planning using conformal technique will yield similar dose for both organs. It is shown that about 60% volume of the critical organ has 100% dose units. The *FoM* technique has much smaller dose to the critical organ. Due to the dose efficiency of the *FoM* technique,

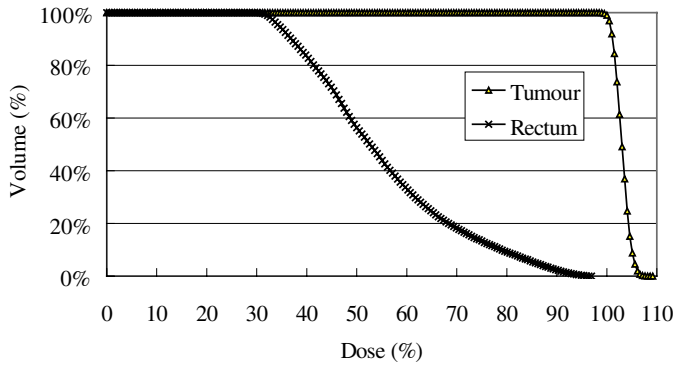

Figure 7. DVH of a Varian CADPLAN inverse planning system using five beams of equisapced angles.

more than 60% volume of the critical organ has dose units smaller than 40%. This result agrees with the difference in *DR* between the two techniques.

Figure 7 shows the resultant DVH of a Varian CADPLAN inverse planning on the prostate cancer using five beams of equispaced angles. The CV of the tumour dose distribution is 1.44% and the average rectum dose is 55.0%. It takes about 20 min to complete the treatment planning with 300 iterations. Compared to the commercial algorithm, our method yields less rectum's dose but greater inhomogeneity in the tumour dose distribution.

3.5. Beam intensity

Figure 8 plots the beam intensity from each angle using *FoM* technique for prostate case with beam number equal to 9. The field size of irradiation (beam's eye view) is varied with the port angle with maximum at 80° and 280°. The modulations of the beam intensity as influenced by the critical organ are obvious. Due to the large value of dose offset used in *FoM* calculation, this method generates very smooth modulation as needed in the beam delivery system.

3.6. Beam weight

Figure 9 plots the beam weight percentage as a function of port angle for prostate case with beam number equal to 9. In general, if there is no critical organ the weight percentage of each beam should be roughly the same (since the tumour is at the centre of the body). Thus the variation of beam weight percentage becomes higher as w_T of the critical organ increases from 3 to 5. Due to its high dose efficiency, the 0° beam has maximum weight for both tissue weighting factors. When w_T is increased from 3 to 5, the beam weighting of each angle adjusts slightly. More weights are shifted from anterior direction (200°–320°) to posterior direction (0°–160°). The change depends upon the location of the critical organ. The critical organ is located behind the tumour and closer to the beam source when the radiation is from posterior direction. Stein *et al* (1997) observed similar phenomenon. They explained: 'beams coming from the direction of rectum are preferable since they allow greater control over dose distribution in the regions close to the sensitive structure'.

3.7. Effect of beam number

The average critical organ doses versus *CV* are plotted in figure 10 for prostate treatment planning with beam numbers equal to 5, 9, 13 and 17. Each data point on the curve represents

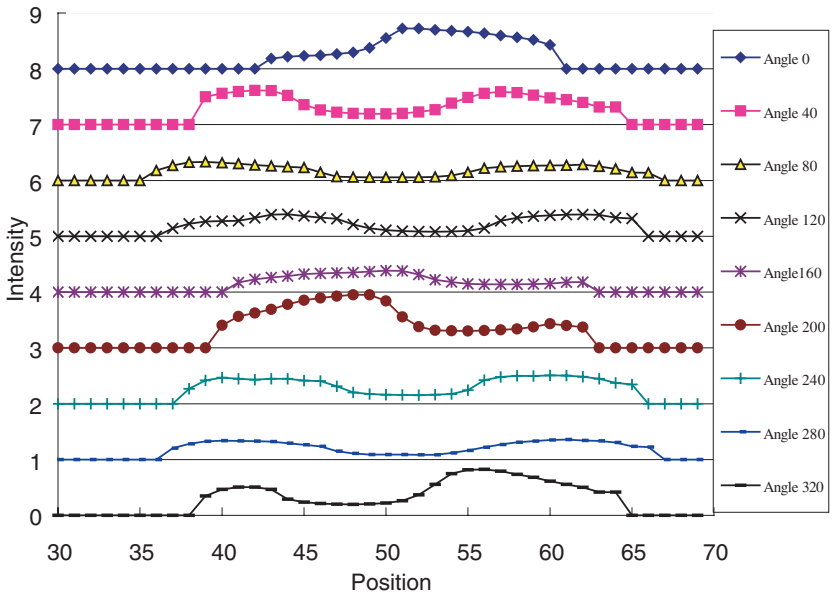

Figure 8. Beam intensity calculated using *FoM* technique for the prostate case ($K = 9$, $d_{\text{off}} = 100$, $w = 3$ for the rectum and $w = 1$ for the rests).

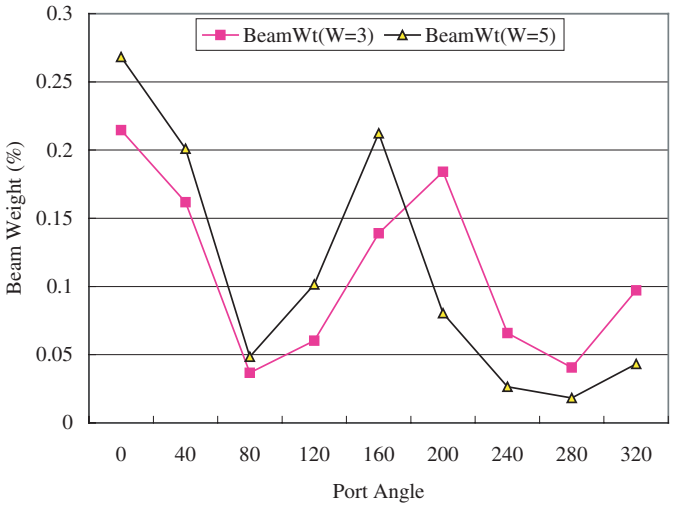

Figure 9. Percentage of beam weight from each angle for the prostate case with tissue weighting factors (w) equal to 3 and 5 ($K = 9$, $d_{\text{off}} = 100$ and $w = 1$ for the rests).

various percentages of beam weight for the specified beam number. For a given beam number, the uniformity of tumour dose improves at the expense of slightly increasing dose in the critical organ. Note that the dose to the sensitive organ increases with increasing beam number (when the uniformity is held constant). This is contradicted to previous report. Stein *et al* (1997) studied the number and orientations of beams in IMRT and showed that the rectum dose decreases as the number of beams increases. This contradiction is due to the fact that

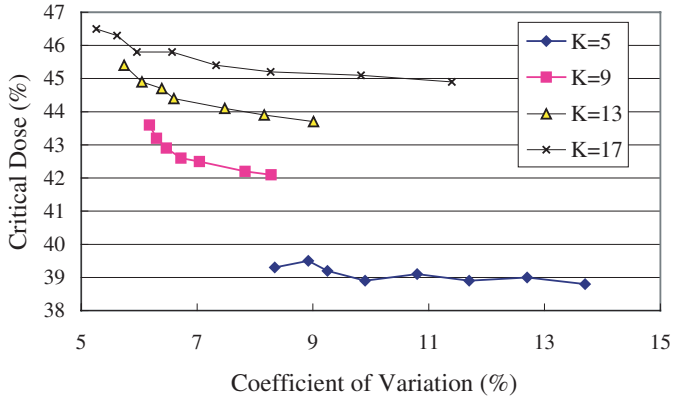

Figure 10. Average critical organ dose versus coefficient of variation for various beam numbers (K) for the prostate case ($d_{\text{off}} = 100$, $w = 3$ for the rectum and $w = 1$ for the rests).

their objective function is a function of both target dose uniformity and critical organ dose tolerance. In our case, we assume that the dose tolerance for the critical organ has been taken care of by the use of *FoM* and our optimization is aiming only at improving the tumour uniformity. The critical organ is located very close to the tumour and there is no way to avoid the critical organ completely. Consequently, the dose to the critical organ does not spread out as beam number increases. Furthermore, during beam weight optimization, the beams with poor dose distribution (usually are more dose efficient, e.g., the 80° beam) will be reduced more than those with uniform dose distribution and cause an increase in the critical organ dose.

3.8. NPC case

The CT image of the NPC patient is shown in figure 11. Seven beams of equispaced angles are employed for this case. The tissue weighting factors used for this case are 5.0 for the brain stem and 3.0 for the parotid glands. The resultant DVHs of the tumour and brain stem for both treatment techniques are shown in figure 12. As before, the conformal therapy has a better dose distribution for the tumour. The average doses for brain stem are 16.4% and 47.1% for IMRT and conformal therapy, respectively. The *FoM* technique shows a significant dose reduction for critical organ.

4. Discussion

The purpose of treatment planning is to deliver a prescribed dose to the target without delivering too much dose to sensitive organs. Most inverse planning methods iteratively create non-uniform beam intensity profiles using an object function to drive the optimization process. In this work, *FoM* is actually the intensity profile and computed in a single step. A ray with higher value of *FoM* is considered to be more dose efficient because it delivers more doses to tumour and less doses to normal tissue. It is natural to set the beam intensity to be proportional to *FoM*. Since the *FoM* calculation already takes into consideration the normal tissue dose, the optimization needs only to deal with the uniformity of the tumour dose. Furthermore, the dose distribution from each beam is needed to calculate one time only. The iterative optimization is simply to determine the weights of each non-uniform beam.

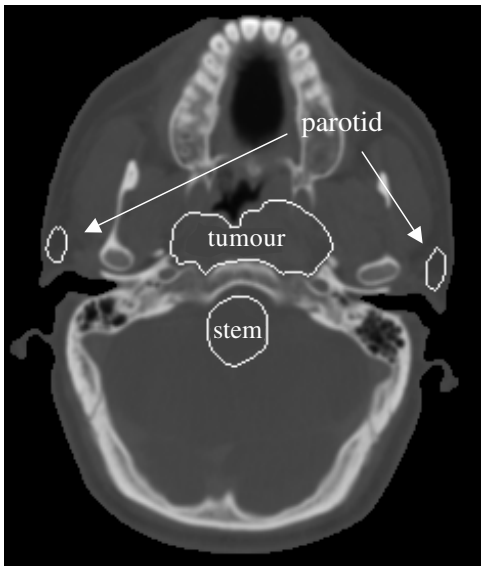

Figure 11. The CT image of a NPC case with tumour, brain stem and both parotid glands outlined.

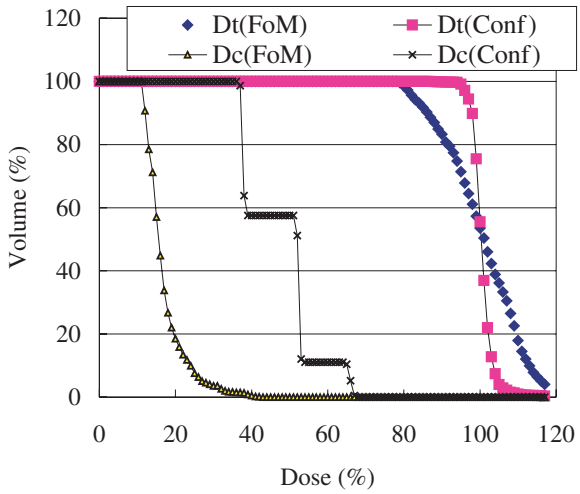

Figure 12. The DVH of the tumour (D_t) and brain stem (D_c) for the NPC case ($K = 7, d_{off} = 100, w = 5$ for the brain stem, $w = 3$ for the parotid glands and $w = 1$ for the rests, where 'Conf' is a conformal technique).

In this method, the rules that determine the modifications made to the weight are deterministic and the optimization can be finished in a few iterations. While the stochastic algorithms are very slow as they spend a lot of time evaluating the bad treatment plans. Furthermore, the search spaces of this method (number of beam weight $= K$) are much smaller than conventional inverse planning (total number of beamlet intensity $= K \times I$). As a result, our method is much faster than the stochastic techniques and can be finished in a very short time (<1 min in a PC).

The optimum treatment planning is a trade-off between the dose uniformity and dose efficacy. The essence of using *FoM* to estimate the beam intensity is to maximize the radiation dose effectiveness. Our method then fixes the beamlet weights and optimizes the relative weights of each beam. It is not optimal in terms of tumour uniformity. However, this treatment planning is optimal from the point of dose efficiency, i.e. the intensity of each beamlet is proportional to the figure of merit of that ray and the normal tissues will receive less doses.

The use of equispaced angles in this study is justifiable. The IMRT usually employs a large number of incident beams and the beam direction is less important. In fact, for relatively large number of ports, it can be shown that an evenly spaced number of ports around the patient seems to be nearly optimum (Bortfeld and Schlegel 1993).

It is difficult to generate a rapidly varying intensity profile due to scattered radiation (Spirou and Chui 1998). Furthermore, this kind of profiles may increase beam-on time. A smoothing filter is usually applied to modify the beam profiles and this will alter the optimization. In this study, the smoothness of the intensity profiles can be controlled by dose offset in the calculation of *FoM*. The higher the value of the dose offset, the smoother the beam profile. However, higher offset means less dose efficiency of the beam and thus causes more doses to the critical organ.

Ling *et al* (2000) proposed the use of biological image to outline the multiple target sizes such that different dose descriptions can be delivered. These different dose levels can be interpreted as weighting factors (w_{pr}) assigned to tumour volume and to be incorporated as the correction matrix in the calculation of *FoM*.

Currently the calculation of *FoM* is based solely on physical dose. The planning can be expanded to consider biological outcome. The tumour control probability (TCP) and normal tissue complication probability (NTCP) are good candidates for that purpose. We can define a biological *FoM* as the ratio between TCP and NTCP. The other possibility is to use probability of complication-free tumour control (P_+) defined as (Brahme 1999, Webb 2001)

$$P_+ = \mathrm{TCP} - \mathrm{NTCP} + \delta(1 - \mathrm{TCP})\mathrm{NTCP} \tag{11}$$

where parameter δ controls the statistical independence of benefit (TCP) and injury (NTCP).

Our future work will be to incorporate constraints such as hard dose constraints and DVH constraints into consideration. The basic idea is that if each individual beam fulfils the constraints, then their linear combinations (multiple beams) will also satisfy the constraints. For those beams that violate the constraints, their intensity profiles can be modified by gradually increasing the tissue weighting factor (w_{T}) of the sensitive organ in the *FoM* calculation until these constraints are satisfied.

5. Summary

In this study, we described a single-step method for the determination of beam intensity for IMRT. The definition of *FoM* warrants the effective usage of radiation dose in the treatment. Various weighting factors can be incorporated to reflect different clinical conditions. Preliminary results indicate that this algorithm can generate uniform tumour dose with critical organ dose kept below a tolerance limit. As compared to other inverse planning techniques, the method is fast and straightforward.

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Endnotes

(1) Author: Please provide the place of publication in Johns and Cunningham (1983).