Radiotherapy delivery during motion

Sofie Ceberg and Sven Å J Bäck

Department of Medical Radiation Physics, Lund University, Skåne University Hospital, Malmö, Sweden

Sofie.ceberg@med.lu.se

Abstract. This paper discusses the 3D dosimetric consequences of radiotherapy delivery during two kinds of motion, (i) the respiratory motion by the patient and (ii) the motion by the gantry while rotating around the patient.

Respiratory motion primarily compromises treatments in the thorax and abdomen regions. Several strategies to reduce respiratory motion effects have been developed or are under development. The organ motion could for instance be measured and incorporated in the treatment planning, or adapted to by using respiratory gating and tumour-tracking delivery techniques.

Gantry motion is involved in various forms of intensity-modulated arc-therapy techniques. The purpose is to increase the modulation by simultaneously varying the MLC positions, the rotation speed of the gantry, and the dose rate during the treatment.

The advantage of these techniques is the increased possibility to deliver a high absorbed dose to the target volume while minimizing the dose to normal tissues. However, the dosimetric uncertainties associated with motion, small fields and steep dose gradients, has to be evaluated in detail, and this requires adequate true 3D dose-verification tools.

1. Incorporating organ motion in treatment planning

The most widely used approach to account for organ motion is to collect a large data set of measured respiratory motion, calculate the statistical distribution, and integrate this into the treatment planning procedure as an additional margin to the PTV. The measured amplitude, applied to the internal margin (IM), can vary as a function of various factors related to the tumour (e.g. tumour site and the attachment to the chest wall), patient (e.g. respiratory capacity, oxygenation and anxiety) or treatment protocol (e.g the immobilisation and patient positioning) [1]. However, this approach, based on an addition to the geometrical margin, does not describe the uncertainties in the dose to normal tissue.

Another method is to directly introduce respiratory uncertainties into the dose calculation, by convolving the static dose distribution with a motion function [2]. This approach assumes that the target moves without deformation during treatment and that the medium is homogeneous.

2. Breathing adapted radiotherapy

2.1. Breath-hold and respiratory gating

The most obvious approach to reduce uncertainty related to movements induced by breathing is to ask the patient not to breathe during the actual beam-on time. This periods can generally be made to have a duration of 15-30 s. This breath-hold technique can either be voluntary [3] or active [4]. Another approach is to turn the beam on only when the target is in a favorable position during the patient’s
respiration cycle, i.e. the so called respiratory gating technique [5]. The advantages of these techniques are a) a reproducible immobilization, with the target volume and organs at risk (OAR) in a known position throughout the treatment, b) the irradiated lung volume will be reduced, and c) the cardiac toxicity will be reduced [6].

During deep inspiration, as with breath-hold and respiratory gating methods, the irradiated lung density will be reduced due to the 40-50% increased lung volume compared to the resting value [1]. This reduction of lung density can affect the dose to peripheral parts of the target volume due to loss of electronic equilibrium. Monte Carlo simulations have verified this impact on TPS calculations and demonstrated the risk of target under-dosage [7]. No 3D dosimetric measurements have verified this under-dosage, however, a low-density (lung tissue equivalent) polymer gel have been presented [8, 9], which may potentially enable the required measurements.

Further, depending on the size of the gating window during breath-hold or respiratory gating a certain dose smearing of the field edges due to movement during the beam-on time will remain [10]. This dose smearing effect is not accounted for in conventional treatment planning systems and dosimetric verifications is required. Experimental data have so far mostly been obtained using ionization chambers and films [10, 11]. Recently published data obtained using 3D detector systems such as PRESAGE™/optical-CT dosimetry [12] and polymer gel dosimetry [13] show very good agreement between static and gated measurements.

2.2. Tumour tracking

Another method to compensate for respiratory motion during treatment is tumour tracking. A novel promising strategy uses the dynamic multileaf collimator (DMLC) to continuously align and reshape the treatment machine aperture to follow the target motion in real time [14]. The advantage of this technique is the ability to allow for a tighter margin around the target by continuously following, and adapting the dose delivery to its motion (Figure 1, *ibid* IC3DDose proceedings, Ceberg et al. 2010).

![Figure 1](image_url). Dose distributions as measured using gel al dosimetry with and without tumour tracking delivery.

Compared to the breath-hold and respiratory gating methods, the tumour tracking technique potentially offers additional benefits such as higher delivery efficiency and less residual target motion. However, real-time beam adaptation is not feasible without precise real-time localisation of the tumour position in 3D, which includes parallel and perpendicular motion to the MLC leaf travel direction, in-and out-of-plane rotation as well as translation along the beam direction.

4D IMRT treatment planning methods that accounts for 3D tumour motion can be adopted [15]. The proposed method uses 4D CT and was integrated with the DMLC tumour-tracking delivery and opens up for a clinical implementation. Still, very few dosimetric measurements of the DMLC tumour-tracking system have been carried out. Dosimetric uncertainties associated with a DMLC tracking system arise from the estimation of the actual target position, possible delay between target
motion detection, and beam repositioning and output variations due to e.g. dissimilar head-scatter conditions during off-axis irradiation.

A dosimetric study to investigate the ability of a DMLC-tracking system to account for target motion during radiotherapy using 3D polymer dosimetry and a bi-planar detector array (Delta4®, Scandidos) was recently undertaken (this conference). The DMLC real-time motion-tracking system was used together with the optical part of the ExacTrac system (BrainLab, Germany). The respiratory-like target motion was simulated mechanically. Good agreement in absorbed dose was found between the static and tracked measurements (ibid IC3DDose proceedings, Ceberg et al. 2010).

3. Volumetric modulated-arc therapy

Another category of motion during radiotherapy is ascribed to the gantry motion, as it rotates around the patient during a volumetric modulated-arc therapy (VMAT) delivery. The delivery could be accomplished using designated rotational therapy as Tomotherapy™ or standard linear accelerators. An early paper describing the dosimetric challenges with such techniques and the potential to use gel dosimetry was presented in 2004 [16]. RapidArc™ is a novel radiation therapy technique where the treatment is delivered during one or a few rotations of the linear accelerator gantry. The dose distribution is modulated by simultaneously varying the MLC positions, dose rate and gantry rotation speed [17]. Since the RapidArc™ optimization algorithm uses a stochastic element during the generation of the apertures it is common to see plans with apertures including single isolated leaves and disconnected small segments. Furthermore, RapidArc™ plans are delivered dynamically with leaf and gantry motion up to approximately 1 cm and 2 degrees per second, respectively. This kind of treatment delivery represents a new level of complexity, and a thorough dosimetric verification is therefore highly desirable.

In recently published work, semi-3D dosimetric verifications of RapidArc™ treatment delivery using back-projection portal dosimetry [18] and Delta4© bi-planar diode array phantom [19] have been presented. The reported results have shown very good agreement between TPS and dose measurements.

Recently a 3D verification of a RapidArc™ treatment plan and delivery using polymer gel dosimetry and Monte Carlo simulation was undertaken. Inter-comparisons between all datasets – TPS calculations, gel measurement and MC simulation - showed very good agreement [20].

4. Breathing adapted modulated-arc therapy

The combination of volumetric modulated-arc with breathing adapted methods will pose an even greater challenge to 3D dose verification.

A pre-clinical investigation, in a non-clinically released framework, evaluated the machine capability to deliver gated RapidArc™ [21]. A conventional ion chamber and a 2D ion chamber array (Seven29, PTW) were used for dosimetric verifications. Both detectors were used in combination with the Octavius phantom developed by PTW for rotational therapy verification. The results showed good agreement with TPS even for very short beam-on durations (5 s), which concluded that RapidArc™ delivery is dosimetrically accurate also when applied in combination with gating procedures.

The absorbed dose to moving targets undergoing a RapidArc™ delivery with a pre-clinical 3D DMLC tracking application has been evaluated [22]. A 2D ion chamber array (Seven29, PTW) and the bi-planar diode array Delta4© were used for dosimetric verifications. The results showed that DMLC-tracking together with RapidArc™ make a feasible combination. An additional study, using the same detector system, verified that the dosimetric accuracy was independent of the magnitude of the peak-to-peak displacement (5-25 mm) of the target and not significantly affected by the angle between the leaf trajectory and the target movements [23].

Adding an extra safety layer in the dosimetric verification of DMLC-tracked RapidArc™ delivery a true 3D dose verification using polymer gels was carried out (ibid IC3DDose proceedings, Ceberg et al. 2010). A high gamma (2%/1mm) pass rate was obtained comparing tracked and static measurements, demonstrating that the motion was well compensated for by the tracking performance.
5. Conclusions

True 3D detector systems with high resolution, where the response is independent of the direction of the incident radiation and can integrate the contribution from several beams, are highly desirable. Recently published result using PRESAGE©/optical-CT dosimetry and polymer gel dosimetry show a very promising future involving true 3D dosimetric verification of advanced dose deliveries during motion.

References

[1] P. Giraud, E. Yorke, S. Jiang, L. Simon, K. Rosenzweig, and G. Mageras, Cancer Radiother 10 (2006) 269.
[2] S. D. McCarter and W. A. Beckham, Phys Med Biol 45 (2000) 923.
[3] A. N. Pedersen, S. Korreman, H. Nystrom, and L. Specht, Radiother Oncol 72 (2004) 53.
[4] J. W. Wong, M. B. Sharpe, D. A. Jaffray, V. R. Kini, J. M. Robertson, J. S. Stromberg, and A. A. Martinez, Int J Radiat Oncol Biol Phys 44 (1999) 911.
[5] K. Ohara, T. Okumura, M. Akisada, T. Inada, T. Mori, H. Yokota, and M. J. Calaguas, Int J Radiat Oncol Biol Phys 17 (1989) 853.
[6] S. S. Korreman, A. N. Pedersen, L. R. Aarup, T. J. Nottrup, L. Specht, and H. Nystrom, Int J Radiat Oncol Biol Phys 65 (2006) 1375.
[7] E. D. Yorke, L. Wang, K. E. Rosenzweig, D. Mah, J. B. Paoli, and C. S. Chui, Int J Radiat Oncol Biol Phys 53 (2002) 1058.
[8] Y. De Deene, K. Vergote, C. Claeyts, and C. De Wagter, Med Phys 33 (2006) 2586.
[9] P. Haraldsson, A. Karlsson, E. Wieslander, H. Gustavsson, and S. A. Back, Phys Med Biol 51 (2006) 919.
[10] X. A. Li, C. Stepaniak, and E. Gore, Med Phys 33 (2006) 145.
[11] J. Duan, S. Shen, J. B. Fiveash, R. A. Popple, and I. A. Brezovich, Med Phys 33 (2006) 1380.
[12] S. L. Brady, W. E. Brown, C. G. Clift, S. Yoo, and M. Oldham, Phys Med Biol 55 (2010) 2187.
[13] S. Ceberg, A. Karlsson, H. Gustavsson, L. Wittgren, and S. A. J. Bäck, Physics in medicine and biology 53 (2008) N387.
[14] A. Sawant, R. Venkat, V. Srivastava, D. Carlson, S. Povzner, H. Cattell, and P. Keall, Med Phys 35 (2008) 2050.
[15] Y. Suh, A. Sawant, R. Venkat, and P. J. Keall, Phys Med Biol 54 (2009) 3821.
[16] K. Vergote, Y. De Deene, W. Duthoy, W. De Gersem, W. De Neve, E. Achten, and C. De Wagter, Phys Med Biol 49 (2004) 287.
[17] K. Otto, Medical Physics 35 (2008) 310.
[18] A. Mans, P. Remeijer, I. Olaciregui-Ruiz, M. Wendling, J. J. Sonke, B. Mijnheer, M. van Herk, and J. C. Stroom, Radiother Oncol 94 (2010) 181.
[19] S. Korreman, J. Medin, and F. Kjaer-Kristoffersen, Acta Oncologica 48 (2009) 185.
[20] S. Ceberg, I. Gagne, H. Gustafsson, J. B. Scherman, S. Korreman, F. Kjær-Kristofferson, M. Hilts, and S. Å. J. Bäck, Phys Med Biol (2010).
[21] G. Nicolini, E. Vanetti, A. Clivio, A. Fogliata, and L. Cozzi, Phys Med Biol 55 (2010) N347.
[22] J. Zimmerman, S. Korreman, G. Persson, H. Cattell, M. Svatos, A. Sawant, R. Venkat, D. Carlson, and P. Keall, Acta Oncol 48 (2009) 245.
[23] M. Falk, P. M. af Rosenschoild, P. Keall, H. Cattell, B. C. Cho, P. Poulsen, S. Povzner, A. Sawant, J. Zimmerman, and S. Korreman, Radiother Oncol 94 (2010) 218.