

# Image-guided Positioning and Tracking

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## **Abstract.**

Radiation therapy aims at maximizing tumor control while minimizing normal tissue complication. The introduction of stereotactic treatment explores the volume effect, and achieves dose escalation to tumor target with small margins. The use of ablative irradiation dose and sharp dose gradients requires good definition of the tumor and accurate alignment between patient and treatment geometry. Patient geometry variation during treatment may significantly compromise the conformality of delivered dose, and must be managed properly. Setup error, inter-/intra-fraction motion is incorporated in the target definition process by expanding the clinical target volume (CTV) to planning target volume (PTV); while the alignment between patient and treatment geometry is obtained with an adaptive control process, by taking immediate actions in response to closely monitored patient geometry. This article focuses on the monitoring and adaptive response aspect of the problem. The term “image-guidance” will be used in a most general sense, to be inclusive of some important point-based monitoring systems that can be considered as degenerate cases of imaging. Image-guided motion adaptive control, as a comprehensive system, involves a hierarchy of decisions, each of which balances simplicity vs. flexibility, accuracy vs. robustness. Patient specifics and machine specifics at the treatment facility also needs to be incorporated into the decision-making process. Identifying operation bottleneck from a system perspective and making informed compromises are crucial in the proper selection of image-guidance modality, the motion management mechanism and the respective operation modes. Not intended as an exhaustive exposition, this article focuses on discussing the major issues and development principles for image-guided motion management systems. We hope these information and methodologies will facilitate conscientious practitioners to adopt image-guided motion management systems accounting for patient and institute specifics; and to embrace advances in knowledge and new technologies subsequent to the publication of this article.

## **1. Introduction**

Stereotactic radiotherapy, unlike conventional 3D conformal treatment, provides a vehicle to harvest the benefit of volume effect by targeting the ablative radiation beams to the tumor target, and geographically sparing the normal cells from such exposure. With its high dose and sharp gradient, stereotactic radiotherapies are often understood as the noninvasive counterpart of surgery. Just like tumors need to be clearly delineated before their surgical removal, radiation targets are defined carefully prior to the stereotactic treatment, and radiotherapy plan is devised based on such definition. However, in contrast to the surgical procedure, where the

surgeons define and remove the tumors in the same geometry in the operation room, target definition and treatment planning are performed on a pre-treatment simulation image (often CT), yet the therapy is delivered to the instantaneous patient geometry later. The planning and delivery geometry may have discrepancy caused by patient setup, inter- and intra- fraction motion, in addition to other variations such tissue growth or shrinkage. It is thus critical to minimize such discrepancy to realize the improved balance between cure and complication from the surgery-like stereotactic radiotherapy.

Image guidance, in its many modalities and forms, provides either direct and indirect information about the patient geometry at the time of image acquisition ‡. This information is used for (1) patient setup at the beginning of each treatment fraction; (2) patient position monitoring and adjustment between beams; (3) making real-time gating decisions; (4) making real-time delivery adjustment; (5) patient position verification at the end of the fraction; (6) logging and reporting patient motion for retrospective evaluation and assessment. Note that each of this utility involves a combination of image guidance and responsive actions. For example, patient setup is most possibly an iterative process of { imaging current geometry → comparison with planning geometry → patient position adjustment → re-imaging... }

Section 2 provides a general review of image guidance modalities and techniques available or under development. It also discusses image processing techniques involved to extract patient position and tumor motion information from acquired images. Section 3 presents the the control action and focuses on the motion coping mechanisms in response to the information from Section 2.

## 2. Image guidance

### 2.1. Image guidance modalities and operation modes

Rather than an exhaustive list of the available image guidance techniques, this section focuses the major image guidance modalities and the key principles. We intentionally avoid the explicit reference to specific commercial systems unless absolutely necessary to maintain the generality of the arguments.

Image-guidance systems can be classified according to the amount of information contents. Specifically, point-based imaging systems, such as the ones based on detecting locations of implanted Electromagnetic transponders (Willoughby et al., 2006) or photon emission seeds (Alezra et al., 2009), provides location of a single point within the patient. X-ray planar imaging, including floor-mounted (Adler et al., 1999; Yan et al., 2003) or gantry-mounted (Sorcini and Tilikidis, 2006) kilo-voltage (kV) imagers operated in radiographic or fluoroscopic mode or mega-voltage (MV) (Lam et al., 1985) images obtained via electronic portal imaging devices (EPID), generates a 2D intensity array. More recently, in-room volumetric imaging systems, such as MV computed tomography (MV CT) (Langen et al., 2005), kV (Oelfke et al., 2006; Guckenberger et al., 2006; Huntzinger et al., 2006; Letourneau

‡ The term “image” in “image-guidance” is used in the most general sense in this article, and covers a wide scope of observation modes, from point-base measurements to volumetric imaging.

et al., 2005) and MV Cone-beam CT (Groh et al., 2002), and the new development of magnetic resonance imaging (Dempsey et al., 2006) provides volumetric images of patient geometry. Note that this classification is based on the physics capability of the imaging systems and does not reflect how such information is utilized and potentially reduced in estimating the motion information. This distinction can be seen most clearly from fiducial-based motion estimation with X-ray imaging. Despite the acquisition of 2D X-ray images, only locations of the fiducial markers are extracted and propagated to the next processing/control stage - information on any other pixels are discarded. The physical imaging system is 2D, yet the motion estimation behavior presents more resemblance to a point-based system.

In addition to the aforementioned systems which acquires either location or tomographic information of the patient, there are monitoring devices that observe motion of surrogate organs or measure surrogate quantities. Examples of the former include optical monitoring devices measuring displacement of skin surfaces; and surrogate quantities measured include air flow and pressure. The use of surrogate quantities requires an inference model to *convert* the surrogate measurements to patient internal geometry. The construction of such inference model often involves site-specific assumptions and is subject to careful validation. This particular topic is outside the scope of the present article.

## 2.2. Image processing and registration

The instantaneous patient geometry is extracted from the acquired images with an image processing module. The complexity of this module depends on the following factors: (1) input type, *i.e.*, information content of the acquired image; (2) input features extracted from the input; (3) output type, *i.e.*, motion to be extracted; (4) algorithm adopted. Apparently, the process of moving from physics observation (1) to extracted input features (2) is an information-reduction step. The complexity and the relative dimensionality between input features (2) and output type (3) determines the behavior and performance of the motion estimation system. These selections have a direct impact on the balance between model flexibility, physiological soundness *v.s.* performance stability and robustness to noise.

We discuss some typical settings below:

- (i) {input image = Cone Beam CT; input feature = intensity; output = nonrigid motion field; algorithm = nonrigid image registration }

This procedure is best understood in terms of 3D fusion, where the in-room CBCT image is registered to the reference CT volume from simulation to detect large discrepancies in both soft-tissue and bone structure configurations (Borst et al., 2007). With the improved CBCT reconstruction algorithms, the registration can be performed directly in the reconstructed image domain. For Stereotactic treatment setup, optimization is often performed to obtain a global rigid transformation. More specifically, a 3D (CT) to 3D (CBCT) registration is performed to find the transformation  $\mathcal{T}$  to

$$\text{minimize } \mathcal{D}_{3D}(CBCT, CT_{\text{ref}}(\mathcal{T})),$$

where  $\mathcal{D}$  is a distance metric between 3D real-valued intensity matrices - a common choice being the  $L_2$  distance.

- (ii) {input image = planar x-ray; input feature = intensity; output = rigid motion field; algorithm = image registration }

This is the setting where localization is to be estimated based on planar x-ray images, in the absence of fiducial markers (Muacevic et al., 2006). This mode is preferable when the image has reasonably rich structures and gradient content, e.g., in the radiography of the thorax where the projection of the rib cages yields good contrast. As compared to the 3D-3D registration previously described, a forward projector needs to be evoked to connect the 3D reference CT volume to the observed X-ray planar image. This 3D (CT) to 2D (X-ray) registration process aims to find the transformation  $\mathcal{T}$  to

$$\text{minimize } \mathcal{D}_{2D}(\text{planar x-ray}, \mathcal{P}\{CT_{\text{ref}}(\mathcal{T})\}),$$

where  $\mathcal{D}$  is a distance metric for 2D intensity.

- (iii) {input image = non-coplanar x-ray; input feature = triangulated fiducial location; output = rigid motion; algorithm = point set matching }

This is the setting for most fiducial-based motion monitoring systems (Murphy, 2002). The workflow takes two sequential steps: fiducial markers are first identified in each of the planar x-ray images and the 3D locations of the fiducial set are subsequently reconstructed by triangulation. This triangulation yields an instantaneous coordinate set from which motion can be obtained by finding the transformation  $\mathcal{T}$  to

$$\text{minimize } \mathcal{D}_{pos}(\mathcal{X}_{\text{instantaneous}}, \mathcal{T}\mathcal{X}_{\text{ref}}),$$

where  $\mathcal{D}_{pos}$  is the distance metric for 3D position, with the Euclidean distance being a common choice. The reconstructed 3D instantaneous fiducial set location and the reference fiducial set location *in the same order* are denoted in matrix  $\mathcal{X}_{\text{instantaneous}}$  and  $\mathcal{X}_{\text{mboxref}}$  respectively. If such correspondence is not known from the fiducial marker extraction stage, then the optimization should be performed with respect to both transformation  $\mathcal{T}$  and column permutations of  $\mathcal{X}$  to account for the possible pairings of the fiducial map. Such optimization is often achieved by an iterative closest point procedure or its variations, where one alternates between optimizing with respect to correspondence map and finding the best transformation  $\mathcal{T}$  for that given correspondence.

- (iv) {input image = point location readout; input feature = single- point location; output = rigid motion; algorithm = NA }

This is the setting corresponding to intrinsically point-based location monitoring systems or single fiducial-based tracking systems (Kitamura et al., 2002). A typical example of the former is the Calypso tracking system where Electromagnetic transponders are implanted in the vicinity of the tumor target. The transponders are probed sequentially and their positions are agglomerated to provide a single position estimate at an operation frequency. The fusion of multiple transponder locations is transparent to the end-user, so it is reasonable to consider the output as a single virtual location readout for the purpose of image-guided positioning and tracking.

Since the single point location output is itself a time series, motion information can be directly obtained by comparing the instantaneous readout to a reference baseline value.

$$\mathcal{T} = \mathcal{X}_{\text{instantaneous}} - \mathcal{X}_{\text{ref}}.$$

In general, input features with higher information contents admit more complex motion output, in terms of both spatial coverage and motion form. More specifically, when intensity is used as input, such as in settings (i) and (ii), the registration algorithm has access to collective information over the region of interest, and it is possible to recover motion field covering both soft-tissue and tumor target within this region. In contrast, when fiducial location is first extracted as the input feature, the registration module is limited to process local information at those locations only, resulting in a much lower degree-of-freedom (d.o.f) output compared to intensity-based setting.

On the other hand, the flexibility of motion output with higher d.o.f. implies increased sensitivity to noise in the observation images. Therefore, it is desirable to confine the output to a reasonable scope to improve the conditioning and robustness of the motion estimation process. This can be achieved by either explicitly parameterizing the motion field as globally rigid or affine transformation; or by implicitly encouraging physical/physiological properties of the motion output using regularization.

### 3. Positioning and tracking control

The success of stereotactic therapy relies on good consistency between the delivery geometry and planning geometry. An equivalent requirement is the alignment between the instantaneous patient geometry and the treatment beam, as the former reflects real-time delivery geometry and the latter is derived from treatment planning. There are multiple ways to approach such alignment. Conceptually, one may (1) bring the patient geometry to agree with treatment beam; (2) bring the treatment beam to agree with patient geometry; and (3) passively “wait” until they agree. We discuss each of these options below.

#### (i) Aligning patient geometry to treatment beam coordinate.

There are two aspects of this alignment procedure: active positioning of the patient and maintenance of the alignment. Due to patient safety and efficiency considerations, active positioning can only occur sparingly during the treatment course, and is often performed at the setup stage before each fraction and between beams if large enough geometric discrepancy has been detected. Currently, in-room cone-beam CT and planar x-ray imaging are common means of image-guidance for patient setup and couches with 6D (3D translation + roll, pitch, yaw rotation) automated mechanical compensation are available to facilitate fast setup.

Maintenance of this alignment is achieved by immobilizing the patient. For tumors subject to respiratory motion, abdominal compression (Lax et al., 1994) and breath-hold (Barnes et al., 2001) are sometimes used to alleviate the motion effect.

#### (ii) Aligning treatment beam to patient geometry.

This perspective is more viable for intra-fractional and intra-beam alignment since adjustment is applied to the treatment beam rather than the patient. This can be achieved by either manipulate robotics to move the entire linear accelerator, as adopted by the Cyberknife image-guided radiosurgery system (Murphy et al., 2000); or repositioning the multileaf collimator (MLC) (Webb, 2006).

(iii) Passive wait-and-trigger approach for geometry alignment

This passive wait-and-trigger philosophy is the underlying principle for gating-based motion management (Shirato et al., 2000). As patient geometry continuously changes with respect to the beam, there are certain stages, commonly referred to as the *gate*, where the discrepancy between these two geometry is relatively small. When patient geometry falls within this gate, the radiation beam is triggered on. Gating is usually used in managing semi-periodic motion such as those caused by respiration, and gating decisions are often made based on either displacement or phase with respect to the periodic cycle. The development of gating protocols involves the tradeoff between the amount of allowable residual geometry variation within the gate, known as *residual motion*, and the treatment efficiency. Treatment efficiency is quantified with the ratio of the beam-on time within the gate to the overall treatment time, referred to as the *duty cycle* for gated treatment.

Since the positioning and tracking action is executed by mechanical adjustment in response to motion information from imaging, time delay caused by data streaming, processing and mechanical movement is inevitable. This delay is often in the range of a few hundreds of milliseconds. Moreover, image acquisition may be subject to interference from therapeutic beams, occlusion of imaging and limited frequency due to dose considerations, which further increases the delay between the acquisition of the most recent motion estimate and the timing of the actual geometry. The presence of this time delay requires that the tumor position be predicted in advance, so that the positioning or tracking control can synchronize to the actual tumor motion. For tumors with structured motion trajectory, such as respiratory or cardiac-type motion, it is possible to compensate for the delay by developing site/patient/fraction specific models. On the other hand, when tumor motion is approximately random, as in the case of prostate, the benefit of using a complex prediction model is often compromised by the potential risk of model over-fitting and noise sensitivity. An observe-and-hold control mechanism that responds to the most-recent available motion estimate is usually a better strategy.

We also want to note that the above positioning and tracking methods are usually combined in practice. In particular, good initial patient setup (i) is universally desirable as it reduces the amount of beam adjustment required in (ii) and improves efficiency for gating-based system (iii). In addition, beam-off is widely utilized as a fault protection mechanism when desired alignment cannot be maintained in (i) and (ii), which occurs between breath-holds and when motions too large to be coped with treatment beam adjustment.

#### **4. Clinical implementations and management**

Clinical implementations of image-guided motion management involves active participation of radiation oncologists, medical physicists, medical dosimetrists and radiation therapists. As a team, they are responsible for and interact on various levels of the motion management system (Potters et al., 2010).

The oncologists oversee the overall treatment regimen, and are responsible for exploration of treatment options, including applicable image-guided motion management mechanisms. In particular, the oncologists choose the proper IGRT method depending on disease-specific and patient-specific considerations. For example, lung tumor patients with chronic obstructive pulmonary disease may be less compliant with breath coaching, and require more sophisticated adaptive motion management. Patient comfort and the stability/controllability of the potential positioning system needs to be incorporated into balancing between immobilization and adaptive motion management. The oncologists are also responsible of making the tradeoff decision among margin expansion, aggressiveness of tracking/gating, delivery efficiency and the concomitant dose effect for radiographic based imaging (Murphy et al., 2007).

The medical physicists are charged with the technical aspects of the image guided motion management system. They ensure the mechanical precision and accuracy of the IGRT system; the communication among the treatment planning system, image-based motion monitoring and processing system, and the delivery system; and the proper recording and logging of such information. The medical physicists are also responsible to work with the oncologists to develop institution-specific iGRT procedures.

The medical dosimetrists and therapists should be properly educated and knowledgeable about the IGRT system. Particularly, under the supervision of oncologists and physicists, they ensure the consistency between IGRT treatment plan and its execution, including patient immobilization/repositioning and adaptive control.

#### **5. Discussions and comments**

Image guided radiotherapy (IGRT) is a comprehensive system, whose performance involves multiple levels of interactions and tradeoffs. Image-guided positioning and tracking, as the main topic of this article, is the module on the execution end of the IGRT system. Its utility depends on the proper design and implementation of the submodules within itself, as well as its interface and coherence with previous modules in the IGRT workflow, particularly margin design and treatment planning. For example, a treatment plan incorporating motion robustness consideration is less demanding on the subsequent image-guided positioning and tracking module, than a plan without such considerations.

From a system analysis and design perspective, identifying bottlenecks and focusing on reducing or eliminating them are crucial to achieving the utility of image-guided positioning techniques. For example, if the only accessible control variable is 3D translational adjustment, then it would be inadvisable to use an image-guidance system with higher d.o.g than rigid

motion outputs. Using affine or nonrigid motion in this situation should be regarded as over-complex, with respect to the action capability of the control unit, and risks misinterpretation and confusion, in addition to inducing unnecessary sensitivity to imaging noise.

## 6. Summary

To harvest the benefit of volume effect from stereotactic radiotherapy treatment, image-guided positioning and tracking has become prevalent in clinical practice for patient setup, managing inter- and intra- fraction patient geometry variation. The proper design and implementation of image-guidance and the responsive positioning and tracking action requires understanding the system from a comprehensive perspective, and taking into consideration interactions among margin designation, treatment planning and motion management strategies. In addition, patient-specific condition and practical aspects of the clinical setting should be accounted for in choosing image-guidance modalities, motion estimation modes, and the specific techniques for positioning and tracking.

The development and implementation of clinical IGRT protocols is a team effort and involves collaboration among oncologists, physicists, dosimetrists, therapists and other medical/engineering personnel. The fast development of IGRT techniques and the comprehensive multi-level decision making process requires continuing education and training.

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