Fast in vivo volume dose reconstruction via Reference Dose Perturbation

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Abstract. Purpose: Accurate on-line reconstruction of in-vivo volume dose that accounts for both machine and patient discrepancy is not clinically available. We present a simple reference-dose-perturbation algorithm that reconstructs in-vivo volume dose fast and accurately. Methods: We modelled the volume dose as a function of the fluence map and density image. Machine (output variation, jaw/leaf position errors, etc.) and patient (setup error, weight loss, etc.) discrepancies between the plan and delivery were modelled as perturbation of the fluence map and density image, respectively. Delivered dose is modelled as perturbation of the reference dose due to change of the fluence map and density image. We used both simulated and clinical data to validate the algorithm. The planned dose was used as the reference. The reconstruction was perturbed from the reference and accounted for output-variations and the registered daily image. The reconstruction was compared with the ground truth via isodose lines and the Gamma Index. Results: For various plans and geometries, the volume doses were reconstructed in few seconds. The reconstruction generally matched well with the ground truth. For the 3%/3mm criteria, the Gamma pass rates were 98% for simulations and 95% for clinical data. The differences mainly appeared on the surface of the phantom/patient. Conclusions: A novel reference-dose-perturbation dose reconstruction model is presented. The model accounts for machine and patient discrepancy from planning. The algorithm is simple, fast, yet accurate, which makes online in-vivo 3D dose reconstruction clinically feasible.

1. Introduction
The goal of treatment planning is to find an optimal set of machine control parameters (CP), including the beam, gantry, couch, jaw, multi-leaf modulators (MLC), etc., so that the calculated dose distribution satisfies the prescription. Treatment planning requires accurate modeling of treatment components, such as the machine, beam and patient. The machine model includes physical and geometric properties of various LINAC components. The beam model includes curves extracted from the commissioning measurements. The patient model includes both patient setup information and the patient electron density map, which is typically converted from a CT image.

However, even with an optimal plan that involves much hard work, patients may not be treated as planned. There could be patient setup error or uncertainty caused by operators. Patient could experience inter-fraction anatomical changes, such as weight loss or tumor response, or exhibit intra-fraction motion. Additionally, machine behavior, such as beam output, gantry, couch, and/or MLC motion may vary fraction by fraction and different from what was planned.
The modern LINAC is equipped with many hardware/software components that can record in-vivo information during each treatment. For example, machine log files that record information from the sensors, treatment records that contain setup information, online or real-time volume imaging with megavoltage CT (MVCT) or cone beam CT (CBCT), and transit dosimetry using Electronic Portal Imaging Devices (EPID) or linear detectors. Those in-vivo signals provide invaluable information for treatment verification. However, they do not directly translate to dosimetric impact on the patient anatomy, which clinicians are most concerned.

The goal of in vivo dose reconstruction is to translate the delivery errors back to the patient anatomy with a clinically relevant accuracy. This level of accuracy was reported in [1] when using a well calibrated Image Value to Density Table (IVDT) that converts MVCT to a density map to calculate patient dose. The implementation needs to be fast and fully automatic so that it can be used routinely and meet requirements of time-sensitive applications, such as online verification or real-time adaptation. Therefore, we cannot afford to start from scratch as in the treatment planning. On the other hand, the rich information about the machine, beam and patient representations is already contained in the plan dose that is the result of hard work in treatment planning, and it would be a natural candidate as our starting point. Indeed, we are able to incorporate the plan dose as a reference base (reference dose) in a perturbation framework. In this work, we demonstrate a fast in vivo dose reconstruction by perturbing the reference dose with the in-vivo information of each treatment. Note that we assume the availability of machine data and anatomical change for the purpose of this paper; however, that assumption is yet advanced and it requires much effort to realize. Moreover, the dose reconstruction relies on the patient density information retrieved from on-line CT image, such as TomoTherapy MVCT or cone-beam CT (CBCT), which requires comprehensive Image Value to Density Table (IVDT) calibration. As reported from others [1] and verified by ourselves, dose calculation using TomoTherapy MVCT with well calibrated IVDT can fall within 1% accuracy.

2. Methods
The idea of reference dose perturbation (RDP) went back to the adaptive full dose correction as used in plan optimization iterations in the Non-Voxel-based Broad-Beam (NVBB) framework [2]. The reference dose is the base, and the perturbation is some variation, which accounts for fluence and density changes. In fact, the perturbation term is computed as the difference of the approximate dose for two different conditions: the treatment condition and the planned condition.

Let $D(f_0, R_0, x)$ denote the reference dose (plan dose) at location $x$ for the planned fluence $f_0$ and the planning image $R_0$. When the fluence and density change to $f$ and $R$, respectively, the dose incurs a change $\Delta = D(f, R, x) - D(f_0, R_0, x)$. Rather than calculating the true difference $\Delta$, which involves long computation of $D$, we calculate an approximation to the difference $\tilde{\Delta} = \tilde{D}(f, R, x) - \tilde{D}(f_0, R_0, x)$ instead, where $\tilde{D}$ is an approximation to $D$ and ideally, fast to calculate. It was proved that the perturbed dose $D(f_0, R_0, x) + \tilde{\Delta}$ approximates the true dose $D(f, R, x)$ with improved accuracy than the approximation $\tilde{D}(f, R, x)$ does alone by an order of magnitude [2]. The work flow of the RDP algorithm is illustrated in Figure 1.
In principle, the reference plan dose is usually calculated by treatment planning system (TPS) with (time-consuming) model based algorithm, such as the Monte Carlo method or the Collapsed Cone Convolution/Superposition (CCCS) algorithm. Because of the small magnitude in fluence and density perturbation, the accuracy demand on the dose calculation algorithm for the perturbation term is lower than that on TPS dose calculation algorithms, but the speed demand is much higher. As an example of fast approximation, we describe below the Fluence-Convolution Broad Beam (FCBB) dose calculation. As the name suggests, the calculation has a filtering component and a ray-tracing component [3]:

\[
\tilde{D}(f, R, x) = \left( f \otimes k \right) \cdot C(R(x)) \cdot \left( \frac{s_0}{s} \right)^2
\]

Here \(\tilde{D}(f, R, x)\) denotes the FCBB dose at a point \(x\) for the fluence \(f\) and radiological image \(R\). \(s_0\) and \(s\) are the source-to-axis distance and distance of \(x\) from the source, respectively. The fluence is filtered by a lateral kernel \(k\), which is derived from an infinitesimal beam [3], and the term \(C(R(x))\) for attenuation is calculated based on the radiological distance via direct ray-tracing.

3. Results

In Figure 2 and Figure 3, we show the progressive outcome from plan dose, FCBB dose, to RDP, against treatment dose due to change of the fluence map. The plan fluence was a circular field of diameter 10 cm, but the treatment fluence had additional annular modulation (Figure 2) to simulate fluence perturbation. The radiation field was applied on slabs of different densities to simulate inhomogeneity. Then we calculated the treatment dose using FCBB and RDP and compared the results in isodose to the treatment dose [4] calculated by the CCCS method, which is regarded as the gold standard. Figure 3 (a) compares the plan dose with the treatment dose. Without correction, the discrepancy can be as large as 10%. Figure 3 (b) compares the treatment dose calculated by FCBB and by CCCS. The FCBB dose has apparent higher dose in the lung region, as the fluence convolution kernel does not have an accurate model of scattering for inhomogeneity. However, the RDP (Figure 3 (c)) approaches the gold standard within 1% when the change of delivery condition is small from the reference condition.

We retrospectively studied various TomoTherapy treatment cases. The CT, sinogram, plan dose are retrieved from the TomoTherapy archive. The ground truth fraction doses were re-calculated using the CCCS dose calculator as in TomoTherapy TPS, and took 0.5-2 minutes per fraction. The RDP doses were calculated using the presented algorithm, which took about 1-5 seconds per fraction for a 3D volume dose reconstruction.

Figure 4 shows a head and neck case with weight loss and machine output variation. Figure 4 (a) shows the registered online MVCT in green and plan kvCT in red. The red portion in the neck and
shoulder region indicates the mismatch between MVCT and kVCT as a result of patient weight loss. Figure 4 (b) shows that there is a +/- 2% difference in plan sinogram and delivery sinogram due to output variation. Both the anatomical change and machine parameter change caused delivery discrepancy from the plan. Figure 4 (c) shows the difference of the treatment dose and plan dose with ~ 3% dose difference in most region and up to ~5% in the boundary region (skin and air). Using RDP to calculate the treatment dose (Figure 4 (d)), the difference from the ground truth generally falls within 1%, with exceptions in the air region, which is of little clinical significance. Figure 4 (e-f) shows the Gamma Index between the ground truth treatment dose and PDP treatment dose. Figure 4 (e) uses 1%/1mm criteria and Figure 4 (f) uses 3% / 3mm criteria. It is shown that for the most body region, the PDP treatment dose is well within 1%/1mm difference compared with the ground truth. When counting voxels with dose above 10% of the prescribed dose, the Gamma pass rates are above 90% for 1%/1mm criteria and above 95% for 3%/3mm criteria. The voxels with Gamma fail appear mainly outside the body region.

Figure 2 Fluence changed from (a) a uniform, circular field (the plan fluence) to (b) an annularly modulated field (the treatment fluence).

Figure 3 Isodose comparisons with the treatment dose calculated by CCCS. The treatment dose are plotted in solid line. (a) The plan dose, (b) FCBB, and (c) RDP are plotted in dotted line. The “A”, “M”, “B”, and “L” indicate adipose, muscle, bone, and lung, respectively, for the simulated density of different slabs.
4. Conclusions

We presented a fast algorithm for in vivo volume dose reconstruction via reference dose perturbation. It effectively models machine data and anatomical changes into in vivo dosimetry. The automatic and efficient dose reconstruction workflow, combined with an automatic flagging system, enables time-sensitive reviews and clinical decision making.

References

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