Dosimetric quality assurance of highly conformal external beam treatments: from 2D phantom comparisons to 4D patient dose reconstruction

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Abstract. As IMRT technology continues to evolve, so do the dosimetric QA methods. A historical review of those is presented, starting with longstanding techniques such as film and ion chamber in a phantom and progressing towards 3D and 4D dose reconstruction in the patient. Regarding patient-specific QA, we envision that the currently prevalent limited comparison of dose distributions in the phantom by $\gamma$-analysis will be eventually replaced by clinically meaningful patient dose analyses with improved sensitivity and specificity. In a larger sense, we envision a future of QA built upon lessons from the rich history of “quality” as a science and philosophy. This future will aim to improve quality (and ultimately reduce cost) via advanced commissioning processes that succeed in detecting and rooting out systematic errors upstream of patient treatment, thus reducing our reliance on, and the resource burden associated with, per-beam/per-plan inspection.

1. Introduction
Following visionary work by Brahme [1, 2], inverse-planned radiotherapy became a widely clinically accepted technique. From the beginning, it was recognized that the complexities of planning and delivery of segmented beams in conjunction with intended sharp dose gradients would require, in addition to thorough commissioning, empirical dosimetric QA of each plan. This was certainly justified at first as many potential dosimetric pitfalls were subject of active investigation. These days, with IMRT being a staple treatment in the developed world, one could argue that the measurement of every plan is contrary to the principles of quality management and the physicists’ time would be better spent on more thorough commissioning and credentialing of the IMRT delivery chain. We elaborate on this in the last section of the paper. However, given the high human cost of significant errors, the requirements of accreditation bodies, such as ACR/ASTRO in the United States [3], and relentless public scrutiny of “all things radiation”, this is currently largely academic. Therefore, in the meantime it is imperative to continuously improve experimental IMRT dosimetry as it pertains both to system commissioning and individual patients. Realistically, only methods that are widely (i.e. commercially) available have the potential to significantly impact the clinical practice.
2. In-phantom measurements
The current requirement for IMRT program accreditation by the ACR/ASTRO [3] is “patient-specific end-to-end testing”. This wording is somewhat important to those in the US, as it can be argued that it implies physical dose measurement in a phantom replacing the patient on the treatment table. This approach is intuitive to most medical physicists, as this is how all radiotherapy techniques, including IMRT, were historically developed.

2.1. Planar measurements

2.1.1. The humble beginnings – film and ion chamber
The ion chamber point dose measurements are still indispensable for system commissioning [4]. Since only a limited number of data points can be realistically acquired this way, film was originally employed as a planar dosimeter. It has a definite advantage in spatial resolution but also has some drawbacks as an absolute and even precise relative dosimeter, the most significant one being the dependence on the processing conditions. Although continuous use of either silver halide or radiochromic film for IMRT commissioning was advocated in no uncertain terms [5], the same author [6], followed by others [7], have demonstrated that most tasks can be successfully accomplished in the clinic with other tools, such as diode detectors, at least for the previously well-characterized MLCs. The only important parameter that appears to still need film measurements is the intra-leaf leakage. However, in our experience, these measurements are variable from leaf to leaf and noisy enough that a default value from the literature can be used just as well.

One of the practical problems with using silver halide film is the rapid disappearance of film processors from the hospital environment due to widespread switch to digital imaging. Even when a processor is still available, maintaining its stability adequate for dosimetric tasks without a sufficient daily volume of developed films presents an additional challenge. Radiochromic film is also not without its drawbacks, including polarization-induced angular dependence of optical density in the flatbed scanners and heating effects, among them [8]. In either case, using film is a time consuming two-step process, involving off-line scanning and analysis. The desire to obtain dose distributions for near-instantaneous online analysis led to the development of electronic 2D dosimetry arrays.

2.1.2. Planar electronic arrays
The most fundamental characteristics of the array are detector resolution (pixel size) and density (pixel pitch). For IMRT/VMAT QA, a detector ≤1 mm in size can be considered a point, and detector spacing between 5 and 2.5 mm is sufficient to faithfully represent any realistic dose profile [9, 10]. Widely used electronic arrays come with two types of detectors: diodes [11] or ion chambers [12, 13]. Diodes are small (~1 mm) but have energy-dependent response. The ion chambers are energy-independent but have larger active volumes leading to loss of resolution. All planar electronic arrays have angular sensitivity dependence [12, 14, 15]. For the composite IMRT or VMAT measurements it can be negated by either applying angular correction factors (easier done for fixed gantry delivery), or using a phantom that physically compensates for the angular dependence [14, 15]. Even with a perfectly calibrated planar array, when the beam is incident parallel to the detector plane, 2D modulation information degenerates into one dimension. Direct comparison of EPID-measured dose against the prediction can be considered a special case of high-resolution/high pixel density planar dosimeter that is always orthogonal to the beam. In its widely used commercial implementation (Varian Portal Dosimetry), the EPID measurement is compared not to the TPS calculations, but rather to a separate “portal dose predictor” algorithm. Strictly speaking, this violates the integrity of an end-to-end test, since the actual TPS dose distribution is not being evaluated [16, 17]. A great amount of work has been done to address the challenges of portal dosimetry stemming from the fact that a current EPID is fundamentally an imaging rather than a dose measuring device [16, 18], although first experiments on simultaneous imaging and water-equivalent dosimetry have been reported recently [19]. Recently, Gordon et al [20] have analyzed the limits of reliable fluence anomalies detection by
pixel intensity deviations in the presence of geometrical registration errors and EPID noise. Two percent errors can be reliably detected on a minimum of 20 mm² area.

Kruse [21] has demonstrated that gamma analysis [22] passing rates with both an EPID and a chamber array, even with rather tight 2%/2mm dose-error/distance to agreement criteria, failed to reliably differentiate between what the author considered an acceptable and unacceptable plans. In addition, for electronic arrays (both 2D and 3D) with relatively low detector density (true of all ion chamber and diode arrays), the passing rate metric is subject to statistical uncertainties due to the sub-sampling, which can be significant as shown by Bailey et al [23].

2.1.3. Quasi-3D array measurements and 3D dose reconstruction
To counteract some shortcomings of the planar arrays, quasi-3D dosimeters were introduced as the next logical step. Two such devices, both based on diode detectors, are well characterized. The DELTA4 [24-26] has two planar arrays arranged at a right angle and thus the 2D beam modulation information is preserved regardless of the gantry angle. The ArcCHECK [27-30] has a helical diode array and in the beam’s eye view the detector arrangement is essentially invariant with the gantry angle. Certain performance differences stemming purely from the array geometry were pointed out [28].

The DELTA4 detector spacing varies from 5 mm in the center to 10 mm on the periphery. The adjacent ArcCHECK diodes are always 10 mm apart, but the helical arrangement increases the effective detector density in the beam eye view. Three-dimensional dose in a cylindrical phantom can also be estimated via reconstruction using data inputs of a single planar array that rotates in synchrony with the gantry (OCTAVIUS 4D system). All these systems estimate full-density 3D dose in a phantom via various computational reconstruction strategies based on coarsely sampled measurements [24, 31], and these reconstructions (which might be considered “3D dose imaging”) act as surrogates for the more elusive true 3D measurements.

2.1.4. 3D measurements
The only current dosimeters that can directly measure 3D dose distributions are polymer gels, recently reviewed by Baldock et al [32]. They are fabricated from radiation-sensitive chemicals which upon irradiation polymerize as a function of dose. The degree of polymerization can be imaged by various methods. While polymer gels possess some desirable properties, their adoption in the clinic for routine QA and even system commissioning is limited by practical considerations well enumerated by Jursinic et al [14]. Feasibility of relative dosimetry with the optically scanned gel in the Radiologic Physics Center Head and Neck IMRT phantom was demonstrated by Sakhalkar et al [33]. While the strengthening of the IMRT credentialing process would be a welcome development, the requisite accurate calibration to dose, along with robustness to the stresses of physical shipping, remained to be demonstrated. An interesting application of a deformable dosimetric gel is direct validation of dose-warping algorithms [34].

3. In-patient 3D reconstruction
While reliable agreement between the calculated and measured/reconstructed dose in a phantom is the basis for the dosimetric commissioning of an IMRT system, its value for the meaningful patient-specific end-to-end testing is less clear. With the prevalent 3% global dose-error/3mm distance threshold criteria in particular [4, 35], γ-analysis passing rates for either per-beam single-plane [36] or quasi-3D [37] array geometries, had weak, and counter-intuitive, if any, correlation with the conventional clinical DVH metrics. Instances of the 3%/3mm passing rate metric’s failure to detect systematic errors are numerous (see Refs [21, 36, 37] and our paper elsewhere in these proceedings). On the other hand, direct comparison of the planned and deliverable DVHs exhibits higher sensitivity and specificity and is expected to be more clinically meaningful and intuitive to both the physician and the physicist [37-39].
3.1. From fluence
A number of authors have suggested reconstructing the dose on the patient dataset from the measured modulated fluence [16]. Patient dose reconstruction from measured fluence can be broadly divided into two approaches: i) in-air measurements and ii) exit (transit) dosimetry, whereby the fluence modified by the patient is measured.

3.1.1. In-air fluence
For this method, in principle, any planar dosimeter, including film, positioned at any distance from the source can be used [40]. The dose on the patient planning CT dataset can be reconstructed from the dosimeter placed anywhere from the gantry head (for example, COMPASS device [41]) to ≥150 cm from the source (EPID [16]). The reconstruction method is a specialized dose calculation engine ranging in sophistication from pencil beam superposition [40] to Monte Carlo [38, 42, 43]. While the results comparing the dose reconstruction to film dosimetry on a thoracic phantom are encouraging for conventional and IMRT treatments [44], the most advanced and well-characterized EPID-based 3D dose reconstruction systems are not currently commercially available.

3.1.2. In vivo – exit fluence
The reported advantages of the EPID backprojection method is its relative simplicity compared to the Monte Carlo calculations [45] while maintaining the ability to catch errors related to the current state of the patient, which may differ from the time of the planning CT [46]. This method uses exit EPID dosimetry in conjunction with a backprojection reconstruction [47] or reversal of the convolution/superposition dose calculation in an extended phantom [48]. While sufficiently accurate with fixed gantry IMRT even in the presence of inhomogeneities, the backprojection reconstruction appears inadequate for the lung cancer VMAT treatments due to water-based scatter correction kernels [45]. Wendling et al. [45] reported a workaround, whereby a homogeneous dataset was substituted for the heterogeneous patient, which they termed in aqua vivo dosimetry. While improving the agreement between the planned and reconstructed doses, and allowing, for example, to detect the shifts in patient position, this appears to defeat the goal of comparing the planned and deliverable 3D dose distributions on the actual patient CT dataset.

3.2. From dose in phantom
3DVH software (Sun Nuclear) is capable of estimating the volumetric patient dose from the phantom measurements using perturbation methods, thus allowing direct DVH comparison while simultaneously fulfilling the true definition of the end-to-end test. The idea of the method is to use the dose distribution measured inside a QA phantom with a relatively low pixel density detector array to guide (correct when necessary) the TPS dose on the patient dataset, resulting in a high voxel-density patient dose grid [37, 49, 50]. Depending on the delivery technique, two types of phantoms can be used. With fixed-gantry IMRT, measurement-guided composite dose to the patient is reconstructed from the beam-by-beam measurements with a planar array (MAPCHECK) [50]. Dose reconstruction accuracy was verified in a homogeneous phantom with ion chamber and film [50] and using simulation in heterogeneous head/neck patients [37]. While introducing deliberate errors, Carrasco et al. [39] have shown that while per-beam gamma analysis was prone to false positives/negatives, the 3DVH software correctly described the DVH of the plan which included the error.

For VMAT dosimetry, a cylindrical phantom (ArCCHECK) is a more appropriate choice. While the general approach is similar on the surface, this different detector geometry and the arc delivery require an entirely different method for measurement-guided dose reconstruction, called ArCCHECK Planned Dose Perturbation (ACPDP) [31]. ACPDP explicitly relies on the time-resolved nature of the ArCCHECK data, which have updates logged at 50 ms intervals. Along with the dose, the gantry angle determined by the virtual inclinometer [27] is also stored for each data update. At this point, both the low density (~10 mm) dose map on the cylindrical ArCCHECK surface and the gantry angle are known as a function of time. The DICOM RT Plan’s beams’ control points (CPs) are thus
synchronized to absolute, corresponding delivery times, forming the basis of discretizing the delivery process into individual modulated sub-beams at ~2° intervals (which typically corresponds to resolution of 0.2-0.4 sec in terms of delivery time).

A relative 3D dose grid for each sub-beam is independently calculated by convolving a 3D impulse TERMA function throughout the phantom volume with the 3D scatter depth kernels. The next step is the position-dependent measurement-guided dose morphing and absolute scaling that converts full-density relative dose to absolute, using the relevant diode measurements as 3D spatial calibration data points. After that, a full-volume, high-density absolute dose grid is generated in the ArcCHECK phantom by summing all the component, time-resolved sub-beam dose grids. Drawing an analogy with a physical volumetric cylindrical dosimeter, this result could be thought of as a “virtual gel”. A minor post-processing step fine-tunes, if necessary, the virtual 3D phantom dose by a global correction factor that best fits the virtual measurement to the composite (cumulative) doses per diode position. At this point, the high-resolution (sub-millimeter detectors), low density (~10 mm) cylindrical AC dose map is transformed into high voxel density (TPS resolution) volumetric dose grid on a solid cylindrical phantom.

The final step is to use this high density volumetric phantom dose grid to obtain measurement-driven estimate of the dose delivered to the patient. To that end, the voxel by voxel correction factors derived from the ratios of calculated (TPS) and reconstructed doses on the phantom are applied to the TPS dose distribution on the patient CT. This simple approach is based on perturbation theory and basic physics of Compton scatter, and is surprisingly accurate as verified by the direct ion chamber and film measurements in a homogeneous phantom [31] and the ion chamber and small optically stimulated luminescent dosimeters [51] in an anthropomorphic thoracic SBRT phantom. This 3D measurement-guided dose reconstruction system is commercially available.

4. In-patient 4D reconstruction
While 3D dose reconstruction on a patient is an improvement over the standard in-phantom QA techniques, the static approach does not take into account interplay between the organ motion and dynamic delivery. Quantifying any degradation of planned dose due to interplay prior to treatment, based on patient-specific target motion trajectories and patient-plan-specific delivery dynamics, would be of clear value in some cases, e.g. SBRT. Two emerging techniques attempt to address this issue.

4.1. From fluence
EPID-based 4D patient dose reconstruction framework was so far briefly described by Lin et al [42]. Both the tumor (fiducials) and MLC movements were continuously tracked by an EPID. The incident fluence distributions were sorted into their corresponding phases based on the tumor (fiducial) motion pattern detected in real time by the EPID and accumulated as the incident fluence map for each phase. Together with 4DCT, it was then used for Monte Carlo dose calculation using the EPID-detected MLC shapes. Deformable registration was performed to sum up the phase doses for SBRT treatment assessment.

4.2. From dose measurements in phantom
We have demonstrated by direct time-resolved measurements [31] that prototype 3DVH software can accurately estimate 4D VMAT dose in the patient based on the phantom measurements recorded every 50 ms and synchronized with the gantry movement through the virtual inclinometer [27]. The prototype relied on the TPS dose grids exported for each control point but this is mathematically not necessary. Instead, motion can be treated as another perturbation as described in a separate paper in these proceedings. With volumetric time-resolved dose, it is rather simple to accumulate the dose for any moving structure by propagating it through this 4D dose space either as a rigid or deformable object. This approach is currently being generalized for the fixed gantry IMRT delivery.
5. Looking Forward

The majority of the methods summarized in this work serve pre-treatment QA, with some aimed at per-fraction dosimetric verification. Each method has pros and cons in terms of accuracy, effectiveness, efficiency, and basic practicality. However, our summary thus far has been method-centric rather than system-centric, and now we will try to take a step back to see the bigger picture and ruminate on the future of dosimetric QA.

5.1. Modern radiation therapy as a high-tech manufacturing system

Deposition of a highly conformal dose to the patient is the ultimate output of a complex system, comparable to a high-tech manufacturing process involving many steps. At the very least, there are pre-treatment steps (such as imaging, contouring, planning, dosimetric QA) and per-fraction steps (such as patient setup, image guidance, and dose delivery), each one with its own sources of potential error and variation. This analogy to manufacturing allows us to compare our methods to those that helped revolutionize quality in the manufacturing sector starting in the post-WWII era.

Consider one of W.E. Deming’s famous 14 Points [3] that states: “Cease dependence on inspection to achieve quality. Eliminate the need for inspection on a mass basis by building quality into the product in the first place.” Deming recognized that searching for defects at the end of production (i.e. inspection) is too late in the process and, as a result, cost ineffective. One can draw a direct comparison to per-patient dosimetric QA, where detection, diagnosis, and mitigation of issues at the per-plan level clearly qualifies as “inspection on a mass basis” given the sheer number of radiation therapy plans. In these terms, “building quality into the product in the first place” is highly desirable, and would include such goals as optimizing accuracy of the TPS dose algorithms (by the vendors) and their implementation in the site-specific beam models (by the medical physicists), and maximizing accuracy and minimizing variation of the delivery system. As these are the goals of system commissioning, we can imagine that Deming might advise today’s medical physicist to spend a lot more time commissioning the system in the first place, to detect and root out systematic errors by applying tight tolerances and stringent methods. To their part, vendors would strive for higher standards of accuracy and lower incidence of defects and imperfections. In addition, both vendors and clinicians would also employ continual improvement and abandon the concept of “meeting specifications.”

An apt analogy is that of Aguayo’s flywheel [52], wherein a flywheel is made of a rod with three heavy disks. Variation in the radial uniformity of those disks causes the spinning flywheel to wobble at high speeds, an undesirable result. Three strategies could be considered to reduce wobble: 1) inspect each disk, labelling heavy and light regions and using these to customize the mounting to offset the wobble; 2) assemble as usual and just inspect for wobble prior to shipment, rejecting those with wobble; or 3) work with the disk supplier (or fabrication process) to drive down the variation in the uniformity to the point that wobble rarely if ever occurs in the first place. Clearly, option #3 requires more time and resources up front, but is the wisest (and least costly) long term option compared to the very inefficient strategies #1 and 2. Today’s conventions of dosimetric QA resemble #1 and #2 much more than #3.

5.2. Basic tenets of quality violated by common dosimetric QA conventions of today

Basic tenets of quality include: 1) build/design quality into a system, thus reducing reliance on inspection, 2) strive to meet tight tolerances with minimal variation, and 3) abandon the mindset of “meeting specs” and instead work to continually improve. These are largely violated by conventions of dosimetric QA. To start, insensitive metrics – such as the ubiquitous 3%/3mm \( \gamma \)-passing rate [4, 35] – allow many systematic errors to remain undetected [21, 36, 37]. Because system commissioning is in large part based on this metric, it hinders “building quality into the system”, since both vendors and clinical physicists may erroneously believe their system is of higher quality than it is. This by definition violates the goal to strive for tight tolerances and inhibits driving down variation because that variation hides in deceptively high passing rates. Naturally, continual improvement is hamstrung...
as well, as meeting proposed action levels that are based on insensitive metrics [4] gives a false sense that there is nothing to improve. The relatively low performance of multi-institutional tests such as that run by the RPC [53], we believe, is at least in part a direct result of these violations of basic tenets of quality.

5.3. Re-designing commissioning goals

We suggest that the biggest improvements in dosimetric QA can be made by improving the commissioning process and standards. Figure 1 schematizes evolutionary changes in strategy that embrace the major tenets of quality. The noteworthy components of the commissioning process of the future are:

1. Identify specific, testable requirements of the system. Ideally, these would target individual potential errors or singular and testable components. Examples might include, small field PDD, output factors vs. field size and depth, leaf-end penumbra, etc. There will be many of these components and they may vary based on the delivery system (i.e. a conventional MLC-equipped C-arm linac may have a different subset of issues compared for example to a robot-mounted accelerator). The list of requirements should be comprehensive and centered around potential error sources.

2. Each testable requirement will have a test setup designed specifically for it. Clear instructions and, if applicable, common patient or phantom datasets are imperative in order to eliminate risk of variation of how the test is performed on a specific system.

3. Each testable requirement will have a custom method and metric(s) of analysis that are proven sensitive and preferably not binary (pass/fail) but rather a continuum of scores to assess even small variations.

4. When a test is performed and metric results generated, the result(s) will be added to a population database of identical systems. The site can compare itself against the population and see the high performance potential.

5. The population results then serve to drive benchmarking. System highest potentials are quantified and best practices can be shared. This can drive continual improvement not only of the individual system (i.e. improvement of their metric) but also of the population (i.e. reduction in variation and increase in mean performance level). Such metric distributions would also be valuable for regulatory agencies to use in site accreditation/credentialing.

Figure 1: Schematic of an advanced commissioning process.
6. Conclusions
As IMRT technology continues to evolve, so do the dosimetric QA methods. In terms of patient-specific dose QA, we envision that the currently prevalent limited comparison of dose distributions in the phantom by γ-analysis will be eventually replaced by clinically meaningful 3D and 4D patient dose analysis. A number of approaches have been proposed, each with its own advantages and drawbacks, and it remains to be seen which ones will find their way into the wide spread clinical use.

In terms of the bigger picture, we envision a future of QA built up on lessons from the rich history of “quality” as a science and philosophy. This future will aim to improve quality (and ultimately reduce cost) via advanced commissioning processes that succeed in detecting and rooting out systematic errors upstream of patient treatment. Safeguard checks against catastrophic failures such as data corruption during transfer from the TPS to the treatment management system can be easily automated and do not require physical measurements.

7. References
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