Error Detection using EPID-based 3D In Vivo Dose Verification for Lung Stereotactic Body Radiotherapy

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Research Article

Keywords: EPID, Lung SBRT, Error detection, In vivo dosimetry, Quality assurance

Posted Date: January 31st, 2022

DOI: https://doi.org/10.21203/rs.3.rs-1218280/v1

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Abstract

**Purpose** Due to the high doses per fraction and very sharp dose gradients in stereotactic body radiation therapy (SBRT), any errors occur during treatment delivery have a greater radiological impact on patients than in conventional treatment schemes. The current study investigates the efficacy of an EPID-based 3D in vivo dosimetry verification system for volumetric modulated arc therapy (VMAT) SBRT for lung cancer.

**Methods** To investigate the error detectability effectiveness of the iViewDose TM software, thirty errors were intentionally introduced, consisting of dynamic and constant machine errors, to simulate the possible errors that may occur during delivery. The dynamic errors included errors in the output, gantry angle and MLC positions related to gantry inertial and gravitational effects, while the constant errors included errors in the collimator angle, jaw positions, central leaf positions, setup shift and thickness to simulate patient weight loss. These error plans were delivered to a CIRS phantom using the VMAT-SBRT technique for lung cancer. Following irradiation of the error plans, the reconstruction dose distribution calculated by a back-projection algorithm using an EPID image was compared with the original plan dose calculated by the treatment planning system.

**Results** All errors caused by the central leaf positions, setup shifts and patient anatomical changes were successfully detected. Dynamic MLC errors could be distinguished in the 2-5 mm scenarios, but not in the 1 mm scenario. Jaw movements of 5 mm inwards could be detected, but jaw movements outwards could not be detected for all indicator types. However, output errors were not detected with a 1-5% dynamic change which may due to sinusoidal function reduces the magnitude of errors. Dynamic gantry angle and collimator angle errors were not detected in the lung case due to the rotation-symmetric target shape. It was shown that, if the gantry and collimator angle errors are not included, 13 out of 21 (61.9%) and 14 out of 21 (66.7%) errors can be detected by the [EQUATION] and [EQUATION] indicators, respectively.

**Conclusions** In summary, commercial software iViewDose TM is a suitable approach for detecting most kinds of clinical errors in 3D in vivo dose verification for lung cancer SBRT.

1 | Introduction

Advanced radiation therapy techniques, such as volumetric modulated arc therapy (VMAT), have been introduced due to their ability to deliver a high and uniform dose to the target while minimizing the dose to nearby organs at risk (OARs). Together with advancements in immobilization and imaging, VMAT combined with stereotactic body radiotherapy (SBRT), which can deliver large doses in 1-8 fractions, has been widely implemented in a number of clinical institutes [1, 2]. SBRT differs from conventional radiation therapy in that the per fraction doses are much higher and delivered over fewer fractions [3]. To reduce the delivery time, a flattening filter-free (FFF) mode is used to achieve a higher dose rate (e.g., 1000-2000 cGy/min). Because of the high doses per fraction and very sharp dose gradients involved, any error in treatment delivery has a greater radiological impact on the patient than in conventional treatment schemes.
Standard quality assurance (QA) procedures usually include periodic equipment and patient-related pretreatment assessments. Point dose measurement, 2D or 3D gamma agreement index (GAI) analyses were carried out when the approved plans were transferred to a homogeneous phantom. The pre-treatment QA can detect several potential errors; nevertheless, dosimetric uncertainties that may occur during treatment, intra-fraction anatomical variations and position errors may escape from the pre-treatment inspection. A real-time in vivo dosimetry (IVD) system was employed to eliminate these uncertainties. The goal of IVD systems is to detect treatment errors, assist in treatment adaptation and record the actual dose delivered to the patient [4, 5]. Electronic portal imaging devices (EPIDs) are the most widely used IVD devices because they are commonly located on accelerators. Although they were initially designed for setup validation, their dosimetric characteristics made them suitable for pre-treatment [6, 7] and in vivo dosimetric verification [8-10].

There are two methods for performing IVD with EPIDs. The first method, known as the forward method, compares the EPID-measured dose distribution with the treatment planning system (TPS)-predicted exit fluence projected at the EPID level [11, 12]. Although the forward method is relatively straightforward, it does not display the dose distribution in patients. In the second method, a back-projection algorithm reconstructs a 3D dose distribution in the patient and compares it to the planned dose distribution using 3D gamma evaluation [9, 13-15]. Markus Wendling et al. from the Netherlands Cancer Institute-Antoni van Leeuwenhoek Hospital developed a back-projection algorithm for EPID-based in vivo dose measurement systems, which has been used for dose verification of intensity modulated radiation therapy (IMRT) and volumetric arc therapy (VMAT) since 2008 [9, 16]. When in vivo verification involves inhomogeneous tissue volumes, such as the lung, the in aqua method is used [17]. The key feature is that the EPID images obtained during treatment are converted to a scenario in which the patient consisted entirely of water when reconstruction dose in the patient. This software named IviewDose™ allows users to reconstruct the 3D dose distribution in patients with a back-projection algorithm that uses an MV image obtained during irradiation.

At our institution, before IviewDose™ could be used in clinical applications, a commission needed to determine the potential sensitivity limitations to properly use this software. However, vendors generally provide specific guidelines for software implementation, and a few studies describe the sensitivity of different treatment parameter variations [18-20]. In previous publications, possible error scenarios, including dose output, collimator, multi-leaf collimator (MLC), setup, thickness and field size-related inaccuracies, were created to investigate the detectability of IviewDose. However, the errors simulated in those studies are of constant magnitude. During VMAT-SBRT, the MLC positions vary dynamically along with variations in the dose rate and gantry positions. The gravitational effect may trigger extra machine errors that are unique to VMAT delivery. A sinusoid function was used to simulate this error [21, 22]. The aim of this project was to investigate the IviewDose sensitivity of various dynamic and constant machine errors during VMAT-SBRT delivery. To the best of our knowledge, this is the first report on dynamic magnitude errors deliberately introduced in the sensitivity detection of an EPID-based in vivo dose verification system.
2 | Materials And Methods

2.1 | Accelerator and iViewDose software

All measurements were carried out on an Elekta Infinity linear accelerator equipped with an agility collimator consisting of 80 pairs MLC, which features a spatial resolution of 5 mm at the beam isocenter. The EPID is mounted at a source-to-surface distance of 160 cm with $26 \times 26$ cm field size at the beam isocenter. The iViewDose v1.0.1 software works in conjunction with the EPID panel and iViewGT image acquisition system. A unique algorithm named back-projection in the software takes the data captured at the detector panel and back projects it to the depth of the target inside the patient before comparing it with the planned dose distributions. The analysis is automatic and can be reviewed offline at any time with a traffic light system alerting users to potential delivery errors. This procedure requires no additional dose to the patient and has minimal impact on the clinical workflow.

2.2 | Treatment planning for baseline plan

In this study, a CIRS 008A phantom with a lung-equivalent rod containing a spherical target with a diameter of 3 cm was inserted into the lung-equivalent lobe. The planning target volume (PTV) was obtained by expanding the clinical target volume (CTV) by 3 mm, resulting in a spherical target. The dose prescription was 50 Gy divided into five fractions. 6 MV FFF photon beam with 2 arcs, one from 181° to 179° in a clockwise direction and the other from 179° to 181° in a counterclockwise direction, was used. The Monte Carlo algorithm was used for the dose calculation, and the calculation grid was set to 2 mm.

2.3 | Machine error simulation

To achieve the desired fluence in VMAT delivery, both the MLC position and dose rate, as well as the gantry rotation, were modulated. Errors in any of the components will result in serious medical harm to the patient. In theory, machine characteristics will be affected by gravity. Errors will be zero or small enough at angles of 0° and 180° to escape the weekly QA procedure since the standard QA usually perform at these angles, and will be maximizing at 90° and 270°.

In this study, there were thirty intentionally introduced errors, which were classified into two types: dynamic and constant. Errors in the monitor unit (MU), gantry angle and MLC position are dynamically modulated by expressing the gravity effect with a sinusoidal form. The gravity factor $\delta(i)$ can be expressed by the following formula:

\[
\delta(i) = \sin(\frac{CP_i^\circ}{360} \times 2\pi), \ i = 1, 2, 3, \ldots N
\]

where $N$ is the number of total control points (CP) in the plan and $CP_i^\circ$ is the gantry angle of the ith CP. The maximum deviations occur at 90° and 270°, while zero deviations occur at 0° and 180°. Therefore,
the modification function of the MU, gantry angle and MLC position can be expressed as follows:

\[ x'_i = x^0_i - \delta(i) \times EAm, \ i = 1, 2, 3, \ldots N \]

where \( x^0_i \) represents the initial value of the MU, gantry angle or MLC position of the ith CP and \( EAm \) is the introduced error amplitude. The error for the MU ranges from 1-5\%, the error for the gantry angle ranges from 1° to 5°, and the error in the MLC positions ranges from 1-5 mm. \( x'_i \) is the modified value.

Using these equations, the MU decreases from 0° to 180° and increases from 180° to 360°; the gantry angle lags behind the initial angle from 0° to 180° and exceeds the initial angle from 180° to 360°; and the whole MLC shifts toward the gravitational direction without changing the gap between them.

Different scenarios with constant errors for collimator angle, jaw positions, central leaf positions, setup errors and patient weight loss were introduced. To simulate the detectability of collimator angle errors, 1° and 2° angle deviations were made in both clockwise and counterclockwise directions. The Y1 jaw was moved 5 mm inwards (closing) and outwards (wider) to simulate jaw errors. One leaf pair, located at the isocenter, was moved 2, 4, 6, 8, or 10 mm outwards. To simulate setup errors, the phantom was moved to incorrect positions. In this scenario, after acquiring cone beam CT (CBCT) for the initial image registration, the treatment couch was shifted 2 cm toward the anterior, left and superior directions. Patient weight change was simulated with a bolus. A 1 cm bolus was taped to the irradiated area of the phantom. The thirty errors simulated in this study are summarized in Table 1.

After the original plan was finished, the plan was exported to an in-house developed Python software program to modify the parameters to introduce errors described above. Then, the DCMtree application was used to import the modified plans to record and verify system (MOSAIQ version 2.62, Stockholm, Sweden). After irradiation, the transit radiation dose information was collected by the iViewGT image acquisition system and exported to the iViewDose software.

The iViewDose software compares the reconstructed dose of the modified plans with the original plan dose calculated by the TPS. 3D gamma evaluation was used for analysis. \( \gamma_{mean} \), \( \gamma_{1\%} \) and \( \gamma \leq 1\% \) with an evaluation criterion, 3% global dose difference and 3 mm distance to agreement within the 50% isodose surface of the planned maximum dose. Dose reference point (DRP) values were used to compare the point dose measurement results. The alert criteria were 0.5, 2 and 95% for \( \gamma_{mean} \), \( \gamma_{1\%} \) and \( \gamma_{passrate} \) respectively. The DPR was defined in the mass center of the delineated structure. The algorithm calculates the percentage dose difference between the TPS and the dose obtained from the EPID at the defined point. In this study, the acceptance criterion was set at 3\% for the PTV.
### Table 1
Summary of introduced errors.

<table>
<thead>
<tr>
<th>Error type</th>
<th>Errors</th>
<th>Error amplitude</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dynamic errors</td>
<td>MU</td>
<td>range from 1-5%</td>
</tr>
<tr>
<td></td>
<td>Gantry angle</td>
<td>range from 1° to 5°</td>
</tr>
<tr>
<td></td>
<td>All MLC position</td>
<td>range from 1-5 mm</td>
</tr>
<tr>
<td>Constant errors</td>
<td>Collimator angle</td>
<td>1° to 2° (clockwise and counterclockwise directions)</td>
</tr>
<tr>
<td></td>
<td>Jaw movement</td>
<td>Y1 5 mm (closing and wider)</td>
</tr>
<tr>
<td></td>
<td>One central leaf pair</td>
<td>+2, 4, 6, 8, 10 (wider open)</td>
</tr>
<tr>
<td></td>
<td>Setup</td>
<td>2 cm (anterior, left and superior directions)</td>
</tr>
<tr>
<td></td>
<td>Anatomy change</td>
<td>+1 cm bolus</td>
</tr>
</tbody>
</table>

## 3 | Results

### 3.1 | Dynamic errors measurements

The results for the original plan were 0.39, 1.13, 98.4% and 0.6% for the $\gamma_{mean}$, $\gamma_{1\%}$, $\gamma_{passrate}$ and DRP, respectively. The $\gamma_{mean}$, $\gamma_{1\%}$, $1 - \gamma_{passrate}$ and DRP results for dynamic error measurements for all scenarios are shown in Fig. 1. All results for introducing MU errors did not exceed the alert criteria. Changes to the MU had no effect on the $\gamma_{mean}$, $\gamma_{1\%}$ and $\gamma_{passrate}$ results. The DRP shows an increasing trend as the error amplitude increases, with a maximum of 2.6% for a 5% MU error.

All gantry error results were acceptable, as shown in Fig. 1, and the alert criteria were not exceeded for $\gamma_{mean}$, $\gamma_{1\%}$ and $\gamma_{passrate}$. The maximum DRP detected was 0.9% for a 5° gantry angle deviation.

The MLC errors were successfully detected. Only the 1 mm error did not exceed the alert criteria, but the $\gamma_{passrate}$ value was 96.1% lower than the original plan. The $\gamma_{mean}$ values were 0.58, 0.71, 0.83, and 0.96 for the 2-5 mm error scenarios, and all the results exceeded the alert criterion of 0.5. Only the $\gamma_{1\%}$ value in the 5 mm error scenario was over alert criterion of 2, but the results show an increasing trend as the error amplitude increases. The $\gamma_{passrate}$ values, which decreased dramatically, were 88.9%, 75.8%, 63.4% and 54.3% for the 2 to 5 mm error scenarios, respectively. The value for 5 mm scenario was decreased by 44.1% compare to original plan. However, none of the DRPs exceeded the alert criterion.

### 3.2 | Constant errors measurements

All results for constant error scenarios are summarized in Fig. 2. Fig. 2 (a), (b), (c) and (d) represent the $\gamma_{mean}$, $\gamma_{1\%}$, $1 - \gamma_{passrate}$ and DRP, respectively. All results with incorrect collimator angles did not
exceed the alert criteria, and the values were close to the original plan’s results.

There was no alert reminder for the scenario where the Y1 jaw moved 5 mm outwards. The $\gamma_{mean}$ and $\gamma_{passrate}$ values, which were 0.62 and 82.3%, respectively, exceeded the alert criteria for the scenario where the Y1 jaw moved 5 mm inwards, but there were few differences from the original plan for the $\gamma_{1\%}$ value and the DRP.

As shown in Fig. 2, the results for the one leaf pair position error significantly exceeded the alert criterion. For the $\gamma_{mean}$ value, only the 2 mm error scenario was not detected with a value of 0.49, which was close to the 0.5 threshold, while the values for the 4, 6, 8, and 10 mm error scenarios were 0.68, 0.89, 1.1 and 1.33, respectively. The alert light turned red for the 8 mm and 10 mm error cases because the values exceeded the threshold of 1. The $\gamma_{1\%}$ values for the 4, 6, 8, and 10 mm error scenarios were 2.87, 4.01, 5.29, and 6.54, respectively, indicating that the MLC positions were incorrect. The alert light turned red for the 6, 8 and 10 mm error cases because the values exceeded the threshold of 4. When compared to that for the original plan, the worst result was observed for the 10 mm error scenario, which was increased by 4.79 times. The $\gamma_{passrate}$ results were 93.4%, 79.3%, 70.8%, 65.8% and 62% for the various MLC position error cases. When compared to that for the original plan, the lowest value was observed for the 10 mm error scenario, which was decreased by 36.2%. The DRP increased as the MLC position error increased, with values of 6.8%, 13.5%, 20.7%, 28.2% and 35.8% for the 2, 4, 6, 8 and 10 mm MLC position errors, respectively. The alert light turned red in all cases because the values exceeded the threshold of 5%.

A 2 cm setup error was easily detected for the anterior, left and superior directions, with $\gamma_{mean}$ values of 0.53, 1.09 and 0.59, respectively. However, a 2 cm change in the anterior direction could not be detected by the $\gamma_{1\%}$ indicator, with a value of 1.46. The $\gamma_{1\%}$ values for the left and superior directions were 2.79 and 2.25, respectively, which indicated incorrect positions. $\gamma_{passrate}$ successfully identified the 2 cm setup error for the anterior, left and superior directions, with values of 90.6%, 50.3%, and 85.1%, respectively. A 2 cm position change in the left direction could not be detected by the DRP indicator, with a value of 2.8%; however, the anterior and superior directions could be easily detected, with values of 4.7% and 11%, respectively.

The system was quite successful in detecting thickness after taping a 1 cm bolus on the phantom to simulate weight changes in patients, yielding values of 1.07, 2.48, 48.7% and 6.3% for the $\gamma_{mean}, \gamma_{1\%}, \gamma_{passrate}$ and DRP, respectively. Furthermore, $\gamma_{passrate}$ obtains the worst value during the thirty error scenarios.

4 | Discussion

Since in vivo dosimetry verification should an essential part of the standard QA procedure recommended by the International Atomic Energy Agency (IAEA) [23], numerous reports of EPID-based in vivo dose verification have appeared, either reporting experiences of in vivo use [8, 9, 14] or discussing the sensitivity of error detection [18, 19, 24, 25]. However, in those studies, the simulation errors were constant
values that did not account for gantry inertial and gravitational forces during gantry rotation in VMAT. To
the best of our knowledge, this is the first study to investigate the sensitivity of the iViewDose dosimetry
system in lung SBRT for dynamic errors caused by gravity. This study also discussed other errors that
may occur during delivery that are not affected by gantry rotation. The results of the phantom
measurements provided us with sensitivity to possible errors for various iViewDose system scenarios.

In the pre-treatment QA procedure, gantry angles and dose outputs are usually checked daily at static
angles such as 0° or 180° rather than dynamic angles. Possible errors in the output, gantry angle and
MLC positions during VMAT delivery must be represented by periodic functions. Although there are
several types of periodic functions that can be used, the result should not change depending on the
choice of function. A sinusoidal function was used in this study to simulate the output, gantry angle and
MLC position errors.

The results presented here are comparable with those of previous publications. EPID-based in vivo
dosimetry did not detect 1-5% dynamic MU errors in this study; only the DRP showed an increasing trend
as the error amplitude increased, with a maximum of 2.6% for the 5% dynamic MU error. Mijnheer et al.
investigated the detectability of 5% MU errors in four planning models, including lung cases [18]. 5% MU
more could be detected by $\gamma_{1\%}$, MU errors of less than 5% could not be detected. In the sinusoidal form,
the MU errors decreased from 0° to 180° and increased from 180° and 360°, which may counteract the
error results of the full arc. Additionally, the error amplitude may be too small to detect because the
sinusoidal function reduces the magnitude of errors.

As shown in Fig. 1, the gantry angle dynamic errors were not detected for any of the $\gamma_{mean}$, $\gamma_{1\%}$,
$\gamma_{passrate}$ or DRP alerts, and the values were close to the original plan. A similar observation was made
by Mijnheer et al. for lung cases [18]. Theoretically, gantry angle errors are serious errors; however,
sometimes they are not severe enough to have a serious impact on treatment quality. For instance, the
modulation is not overly strong for gantry angle errors, or the anatomy does not change considerably over
angle errors. Both scenarios would yield the same primary dose transmission and could escape detection
in EPID-based transit dosimetry. The model used in this study was a lung case, the body contour was
regular and the target was a small sphere, indicating that even 5° gantry angle errors did not have a
strong influence. Furthermore, except for 90° and 270°, the sinusoidal function used to simulate gantry
errors would reduce the error amplitudes when compared with the constant errors. This would further
reduce the errors in the results.

The dynamic MLC errors were quite successfully captured by the iViewDose dosimetry system for lung
SBRT plans. $\gamma_{mean}$ and $\gamma_{passrate}$ could distinguish all the dynamic MLC error scenarios, except for the 1
mm error case. $\gamma_{1\%}$ exceeded the alert threshold only for the 5 mm error scenario, and none of the DRP
results exceeded the alert criterion. Oliver et al. showed that shifting MLC banks in opposite directions
resulted in smaller or larger field shapes that tended to be more impactful than shifting systematic leaf
banks in the same directions or random MLC errors [26]. In this study, dynamic MLC errors were caused
by shifting all leaf banks toward the gravitational direction without changing the gap between them. The
MLC apertures were not fundamentally alerted, which could make sense for none of the DRP results exceeded the alert threshold.

In contrast to previous studies [18, 19], one leaf pair situated at the position of the isocenter position errors was also designed in our study. The error amplitudes were constant and shifted, resulting in larger field sizes. \( \gamma_{mean} \) and \( \gamma_{1\%} \) could distinguish all MLC error scenarios with the exception of the 2 mm error case, and for the indicator \( \gamma_{passrate} \) and the DRP, all the errors could be captured by EPID-based transit dosimetry. This was consistent with the results of Mijnheer et al. [18]. In that study, single and multiple leaf position errors in lung cases were detected and were much more sensitive than other cases. Small changes in the MLC could be captured due to the smaller average field size.

Incorrect collimator angles for all scenarios escaped from the test. Mijnheer et al. [18] reported that collimator angle errors in lung plans could be detected by the indicator \( \gamma_{1\%} \) when the angle was changed from 20° to 340° or vice versa; however, the DRP did not have a considerable change. In contrast, in the Yedekci et al. [19] study, all incorrect collimator angles from 0.5 to 5° escaped detection in the prostate SBRT plan, which was consistent with our findings. Although the tumor site differed from the lung plan, the target shapes were similar. The targets were regular, and rotation symmetry might be the main reason for the EPID-based transit dosimetry detection failure.

The Y1 jaw moving 5 mm inwards could be detected by the indicators \( \gamma_{mean} \) and \( \gamma_{passrate} \) but the outwards scenario could not be detected for all kinds of indicators. This makes sense because, even when the jaw moves outwards, the leaves still there to keep the field shape and consequently have no influence on the transmit dose. The jaw moving inwards has the opposite effect; moving inwards would have an impact on the field edge fluence, further influencing the \( \gamma_{mean} \) and \( \gamma_{passrate} \) values. The DRP was defined in the mass center of the PTV, which would be affected if the jaw blocked the define point.

A 2 cm setup error for the CIRS phantom in the anterior, left and superior directions was successfully detected. This makes sense when the phantom is moved 2 cm to the left, which would increase the thickness of the body during the irradiation fields and result in a decrease in the dosimetry received by the EPID. Moving two centimeters in the superior direction moved the superior spherical target, resulting in an increased isocenter dose and a decreased anterior shift SSD, resulting in an increased DRP.

Thickness errors were detected by all four indicators in the EPID-based dosimetry system for lung SBRT plans. After taping a 1 cm bolus on the phantom to simulate the weight changes that frequently occur during a treatment course, the expected dose decreased in the isocenter due to an increase in built-up dose. The DRP decreased from 0.6% to -6.3%, which was consistent with the results of the Mijnheer et al. [18] study, which showed a 6.9% difference for lung cases. Furthermore, \( \gamma_{passrate} \) obtains the worst value during all error scenarios simulated in this study.

Gamma analysis is widely used to compare the planned and measured dose distributions in the clinic. Christos Moustakis et al. recommended thin CT slices and tighter criteria (2%/2 mm), as well as local
gamma evaluation in hypofractionated VMAT irradiation for 3D in vivo dosimetry verification[27]. Gamma analysis combines dose-difference and distance-to-agreement in a mathematical model, and contains no information on the dose distribution. Van der et al. compared DVH-based and multiparametric γ-based methodologies for the iViewDose system for pelvic VMAT plans and concluded that $\gamma_{mean}$ and $\gamma_{pssrate}$ were strongly correlated with DVH indicators and equivalent for 3D in vivo dosimetric verification of VMAT pelvic treatment [28]. The iViewDose system has four indicators: $\gamma_{mean}$, $\gamma_{1\%}$, $\gamma_{pssrate}$ and DRP. $\gamma_{mean}$ can detect 13 serious errors out of a total of 30 errors (43.3%), $\gamma_{1\%}$ can detect 8 out of 30 errors (26.7%), $\gamma_{pssrate}$ can detect 14 out of 30 errors (46.7%), and the DRP can detect 8 out of 30 errors (26.7%). Because the site used in the study was lung SBRT, with a regular and rotation-symmetric target, detection failure was expected for gantry angle and collimator angle error scenarios by EPID-based transit dosimetry. Excluding those errors, the detection rates for $\gamma_{mean}$, $\gamma_{pssrate}$ and DRP were 13 out of 21 (61.9%), 8 out of 21 (38.1%), 14 out of 21 (66.7%), and 8 out of 21 (38.1%), respectively. The main error scenarios not detected by the $\gamma_{mean}$ and $\gamma_{pssrate}$ indicators were output errors. As previously discussed, one possible reason was that the amplitude of the simulated MU errors was too small to detect in sinusoidal form. Aside from the output errors, these indicators could detect almost all the error scenarios, with detection rates of 13 out of 16 (81.3%) and 14 out of 16 (87.5%). The ability of the $\gamma_{1\%}$ indicator and the DRP to serve as additional reference indicators appears to be limited, as no error was captured solely by either of them.

Alert criteria are generally a compromise between the need to detect all deviations and the workload required to analyze alerts. Some deviations were technically classified as serious errors, but they were not severe enough to have a serious impact on treatment quality. Treatment site specificity should be considered an important factor when alerts occur, and the alert threshold should also be related to a specific site [14].

5 | Conclusion

In summary, this is the first attempt to introduce dynamic errors in the output, gantry angle and MLC positions related to gantry inertial and gravitational effects, as well as constant errors in the collimator angle, jaw positions, central leaf positions, setup shift and thickness to simulate patient weight loss to test the sensitivity of iViewDose software for 3D in vivo dose verification of lung SBRT plans. All errors caused by central leaf positions, setup and patient anatomical changes were successfully detected. The $\gamma_{mean}$ and $\gamma_{pssrate}$ indicators were able to distinguish dynamic MLC errors in the 2-5 mm scenarios, but not in the 1 mm error case. The Y1 jaw moving 5 mm inwards could be detected, but the outwards scenario could not be detected for all kinds of indicators. Due to the rotation-symmetric target shape in the lung case, gantry angle and collimator angle errors go undetected. It was shown that if the gantry and collimator angle are not included, 13 out of 21 (61.9%) and 14 out of 21 (66.7%) errors can be detected by the $\gamma_{mean}$ and $\gamma_{pssrate}$ indicators, respectively. The main error scenarios that were not detected were output errors, which could have been due to small error amplitudes.
It can be concluded that the commercial software iViewDose is a suitable approach for detecting most kinds of clinical errors in 3D in vivo dose verification of lung SBRT delivery. Errors in output change that were not detectable should be considered in clinical applications. Further investigations are being conducted in the clinic to determine the specific site alert criteria for lung SBRT cases.

**Abbreviations**

SBRT
stereotactic body radiation therapy

VMAT
volumetric modulated arc therapy

OARs
organs at risk

FFF
flattening filter-free

QA
quality assurance

GAI
gamma agreement index

IVD
in vivo dosimetry

EPIIDs
Electronic portal imaging devices

TPS
treatment planning system

IMRT
intensity modulated radiation therapy

MLC
multi-leaf collimator

CTV
clinical target volume

PTV
planning target volume

MU
monitor unit

CP
control points

CBCT
cone beam CT

DRP
Dose reference point
IAEA
the International Atomic Energy Agency.

**Declarations**

**Ethics approval and consent to participate**
Not applicable

**Consent for publication**
Not applicable

**Availability of supporting data**
The datasets generated during and/or analyzed during the current study are not publicly available, but can be inquired from the authors.

**Competing Interests**
The authors have no conflicts of interest to disclose.

**Funding**
Not applicable

**Author Contributions**
Conception and design: Juntian Shi and Yimin Liu. Delivery plan creation: Jianghua Huang and Jinyan Hu. Plan delivery: Shijie Liu and Huanping Lu. Data analysis: Fengying Gong and Xiuxiu Wu. Manuscript preparation: Jianghua Huang. All authors have read and approved the final manuscript.

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Figures
Fig. 1 Results of dynamic error measurements based on $\gamma_{\text{mean}}$, $\gamma_\text{v1}$, $1 - \gamma_{\text{pass rate}}$ and the DRP. The dashed lines indicate alert criteria.

Figure 1

See image above for figure legend
Fig. 2 Results of constant error measurements based on $\gamma_{\text{mean}}$, $\gamma_{1\%}$, $1 - \gamma_{\text{pass rate}}$ and the DRP

Figure 2

See image above for figure legend