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Image guided radiotherapy

IGRT induced dose burden for a variety of imaging protocols at two different anatomical sites

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ABSTRACT

Background and purpose: Increase in positioning accuracy and treatment adaptation is supported by image guidance. The downside is the concomitant imaging dose. In this study, we report on the total dose picture for different styles of image guidance.

Materials and methods: Dose was measured in the Alderson phantom using TLD's. IGRT technology investigated included CBCT at the linac and simulator, multislice-CT and kV and MV planar imaging. Clinically used imaging protocols were applied and the total dose picture was assessed for four different sequences of imaging for a prostate and a head and neck treatment.

Results: The different imaging geometries for the various imaging modalities resulted in fairly different dose distributions. Head and neck doses up to 100 mGy and higher were found for portal imaging and multislice-CT. Depending on the IGRT sequence used maximum total dose varies between 120 and 1500 mGy. In prostate maximum doses between 40 and 100 mGy were found for portal imaging and CBCT at the linac. Here the maximum total dose varies between 120 and 2250 mGy depending on the sequence used.

Discussion: Factors like patient dimensions, age and sex can influence the applicability of presented values. Careful consideration of imaging dose especially for very intense imaging sequences is recommended.

Image guidance techniques have a major impact on radiotherapy practice of many cancer patients worldwide. The use of PET-CT, CT or 4D-CT for improved target definition in the planning phase as well as assuring accurate treatment via CBCT, stereoscopic imaging, portal imaging or kV planar imaging are examples of the increasing popularity of image guidance. Examinations during the treatment follow up might also be included into this category. A survey performed by Simpson et al. [1] showed that use of IGRT technology in the US has increased during the last years. It is expected to grow further and one reason for that is that the full potential of IMRT can just be exploited if it is combined with IGRT. The additional use of IGRT supports safer IMRT application and additional margin reductions might be enabled.

Benefits from intense use of imaging are not only restricted to a gain in positioning accuracy; it also gives the opportunity to adapt the treatment to each patient. A current trend is a more individualized cancer treatment responding, e.g., to anatomical changes during treatment, and repeated imaging is then mandatory [2–5]. Intensive use of IGRT technology in an adaptive radiotherapy (ART) setting is described in the literature. For example, Yan et al. [3] reported acquiring on average 18 CTs in addition to the planning CT and 120 portal images for their prostate patients. Lei and Wu [6] acquired 37 CBCTs per patient in another study. For head and neck daily use of portal images were reported by Court et al. [7].

A downside to the increased imaging is the additional dose burden during the radiotherapy increasing the risk for deterministic effects to organs at risk in- and outside the primary treatment area and the risk for stochastic effects which cannot be excluded even for small doses. In IMRT as well as in ART the additional dose burden which might arise from the use of ionizing radiation for imaging purposes is not to be forgotten. This dose is often neglected; it is simply thought to be small compared to the radiotherapy dose. However, when adding up the contributions from all imaging sessions during intensive IGRT, the total dose might be considerable and if not accounted for deterministic effects are possible. Also, like the concomitant dose produced by linac head leakage and scatter dose, imaging dose contributes to the dose burden in- and outside the treatment area [8]. As pointed out by the AAPM task group report 75 [9] an overview of the imaging dose should be compiled and ways to reduce and optimize dose should be found. Especially

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when organs at risk doses are already pushed to the limit in IMRT the dose burden due to imaging might have additional impact on side effects.

Reports on the total dose picture are still scarce. Most reports on IGRT dose studies are either limited to one device or to one indication. Articles about dose to anthropomorphic phantoms and patients are covering, e.g., Varian equipment [10–12], Elekta equipment [13–17], dose to head and neck [10–13,15,16], chest [10–13,16] and abdomen [10–13,16,17]. Provided that an institution uses similar protocols as reported in those articles, the total doses can be calculated with a limited accuracy by multiplying those doses per scan by the number of imaging studies.

This work aimed at performing a dose assessment of the additional imaging dose to patients on all imaging devices in our department by measuring organ doses in a phantom. We also compared the total dose for certain sequences of imaging sessions. It focuses on prostate patients and head and neck patients where IMRT and IGRT are extensively explored. Historically conformal radiotherapy and IMRT is/was based on MV portal imaging. Therefore traditionally used EPID were included in this study.

Materials and methods

For our imaging dose estimation, clinically used head and neck and prostate protocols were applied[1] for all devices investigated (see below).

kV imaging

Image acquisition with CBCT at the linac

Measurements were performed with a CBCT, type XVI™ (Elekta Ltd., Crawley, UK) attached to a linear accelerator (Synergy™, Elekta Ltd., Crawley, UK). The detector-source distance was 153.6 cm. The imaging panel size was 41×41 cm^2 and provided an image size of 1024×1024 with 16-bit pixel depth. For the head and neck CBCT protocol a centered panel position and beam collimation was used. For the prostate an offset geometry was applied to obtain a larger axial field of view (FoV). Both planar and volumetric imaging were investigated.

Image acquisition with CBCT at the simulator

The second cone-beam device that was considered in the study was a therapy simulator (Simulix Evolution™, Nucletron, Veenendal, Netherland). This device was included to have an additional comparison to linac attached CBCT devices and to take into consideration that still some departments might not have regular access to a CT and therefore use the simulator for planning purposes. The detector-source distance was 152 cm. The amorphous silicon detector size was 41×41 cm^2 and it provided a resolution of 1024×1024, 16-bit. A wedged filter was used to balance the X-ray intensity on the detector.

Image acquisition with multislice CT

A multislice CT used for radiotherapy treatment planning and CT simulation (Siemens Somatom Plus 4 Volume Zoom, Siemens, Erlangen, Germany) with an adaptive detector array was used. During head and neck scanning a sequence with 4 mm beam collimation was used, while a spiral mode with 2.5 mm beam collimation was applied for the prostate.

[1] Tables on parameters for the protocols and correction factors can be found as Supplementary material in the appendix on the journal website.

Image acquisition with ExacTrac X-ray (stereoscopic imaging)

ExacTrac integrates an infrared tracking system with in-room X-ray imaging capabilities [18]. Two orthogonal X-ray tubes are mounted on the floor. The X-ray beam crosses the isocentre and projects X-ray transmission onto the ceiling mounted flat panels. The size of the amorphous flat-panel detectors are 20.4×20.4 cm^2. In our case the source to isocentre distance (SID) was 2.3 m and the source to detector distance (SDD) was 3.6 m.

MV imaging

The standard portal imager (PI) of one of our linear accelerators (Synergy™ S, Elekta AB, Crawley, UK) was used. The detector specifications are similar to the kV imaging device, except the PI uses a 1 mm copper filter [19]. A double exposure technique was evaluated, which results in additional dose in- and outside the treatment field.

Phantoms

For dosimetry we used the Alderson anthropomorphic phantom (RSD, Long Beach, US), in which it is possible to place detectors inside. The phantom consisted of 34 2.5 cm thick axial slices made of tissue equivalent material. The isocentre was marked with small steel BBs using a three point set up. The alignment marks on the surface of the phantom were used to position the phantom for each imaging device with the corresponding in-room laser system. For the head and neck imaging isocentre position was chosen in the centre of slab number 7 and for prostate in the centre of slab number 33 (see also Fig. 1).

Measurement and evaluation protocol

TLD handling

Lithium fluoride TLD chips (TLD 100, Harshaw) with dimensions of $3 \times 3 \times 1$ mm^3 were used for the measurements. The TLD's had a sensitivity variation within ±3%. TLD calibration was done in a ^{60}Co source using a field size of 10 cm × 10 cm. The TLD's were placed in 4 cm depth in a 16 cm × 18 cm Perspex block at isocentre position (SID of 80 cm). For each measurement point three randomly selected TLD's were packed and positioned at a certain organ position in the Alderson phantom. Before reading the TLD's they were pre heated at 100 °C for 10 min. The TLD's were read in a Harshaw TLD 5500 reader (Thermo Fisher Scientific, Erlangen, Germany).

Beam quality estimation

Numerous kV beams with different beam qualities were used within this study. To account for the varying sensitivity of the TLD's, each half value layer (HVL) was measured and correction factors for TLD's calibrated in a ^{60}Co beam were applied. HVL was determined by using a multipurpose detector and a multimeter (Barracuda, RTI, Mölndal, Sweden). As the beam passes through matter the HVL will increase due to beam hardening. We therefore measured the HVL's at 0, 5 and 10 cm depth of a solid water phantom and calculated the mean value, assuming that each TLD is also hit by these different hardened beam qualities. The detector was irradiated in isocentre position, with or without solid water as build up material, and with the same kV settings, bow tie filter and collimation as used in the clinical protocols. The beam quality correction factors[1] were interpolated from the data of Nunn et al. [20] and Mobit et al. [21].

Measurement protocol and TLD positions

TLD's were placed on the surface and inside of the Alderson phantom. These positions were selected in a way to represent doses to various organs at risk. The 25 measurement positions