Calculation of organ doses from radiotherapy of breast cancer using Monte Carlo methodology

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Purpose

Breast cancer comprises the second deadliest type of cancer for the female population of the United States of America [1, 2]. Despite the recent increase in breast cancer incidence, mortality associated with this type of cancer has declined, because of the improvement of diagnostic and therapeutic techniques [1]. Radiation therapy comprises one of the most effective treatments for breast cancer. Reported experience has suggested that a local control rate of 90% to 95% can be achieved for early stage cancer patients [3]. The number of patients receiving radiation therapy for breast cancer has increased during the last decades [4]. About 25% of cancer cases treated by radiotherapy departments each year are associated with breast cancer [5]. The assessment of the risk for second cancer induction might be of clinical value due to the good prognosis for breast cancer cases. The first step towards the determination of second cancer induction risk is the accurate determination of radiation dose received by the patient’s radiosensitive organs.

The aims of the current study were to:

• Construct a simulated medical linear accelerator (LINAC) model.

• Verify the simulated beam's dosimetric characteristics by comparison with data measured on an actual LINAC using a water phantom.

• Use the beam model to simulate breast cancer radiotherapy with lateral and medial fields and to subsequently estimate the radiation doses to organs at risk.

Methods and Materials

Construction of the medical linear accelerator model

The MCNP 5 Monte Carlo (MC) code was employed for the simulation of a 6 MV (Philips SL 75/5, Philips/Elekta, The Netherlands) medical LINAC. The simulation incorporated all the main beam modifying components of the LINAC head. In order to save computer time the simulation was performed in two steps [6]. Figure 1 on page 4 shows the relative positions of the modeled components.
In the first step a 6 MeV electron beam impinged on a heavy metal target. The generated bremsstrahlung photons crossed the flattening filter and its holder and were tallied just under the flattening filter holder. Figure 2 on page 4 shows the annular tallying surfaces and their position under the flattening filter holder. The number of photons crossing the tallying surface was recorded separately for thirty-two 250 KeV energy bins spanning an energy range from 0 to 8 MeV.

The spectra that were calculated in the first step of the simulation were re-emitted from a point source. The distance separating the point source from the simulated secondary collimator was the same with the distance between the actual secondary collimator and the upper surface of the heavy metal target. Each spectrum was emitted in the same angular aperture that it was measured by the annular tally rings. Figure 3 on page 5 shows the setup used in the second step of the simulation.

**Verification of the simulated beam’s dosimetric properties**

Percentage depth dose (PDD) and dose profiles measured on a water phantom (RFA-300, Scanditronix Wellhofer, Uppsala, Sweden) using an ion chamber (CC13-S, Scanditronix Wellhofer, Uppsala, Sweden) were compared to calculations obtained in simulated geometries including a water phantom. Figure 4 on page 5 depicts the setup of the simulated detectors used inside the water phantom for the calculation of PDDs and dose profiles. Calculations were performed for a square 20 x 20 cm$^2$ field. Source to surface distance (SSD) was set to 100 cm for all actual measurements and simulations. Dose profiles were obtained at the depth of 10 cm ($d_{10}$) in the actual and the simulated water phantom as well.

**The mathematical phantom**

The BodyBuilder™ software (White Rock Science, Los Alamos, New Mexico, USA) was employed for the generation of a mathematical phantom representing an average female patient. The phantom was modified in order to include all twenty-eight radiosensitive organs recently defined by the International Commission on Radiological Protection (ICRP) [7]. Lymph nodes [8] and salivary glands [9] were added to the phantom geometry. The dose imparted to the red bone marrow, bone surface, extrathoracic tissue, muscle and oral mucosa was approximated suitably by the dose imparted to nearby organs.

**Breast cancer radiotherapy simulation**
The second part of the LINAC geometry in conjunction with the mathematical phantom, were used for the simulation of breast radiation therapy. Each therapy consisted of irradiation with a lateral and a medial field of the same size. Medial and lateral fields with a dimension of 16 x 10 cm$^2$ and 20 x 10 cm$^2$ were defined in cooperation with an experienced radiation oncologist. The gantry angle was 286° and 115° for the medial and lateral field respectively, so that the inner field edges matched in the transverse plane of the phantom. The target tumor was assumed to lie at the center of the left breast and the prescribed tumor dose was set to 50.4 Gy for each therapy. Any lateral or medial projection contributed 50% to the total tumor dose. SSD was selected to be 100 cm for all projections. Figure 5 on page 6 shows the therapy setup for the 10 x 16 cm$^2$ fields. Mean doses to all radiosensitive organs were calculated.

**Images for this section:**

![Diagram](image)

**Fig. 1:** For computer time management reasons the LINAC head simulation was realized in two steps. The transparent plane designates the two sets of components used in each step of the simulation. The inset shows a magnification of the primary collimator proximity.
Fig. 2: The angular differentiation of the photon spectrum crossing the flattening filter holder was determined by the annular tally rings just underneath the holder. The rings appearing on the right side of the image had the same width of 0.2 cm.

Fig. 3: A point source located at the spot where the bremsstrahlung photons were generated, emitted the photon spectra produced in the first step of the simulation. Each photon spectrum was emitted exactly at the direction it was measured. The cone surfaces contain the different spectra measured by each tally ring. Each color designates a different photon spectrum.
**Fig. 4:** The simulation geometry for the calculation of a) PDDs and b) lateral dose profiles. Detector tallies inside the water phantom are visible. The size of the tally cells was kept small to achieve resolution but large enough to reduce statistical uncertainties in a reasonable amount of time.

**Fig. 5:** The therapeutic a) medial and b) lateral 10 cm x 16 cm fields employed for breast cancer radiotherapy simulation. The gantry angles were suitably selected so that the right breast was outside the primary photon beam.
Results

Verification of the simulated beam's dosimetric properties

The calculations performed for the verification of the beam's dosimetric properties agreed very well with the measured data on a water phantom. Figure 1 on page 7 shows PDD calculations for a 20 x 20 cm$^2$ field superimposed to measurement data for the same field size. Doses were normalized to $d_{10}$. Local differences did not exceed 2%. Figure 2 on page 8 demonstrates the comparison between the measured and the calculated lateral dose profile for the 20 x 20 cm$^2$ field at $d_{10}$. Local differences were less than 2% in the plateau region of the profile.

Breast cancer radiotherapy simulation

As expected the primary irradiated left breast received nearly the prescribed dose of 50.4 Gy. The dose to the contra-lateral breast was found to be equal to 1.4% of the prescribed dose for both field sizes. The radiation dose to critical structures from irradiation with the small and large field sizes is presented in figures 3 on page 9 and 4 on page 10, respectively. For the small field dimensions of 10 x 16 cm$^2$, the doses to organs partially included in the therapeutic field (red bone marrow, skin, lung, bone surface) ranged between 148.6 and 1935.7 mGy. Organs outside the collimated field received lower doses of 18.2-333.3 mGy. The field size was found to influence organ doses. An average increase of 19.6% was observed when the therapeutic field area was increased by 25%.

Images for this section:
Fig. 1: Comparison of the calculated percentage depth dose for the 20 cm x 20 cm to the corresponding percentage depth dose measured on a water phantom.
Fig. 2: Comparison of the calculated lateral dose profile for the 20 cm x 20 cm to the corresponding lateral profile measured on a water phantom. Doses were normalized to the dose at the central point of the profile.
**Fig. 3:** Radiation doses to sensitive organs due to breast cancer radiotherapy with a lateral and a medial field of dimension 10 cm x 16 cm. Organ doses are presented in decreasing order and a logarithmic scale has been utilized to contain the wide dose range observed. The remainder includes all radiosensitive organs for which the ICRP has not published a radiation induced cancer risk.

![Graph showing radiation doses to sensitive organs](image)

**Fig. 4:** Radiation doses to sensitive organs due to breast cancer radiotherapy with a lateral and a medial field of dimension 10 cm x 20 cm. Organ doses are presented in decreasing order and a logarithmic scale has been utilized to contain the wide dose range observed. The remainder includes all radiosensitive organs for which the ICRP has not published a radiation induced cancer risk.
Conclusion

- The constructed MC model of a 6 MV therapeutic photon beam can be used for the direct organ dose calculation associated with radiotherapy of breast cancer. The generated computerized model provided accurate dose estimations for all radiosensitive tissues partially included or excluded from the treatment field.

- The size of the therapeutic fields used may affect organ doses considerably.

- These organ dose calculations may be used by medical physicists and radiotherapists to estimate the magnitude of secondary cancer risk. Accurate risk estimations may be of clinical value for the follow-up of patients surviving from breast cancer.

References

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