Assessment of uncertanties associated with Monte Carlo simulation of clinical CT examinations

Azadeh Akhavanallaf[1], Tianwu Xie[1], Habib Zaidi[1,2,3,4*]
[1] Division of Nuclear Medicine and Molecular Imaging, Geneva University Hospital, CH-1211 Geneva 4, Switzerland
[2] Geneva University Neurocenter, Geneva University, CH-1205 Geneva, Switzerland
[3] Department of Nuclear Medicine and Molecular Imaging, University of Groningen, 9700 RB Groningen, Netherlands
[4] Department of Nuclear Medicine, University of Southern Denmark, 500 Odense, Denmark
*Corresponding author: Habib.Zaidi@hcuge.ch

***Abstract*— Monte Carlo simulations-based estimation of organ level radiation dose from diagnostic imaging procedures using patient-specific models is commonly considered the gold standard for the implementation of a reliable framework that can be adopted in clinical setting. The main purpose of this work is to analyze the uncertainties of the estimated organ doses as the parameters used as input to the simulation are obtained from limited DICOM CT image information in comparison with accurate values derived from CT raw data files. We investigated the dosimetric impact of modeling tube current modulation schemes (complete vs. longitudinal only) and tube start angle position implemented in a validated dose tracking Monte Carlo code. Data of an anthropomorphic physical phantom examined using an abdominal helical CT scan was employed. The absolute mean difference between organ-level doses resulting from accurate raw data-based simulation and approximate image-based simulation corresponding to one rotation in the abdominal region is about 6.92% where this difference for the gallbladder exceeds 19%.**

***Index Terms*— CT dosimetry, Monte Carlo simulations; computational phantoms; modeling**

I. Introduction

Monte Carlo simulations are commonly considered as reference (gold standard) for dose estimation in Computed Tomography (CT) imaging. The accuracy of input parameters related to the geometry of the imaging system and the data acquisition protocol has substantial impact on the accuracy of results [1]. The input parameters could be extracted from CT raw data, which are commonly saved in proprietary format not easily accessible to CT scientists or simply obtained from DICOM information derived from CT images. In this work, we examined the impact of input parameters on the accuracy of organ level doses from a routine abdominal helical CT scan of a physical phantom.

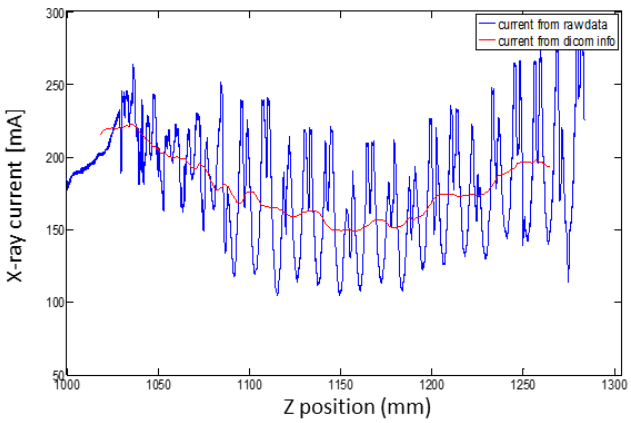

Figure 2. Tube current modulation in routine abdominal helical CT scan.

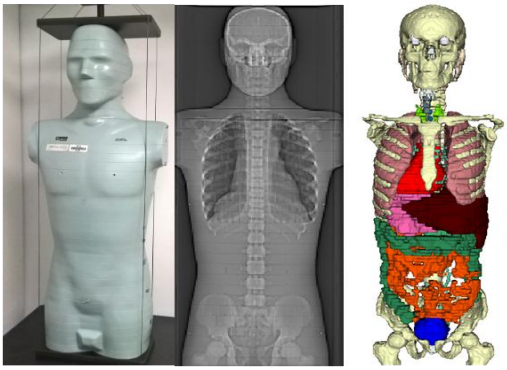

Figure 1. Physical and computational ATOM phantom.

II. Methods

A. Construction of the computational phantom

We used the ATOM adult anthropomorphic phantom (Computerized Imaging Reference Systems, Inc, Norfolk, VA) for a routine abdominal helical CT scan. The ICRP adult male computational phantom was employed as an anchor model to be registered to the ATOM phantom CT images using a 3-D deformable registration algorithm for the construction of a whole-body voxel model with well-defined anatomical structures. Image registration was performed using a previously developed algorithm based on the Insight Toolkit (ITK, https://itk.org) [2]. As shown in Figure 1, the registered voxel phantom was used as input for MCNPX-based Monte Carlo calculations of CT radiation dose.

B. Monte Carlo simulations

Computational models are commonly coupled with Monte Carlo simulations for dosimetry calculations through full simulation of the CT scanner and input parameters used by the scanning protocol. This study was acquired on the GE 750HD CT scanner (GE Healthcare, Waukesha, WI). The geometry of the system was accurately modeled and validated against experimental measurements as described in previous work [3]. The acquisition parameters, including the

revolution time, pitch factor, total collimation width, table speed, tube voltage and modulated tube current and tube start angle extracted from two sources of information: CT raw data files and image DICOM headers. Using CT raw data, we extracted the simulation input parameters for 2305 source point positions per rotation. Using DICOM information, we accessed the data for each slice keeping in mind that the accuracy of the input data depends on the slice thickness. The angular and longitudinal tube current modulation (TCM) profiles were obtained from CT raw data. Only the longitudinal TCM of each slice is accessible from the DICOM header file of CT images. Furthermore, the tube start angle, an important factor in simulations is only available from the CT raw data. Therefore, organ level absorbed doses and effective dose were calculated using a previously validated C++ computer code [3].

III. Results

The complete TCM obtained from CT raw data is illustrated in Figure 2 where the longitudinal TCM extracted from the DICOM information corresponding to the average movement of the complete TCM per rotation is mapped on it. In raw data-based simulation framework, 2305 point source positions per rotation with complete TCM was modeled where the tube start angle was extracted from raw data. In the image-based framework, the number of simulation point positions was determined based on the slice thickness where the tube start angle was modeled in four different arbitrary angles with 90° differences. The average of four different tube start angles is used for dosimetry calculations. The simulated absorbed dose per particle in one rotation is illustrated in Figure 3. A 4% error results from ignoring the complete TCM. The absolute mean difference between organ-level doses resulting from raw data-based and image-based simulations for one rotation in the abdominal region is about 6.92%, while this difference for the gallbladder exceeds 19%. Organ absorbed doses and effective dose are shown in Figure 4.

IV. Conclusion

We investigated the dosimetric impact of complete TCM and tube start angle modeling in MC simulation-based CT dosimetry. The methodology can be further expanded to produce an accurate MC simulation with a reduced computational burden.

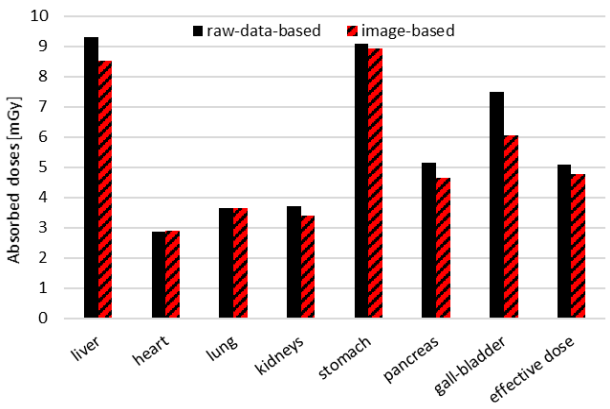

Figure 4. Comparison of absorbed organ doses and effective dose derived from raw data- and image-based simulations.

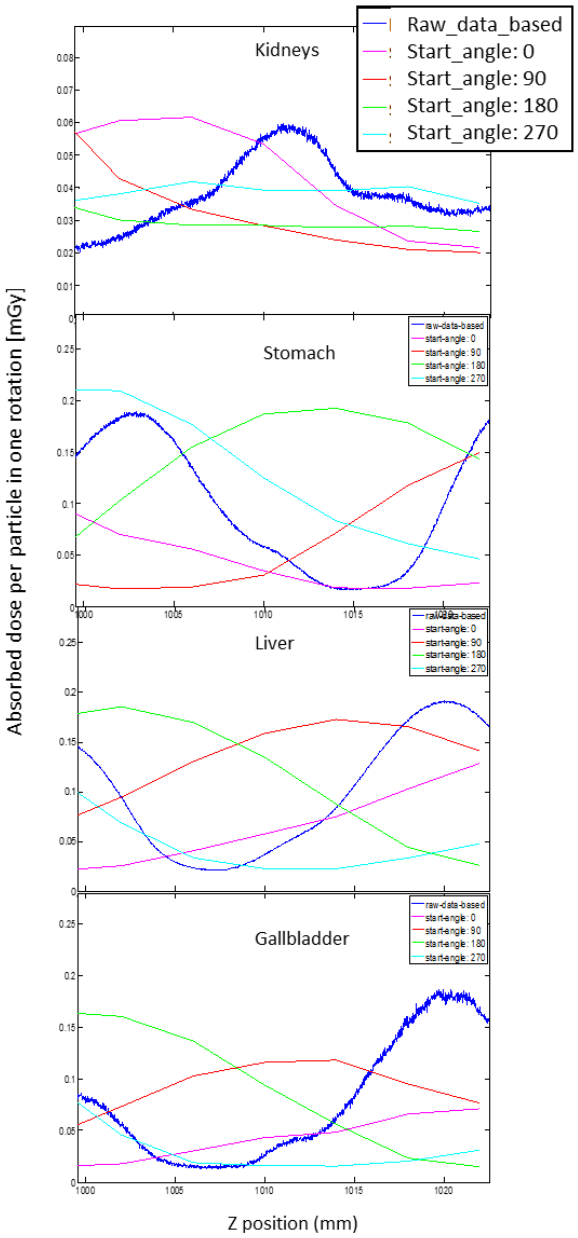

Figure 3. Simulated dose per particle in one rotation for the kidneys, stomach, liver, and gallbladder.

Acknowledgment

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References

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