A Hybrid Approach for Accurate Estimation of the Scatter Component in X-ray CT Combining Experimental Measurements and Monte Carlo Simulations

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Abstract– **The corruption of projection data in x-ray CT with scattered radiation decreases low contrast detectability (LCD) in reconstructed images. Scatter removal is therefore mandatory in CT imaging because of the need to generate images presenting clinically acceptable LCD. The assessment of the scatter component in CT is an active research area in quantitative imaging, however very few contributions have been made in this domain. Thorough knowledge on the scatter distribution is essential for development of scatter correction algorithms. Conventional experimental measurement (EM) based methods using one lead bar as a blocker are not able to extract scattered profiles in a single exposure. A novel method previously proposed by our group using an array of blockers seems to be a reasonable way to efficiently measure scatter profiles. However, the lead blocker is a secondary source of scattered radiation which propagates substantial errors in the EM procedure. Monte Carlo (MC) simulation is the only way enabling to estimate and correct the errors arising from EM. In this study, a hybrid approach combining EM and MC simulations is proposed to accurately calculate scatter profiles in x-ray CT scanners. The hybrid method involves two steps: (a) EM using lead arrays to calculate the scattered radiation profile and (b) MC simulations to estimate and correct scattered radiation induced from lead arrays. The MC simulator was validated through comparison with EM. MC simulations of a 64-slice CT scanner geometry showed that the magnitude of scattered radiation arising from the lead array used for measurement of scatter profile is 54% of total scattered radiation, which can not be ignored. The measured scatter profiles in different tube voltages were corrected using the suggested hybrid approach. In conclusion, the hybrid method seems to be a precise and robust method for accurate estimation of scattered radiation profile. The method is still undergoing some**

improvements to reduce the computational time and to optimize the combination of MC and measured data.

I. INTRODUCTION

The corruption of projection data in x-ray CT with scattered radiation decreases low contrast detectability, reduces CT numbers and introduces cupping artifacts in reconstructed images. Scatter removal is therefore mandatory in x-ray CT imaging because of the need to produce clinical images with acceptable low contrast detectability. Therefore, correcting these artifacts was targeted since the earliest designs of CT scanners, either by optimizing the design geometry or by using improved image reconstruction algorithms. One of the most important source of artifacts is scattered radiation which depends upon the system geometry, object under study and exposure conditions [1, 2].

Contamination of CT data with scattered radiation causes cupping artifacts in reconstructed images [3]. Since exposing a large volume of the patient will produce more scattered radiation, these artifacts are much more significant in cone-beam configurations of CT scanners equipped with two-dimensional detector arrays. In order to develop vigorous scatter correction algorithms, it is essential to be acquainted with the magnitude and spatial distribution of scattered radiation. In the context of x-ray CT imaging, different groups have proposed methods for assessment of the scatter component for both fan- and cone-beam configurations, including experimental measurements, mathematical modeling and Monte Carlo simulations. However, most published papers investigating the distribution of scattered radiation in the fan-beam geometry used either simple experimental measurements based on the use of a single blocker [3, 4, 5] or comprehensive Monte Carlo simulations [5, 6, 7]. Conventional experimental measurement methods based on one lead bar as a blocker of primary photons are able to extract scattered profiles in merely a very small region of the detector array [1]. A novel method was recently proposed by our group that uses an array of blockers is a reasonable way to efficiently measure scatter profiles [8]. Experimental measurement of the scatter profile utilizing lead as a blocker of primary photons produces secondary scattered radiation which propagates remarkable errors in experimental measurements. Scattered photons originating from the lead blocker are attenuated through the phantom and finally reach the detector array and contaminate the scatter profile. It is possible to extract the scattered

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radiation profile induced from lead blockers using experimental methods; however, it is impossible to appropriately model attenuation of these photons within the phantom. The only accurate way to predict phantom attenuation is Monte Carlo simulations. In this study, a hybrid approach combining experimental measurements and Monte Carlo simulations is proposed to efficiently calculate scatter profiles in x-ray CT scanners. The hybrid method involves two steps: (i) experimental measurements using lead arrays in order to calculate the scattered radiation profile [8] and (ii) Monte Carlo simulation to estimate and correct scattered radiation induced from lead arrays. The method can in principle be used for any scanner geometry and configuration. In contrast to previously published methods for measuring scattered radiation (limited to one measurement point), the proposed method measures scattered radiation in almost all detector channels.

II. MATERIALS AND METHODS

The hybrid method was used for the calculation of scatter profiles in the GE LightSpeed VCT 64-slice cardiac CT scanner (GE Healthcare Technologies, Waukesha, WI). This third generation CT scanner has 540 mm source to iso-center and 950 mm source to detector distance, 58,368 individual detector elements arranged in 64 rows of 0.625 mm thickness at iso-center, each containing 888 active patient elements and 24 reference elements with Highlight (Y2Gd2O3:Eu) ceramic scintillator. In this method, measurement of the scatter profile was carried out through the use of an array of lead bars which was inserted under the collimator and the water phantom in the iso-center. The raw data file which contained detector array readout, was transferred to a PC and read using a dedicated graphical user interface running under MatLab 7.5 (The Math Works Inc., Natick, MA, USA). The 64×912 matrix reflecting detector readings was divided into two areas: (i) shadowed area whose detector channels were exactly within the shadow of lead bars in the blocker array and (ii) exposed area whose detector channels were exposed and recorded both scatter and primary (i.e. total) radiation. It was assumed that matrix elements assigned to the shadowed area contain essentially scattered radiation.
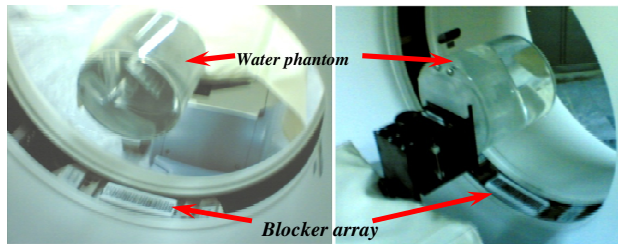

Fig. 1. Experimental setup for measurement of scatter profiles in the GE VCT 64-slice cardiac CT scanner. The array of lead blocker was put on the top of collimator and below the water phantom.

The Monte Carlo simulation was based on the MCNP4C general-purpose Monte Carlo code, which is a continuous-energy, generalized-geometry, time-dependent, coupled neutron/photon/electron Monte Carlo transport code. For photon transport, the code considers photoelectric absorption, with the possibility of K- and L-shell fluorescent emission or Auger electron, coherent and incoherent scattering and pair production all together. In the first step, the user is supposed to provide an input file which includes information about the problem such as geometry specification, materials characteristics, and type of answer or tally as well as any variance reduction techniques to be used. The geometry of MCNP treats an arbitrary three-dimensional configuration in Cartesian coordinate system. The total number of cells in this code should be less than 10^5. This limit might be exceeded in some geometries. The lattice geometry should be used under this condition. The lattice uses a smaller number of cells but prolongs the MCNP execution time. MCNP takes the advantage of a powerful source specification card referred as surface source card which is able to separate primary photons from scattered photons entering into a particular cell or surface. Other important standard features which can be considered as strong points of MCNP that make it very flexible and user friendly can be counted as powerful general source and surface source, both geometry and output tally plotters, a well provided collection of variance reduction techniques, a multi-purpose tally structure and an extensive collection of cross-section data. In order to reduce computation time, MCNP takes advantage of parallel computer architectures. It is supported in multitasking mode on some mainframes and in multiprocessing mode on a cluster of workstations where distributed processing uses the Parallel Virtual Machine (PVM) software.

The creation of input files for the MCNP4C code using first and second degree surfaces is a complicated and time consuming task, especially when the user deals with the complex geometries typical of an x-ray CT scanner. As a matter of fact, the description of 64×912 detector elements (small cubes) requires an extremely large number of plans to create detector cells. A user friendly GUI running under MatLab 7.5 (The Math Works Inc., Natick, MA, USA) was developed in order to generate the MCNP input file [5]. The user is prompted to enter the x-ray CT scanner design parameters and the GUI creates the scanner geometry according to the information provided and options selected as MCNP4C input file. This user friendly GUI is capable of providing the x-ray spectrum for different kVp and filtrations, phantom properties and tally specifications. The GUI is also able to save and reload known characteristics of standard commercial scanners. The MC simulator was validated through comparison with experimental measurements [5].
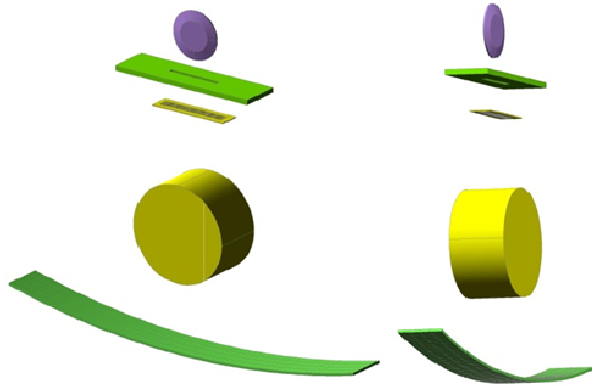
Fig. 2. Two schematic views of the geometry used for both experimental measurements and Monte Carlo simulations.

The validity of MCNP4C-based Monte Carlo simulations was verified by analogy between the simulated and measured distributions obtained using a cylindrical water phantom. The normalized error (NE) calculated for each detector element was used as a figure of merit to evaluate differences between measured and simulated projections using the following formulation [6]:

$$NE(u,v) = \frac{P_{Measured}(u,v) - P_{Simulation}(u,v)}{P_{Measured}(u,v)} \quad (1)$$

where u and v are detector elements' coordinates and $P_{Measured}$ and $P_{Simulation}$ are the measured and simulated projection data for each detector element normalized to the blank scan. To reduce statistical fluctuations and quantum noise, the measured projection data were averaged in 1000 views given the symmetry of the phantom. The input file containing detailed information about the geometry and source characteristics of the CT scanner along with all other components such as blocker array and phantom was created. Moreover, a surface source was applied after the blocker in order to separate the scattered and primary photons leaving the blocker array. Through the next stage, the scattered photons induced from the blocker array were transported through the phantom and then the total energy released in the detector was scored by the F6 tally. The simulation was carried out again after removing the phantom allowing to calculate attenuation factors by dividing the scattered distribution computed during the first stage by that obtained during the second stage. In all cases, $\sim 1.9 \times 10^9$ photons were simulated from the focal spot. Increasing the number of photon histories will decrease the statistical fluctuations and quantum noise in the simulated profiles and thus decrease the standards deviations of the estimates. Although 1.9×10^9 photons were generated from the focal spot, around 5×10^6 photons scattered in the blocker. Hence statistical fluctuations were unavoidable unless some form of denoising or curve fitting is used.

III.   RESULTS

Fig. 3 compares attenuation profiles obtained from experimental measurements and Monte Carlo simulations for the uniform water-filled cylindrical phantom. Fig. 4 shows the normalized error between measured and simulated profiles calculated using Eq. (1). The maximum relative difference between experimental and simulated results (in the region of phantom shadow) was less than 5%.
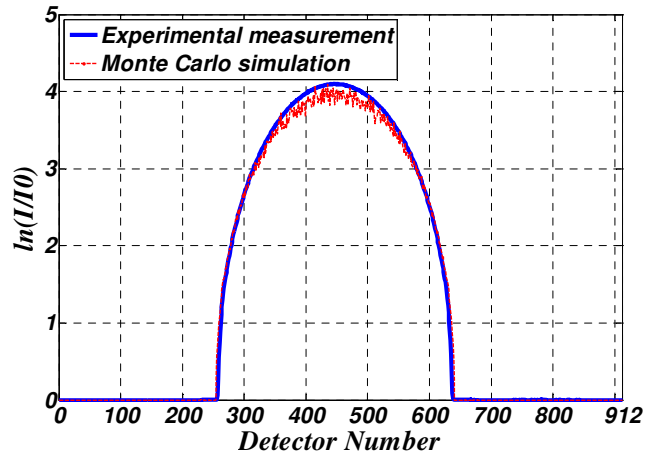

Fig. 3. Comparison of attenuation profiles for a uniform cylindrical water phantom using experimental measurements (solid blue line) and MC simulations (dashed red line).
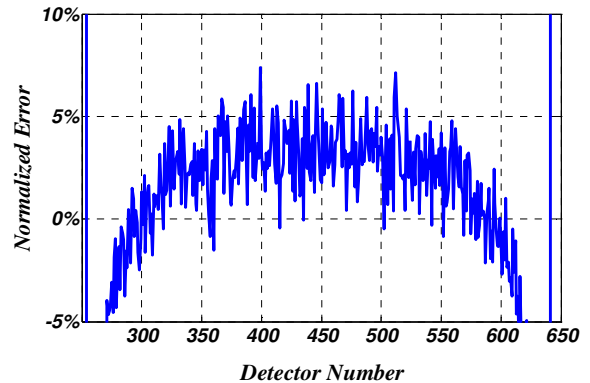

Fig. 4. Normalized error (%) between measured and MC simulated profiles shown in Fig. 3.

Fig. 5 illustrates MC simulated scattered radiation induced from the added blocker. It can be seen that the contribution of scattered radiation from the blocker array is substantial (scattered radiation from the blocker is ~54% of total scattered radiation). Fig. 6 compares corrected versus non-corrected measured profiles obtained using the hybrid method for a tube voltage of 140 kVp.
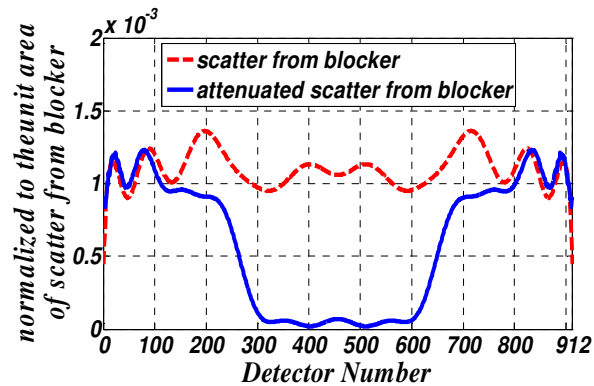
Fig. 5. MC simulated scatter component originating from lead array blocker with and without cylindrical water phantom in the field of view.
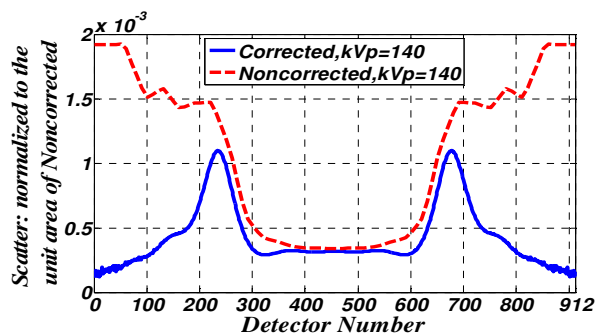

Fig. 6. Plots of corrected versus non-corrected measured profile using the hybrid approach at 140 kVp.

IV. DISCUSSION

Scatter extraction using our recently developed experimental technique is on the one hand a straightforward approach to measure the scatter distribution [8]. On the other hand, in this method the blocker adds remarkable amount of scattered photons to the measurements which renders the accuracy the techniques questionable. As mentioned above, the motivation behind the use of Monte Carlo simulations is that simulation is the mere solution to compute the attenuation caused by the phantom for scattered photons originating from the blocker array. The validation of our Monte Carlo calculations indicates a good agreement between experimental measurements performed on the 64-slice CT scanner and Monte Carlo simulations (see Figs. 3 and 4). As slight differences between the simulated and measured results (under 5%) are due to the fact that in our simulations we were lacking thorough knowledge of the material comprising the phantom wall and considered it to be Plexiglas. We believe that the material used for the phantom wall is such that its attenuation coefficients are slightly higher than that of standard Plexiglas used by manufacturers. Moreover, although we obtained detailed information from the scanner vendor about CT scanner design parameters, any small difference between the

simulated geometry and materials and actual design parameters can be a source of error. The normalized error is high near the phantom wall as shown in Fig. 4. This overshoot in error is the result of mismatched positions of the phantom between experimental measurements and Monte Carlo simulations. Fig. 5 shows the net scatter radiation from the array of lead bars. The contribution of scatter in the phantom shadow is lower than that in the non-shadowed area and this is a reasonable evidence of the attenuating role of the phantom for scattered photons. Fig. 6 illustrates the ultimate correction of scatter distribution obtained from experimental measurements. As a matter of fact, the impact of correction is negligible within the phantom shadow whereas it becomes evident outside this area.

V. CONCLUSION

As conclusion, the proposed hybrid approach seems to be a precise and robust method for accurate estimation of scattered radiation distribution in x-ray CT. The method is still undergoing some improvements to reduce the computation time and optimize the strategy for combining Monte Carlo and measured results. Moreover, the presented approach might be used for scatter correction in CT scanners with 64-slice capability or more. The method is also applicable to other geometries of multi-slice CT scanners. In addition, accurate estimation of the magnitude and spatial distribution of scattered radiation using the proposed method can be used for optimization of scanner design geometry and development and optimization of robust scatter correction techniques which is certainly a curtail for new generation CT scanners with 256 and 320 slice capability [9].

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