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An Improved Method of Heterogeneity Compensation for the Convolution / Superposition Algorithm

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Abstract.

Purpose: To improve the accuracy of convolution/superposition (C/S) in heterogeneous material by developing a new algorithm: heterogeneity compensated superposition (HCS).

Methods: C/S has proven to be a good estimator of the dose deposited in a homogeneous volume. However, near heterogeneities electron disequilibrium occurs, leading to the faster fall-off and re-buildup of dose. We propose to filter the actual patient density in a position and direction sensitive manner, allowing the dose deposited near interfaces to be increased or decreased relative to C/S. We implemented the effective density function as a multivariate first-order recursive filter and incorporated it into GPU-accelerated, multi-energetic C/S implementation. We compared HCS against C/S using the ICCR 2000 Monte-Carlo accuracy benchmark, 23 similar accuracy benchmarks and 5 patient cases.

Results: Multi-energetic HCS increased the dosimetric accuracy for the vast majority of voxels; in many cases near Monte-Carlo results were achieved. We defined the per-voxel error, %/mm, as the minimum of the distance to agreement in mm and the dosimetric percentage error relative to the maximum MC dose. HCS improved the average mean error by 0.79 %/mm for the patient volumes; reducing the average mean error from 1.93 %/mm to 1.14 %/mm. Very low densities (i.e. <0.1 g / cm³) remained problematic, but may be solvable with a better filter function.

Conclusions: HCS improved upon C/S’s density scaled heterogeneity correction with a position and direction sensitive density filter. This method significantly improved the accuracy of the GPU based algorithm reaching the accuracy levels of Monte Carlo based methods with performance in a few tenths of seconds per beam.

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1. Overview
It is challenging for the convolution/superposition (C/S) algorithm to maintain accuracy in regions of electronic disequilibrium. Electronic disequilibrium occurs in areas where there are gradients in both TERMA and density. This leads to significant errors near material interfaces for small or highly modulated fields. Empirically, these errors appear as an overestimation of the dose after a high-density to low-density transition and an underestimation of the dose after a low-density to high-density transition (Figure 1).

We propose to improve C/S by filtering the density ($\rho$) used by the superposition component of the calculation in a position and direction sensitive manner. The use of an effective density ($\rho_{eff}$) value allows the dose deposited to be empirically increased or decreased, relative to traditional C/S. This change in the deposited dose occurs due to the change in the integration width of the superposition kernel across the voxel; the superposition kernels themselves are not modified. Therefore, the total integral of the superposition kernel never changes and both heterogeneity compensated superposition (HCS) and C/S compute similar dose distributions in regions of electronic equilibrium.

2. Materials and Methods
HCS was implemented as an extension to our previous work in GPU accelerated convolution/superposition (C/S).[1,2] Previously, we leveraged the hardware functionality of the GPU texture unit to allow our C/S implementation to efficiently perform multiple superpositions simultaneously. By computing the superposition for each single energy bin independently (multi-energetic) and summing the resulting doses, the kernel is effectively hardened with depth as the lower energy bins will have less contribution to the TERMA at depth. For efficiency, we experimented with simply dividing the spectrum into 2 (dual-energetic) and 4 (quad-energetic) energy sub-spectra and performing the superposition independently for each to more efficiently handle the kernel hardening. These divisions were applied after the TERMA was computed using the exact, full-spectrum attenuation. Use of multi-energetic transport substantially increased the accuracy of C/S in homogeneous phantoms when the source model used the true Monte-Carlo spectrum as opposed to a manually commissioned spectrum.

In order to further improve the accuracy near density changes, we formulated the $\rho_{eff}$ function at position $t$ as a first-order exponential linear recursive filter with a varying time constant as it can be implemented using a small, fixed number of GPU registers and can handle the irregular signal-sampling rate produced by our incremental ray-tracing algorithm. The time constant, $\tau$, is dependent on the voxel width, $\Delta$, an adaptive $\alpha$ term and a decaying $\beta$ term.

Figure 1. (top) Depth dose for 2 commercial C/S methods vs Monte Carlo for the ICCR benchmark showing the error in dose near heterogeneities: (bottom) GPU accelerated superposition with normal density scaling and HCS showing accuracy improvements.
Equation 1 shows the traditional form of the convolution function using density scaling of the water kernels. Equation 2 shows how it has been adapted to accommodate the $\rho_{\text{eff}}$ function defined in Equations 3-8. The coefficients were optimized to best fit the low and high dose gradient regions of the 18MV Water-Lung-Bone-Water and the 24MV Water-Bone-Lung-Water benchmarks. Table 1 indicates the optimized values of the coefficients used in the $\rho_{\text{eff}}$ function.

$$D_{\text{CS}}(r) \equiv \int \int T(r + iv)K_d\left(\int^\infty_{r'}\rho(r')dr'\right)dt$$  \hspace{1cm} (1)$$

$$D_{\text{HCS}}(r) \equiv \int \int T(r + iv)K_d\left(\int^\infty_{r'}\rho_{\text{eff}}(r,v,r')dr'\right)dt$$  \hspace{1cm} (2)$$

$$\rho_{\text{eff},i+1} (r') = (1 - \tau_i)\rho_i + \tau_i\rho_{\text{eff},i}$$  \hspace{1cm} (3)$$

$$\tau_i = \max\left\{0, 1 - \Delta\left(\alpha_i(1 + \beta_i)\frac{\rho_{\text{eff},i}}{\rho_{\text{eff},i} + \rho_i}\right)\right\}$$  \hspace{1cm} (4)$$

$$\alpha(\rho) = \left(C_{\alpha,0} + C_{\alpha,1}\rho - C_{\alpha,2}\rho^2\right)$$  \hspace{1cm} (5)$$

$$\beta(\rho) = \left(C_{\beta,0} + C_{\beta,1}\rho - C_{\beta,2}\rho^2\right)$$  \hspace{1cm} (6)$$

$$\beta_{i+1} = \max\left\{0, \beta_i - \Delta\left(C_{\beta,0} + C_{\beta,1}\beta_i + C_{\beta,2}\beta_i^2\frac{\rho_{\text{eff},i}}{\rho_i + \rho_{\text{eff},i}}\right)\right\}$$  \hspace{1cm} (7)$$

$$\tau_{i+1} = \max\left\{0, 1 - \Delta\left(C_{\tau,0} + C_{\tau,1}\alpha_{i+1} + C_{\tau,2}\frac{\rho_{\text{eff},i}}{\rho_i + \rho_{\text{eff},i}}\right)\right\}$$  \hspace{1cm} (8)$$

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</table>

Table 1: Optimized coefficients used to calculate effective density.

Figure 2 shows the effects of the filter on $\rho_{\text{eff}}$ and the resulting effective distance used in the kernel lookup during the superposition ray-tracing. When entering a high density region from a lower density, the filter causes the $\rho_{\text{eff}}$ to gradually increase, and similarly when going from a high to low density, it will smoothly decrease. This alters the kernel lookup from the more traditional straight density scaling.

3. Results

Accuracy of the method was tested using both the ICCR benchmark Monte Carlo phantoms [3] and human CT scans. Comparisons were made for the identical spectra and source model for the HCS and EGS3 Monte Carlo dose calculation algorithms.

3.1. Slab Phantom Benchmark Results

Figure 3 shows the results from selected slab phantoms for the dual and quad energetic HCS methods. The improvement in the model by breaking it up into the multiple energy groups is...
evident in the upper left plot where the traditional single spectrum poly-energetic HCS is displayed. Agreement with Monte Carlo is within 1% for the majority of the profiles in the various field sizes and phantoms displayed. Some error remained in the drop off from bone to lung in the 4MV results using the 1.5 cm field size.

Figure 3. Phantom results comparing HCS to Monte Carlo for 18, 4 and 24MV for both Dual (green) and Quad (red) Energetic modes. Upper left: the single poly-energetic (orange) shows the impact of not breaking up the spectrum. Top row shows field sizes of 1.5, 3 and 5 cm for an 18 MV beam on the water-bone-lung-water phantom. Bottom 2 rows shows 4 and 24 MV on the water-bone-lung-water, water-half_bone-water, and water-half_lung-water phantoms for a 1.5 cm field size.

3.2. Patient Benchmark Results
Figure 4 shows patient accuracy benchmarks comparing C/S, HCS indicating the improvement in accuracy between the C/S and HCS methods in the regions of electronic disequilibrium due to heterogeneities. All fields were 18MV and 5 cm x 5 cm at the volume’s surface. HCS improved the average mean error by 0.79 %/mm for the patient volumes; reducing the average mean error from 1.93 %/mm to 1.14 %/mm.
3.3. Discussion
HCS was compared to traditional C/S using 29 slab phantom benchmarks, 5 patient cases and 1 water phantom (not all presented here). In many cases the accuracy of HCS was comparable to Monte-Carlo methods. However, discrepancies were observed between HCS and Monte Carlo in regions of low density ($\rho \leq 0.1 \text{ g/cm}^3$). This was due in part to the geometric scale of electronic disequilibrium being the largest at low densities and in part due to our formula for the HCS parameters not robustly extrapolating to very low densities. The build-up regions of the patient datasets were also systematically overestimated. This was due to a layer of low density voxels surrounding the patient that caused the build-up region to be treated as a re-build-up region. We intend to introduce a depth dependence to the HCS filter function to correct for this issue.

4. Conclusions
We have developed a novel improvement to the C/S that better accounts for the electronic disequilibrium caused by patient heterogeneity. This work shows that the convolution/superposition algorithm can rival Monte Carlo in accuracy when the traditional approximations are removed from the C/S algorithm and improved models for handling kernel hardening and kernel scaling with density have been applied.

5. References

Figure 4. Comparison of C/S and HCS methods to Monte Carlo on patient CT scans showing increased accuracy of HCS in the regions near heterogeneities for an 18MV beam and a 5x5 cm field.