Published in final edited form as:

Med Phys. 2007 April ; 34(4): 1439–1449.

Effects of Hounsfield number conversion on CT based proton Monte Carlo dose calculations

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Abstract

The Monte Carlo method provides the most accurate dose calculations on a patient computed tomography (CT) geometry. The increase in accuracy is, at least in part, due to the fact that instead of treating human tissues as water of various densities as in analytical algorithms, the Monte Carlo method allows human tissues to be characterized by elemental composition and mass density, and hence allows the accurate consideration of all relevant electromagnetic and nuclear interactions. On the other hand, the algorithm to convert CT Hounsfield numbers to tissue materials for Monte Carlo dose calculation introduces uncertainties. There is not a simple one to one correspondence between Hounsfield numbers and tissue materials. To investigate the effects of Hounsfield number conversion for proton Monte Carlo dose calculations, clinical proton treatment plans were simulated using the Geant4 Monte Carlo code. Three Hounsfield number to material conversion methods were studied. The results were compared in forms of dose volume histograms of gross tumor volume and clinical target volume. The differences found are generally small but can be dosimetrically significant. Further, different methods may cause deviations in the predicted proton beam range in particular for deep proton fields. Typically, slight discrepancies in mass density assignments play only a minor role in the target region, whereas more significant effects are caused by different assignments in elemental compositions. In the presence of large tissue inhomogeneities, for head and neck treatments, treatment planning decisions could be affected by these differences because of deviations in the predicted tumor coverage. Outside the target area, differences in elemental composition and mass density assignments both may play a role. This can lead to pronounced effects for organs at risk, in particular in the spread-out Bragg peak penumbra or distal regions. In addition, the significance of the elemental composition effect (dose to water vs. dose to tissue) is tissue-type dependent and is also affected by nuclear reactions.

Keywords

Geant4; Monte Carlo; proton therapy; CT Hounsfield conversion

I. INTRODUCTION

Monte Carlo methods provide the most accurate radiation dose calculations. Such simulations can take into account tissue inhomogeneities by using detailed material properties (e.g., material composition, electron density, mass density, and ionization potential). Because of the usually long computational time associated with Monte Carlo dose calculations, it is still not practical to carry out Monte Carlo based radiotherapy treatment planning routinely. However, in recent years, Monte Carlo methods have been widely applied to verify the results of the approximate dose calculation algorithms implemented in treatment planning programs.¹⁻⁶

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Most dose calculation algorithms, like the ones implemented in commercial treatment planning programs, import the patient specific anatomic information in the form of digital imaging and communications in medicine (DICOM) formatted computed tomography (CT) datasets, where the patient is represented by a large number of small rectangular volumes (voxels) with CT information characterized by Hounsfield numbers. The common practice in photon therapy is to use the electron density⁷ or mass density⁸ to characterize the radiation energy loss in tissues. A CT scanner can be easily calibrated, to convert CT Hounsfield numbers into electron density, with the use of phantom materials of well-known electron densities. This CT calibration method is valid for photon or electron beams because the dominant energy loss process is interaction with electrons (Compton scattering for x rays,⁷ and ionization for charged particles⁸). With this measure, dose calculation algorithms are fast but influences of tissue elemental compositions to radiation dose distributions are not fully accounted for.

A conversion from Hounsfield numbers to electron density is insufficient when it comes to proton radiation. Protons lose energy by ionizations, multiple Coulomb scattering and nonelastic nuclear reactions. Instead of electron density or mass density, relative stopping power can be used to accurately define water equivalent tissue properties. Relative stopping power depends on electron density and a Z-dependent kinetic term. At the Francis H. Burr Proton Therapy Center (Former Northeast Proton Therapy Center), Massachusetts General Hospital (MGH) treatment planning is done using FOCUS/XiO (Computerized Medical Systems, Inc.). This package is capable of handling photon and proton treatments and thus has to incorporate two different methods of CT conversion. For every CT voxel a lookup table is used to convert Hounsfield numbers into electron density (for photon calculations) and into relative stopping power (for proton calculations). The proton dose calculation implemented in the treatment planning program is based on an in-house developed pencil beam algorithm.⁹

In contrast to analytical dose calculation algorithms, Monte Carlo dose calculations are typically not based on water equivalent properties but on the exact material composition. Thus, a reliable conversion from Hounsfield numbers to human tissues must be obtained. There is no simple one to one correspondence between CT Hounsfield numbers and tissue materials. The Hounsfield number only reflects the attenuation coefficient of human tissues to diagnostic x rays. Thus, Hounsfield number may be identical for a variety of combinations of elemental compositions, elemental weights and mass densities, not to mention that the measured Hounsfield number may depend on other factors such as the location of the tissue in the image (beam hardening effect). Applying a realistic Hounsfield number conversion is not only important in order to obtain reliable results when using Monte Carlo dose calculations, it is also important in order to understand potential differences when comparing a dose calculation engine implemented in a treatment planning program with Monte Carlo dose calculation.

The accuracy of Monte Carlo dose calculations is affected by the ability to precisely define materials based on the Hounsfield number information. Thus, in addition to electron or mass density, composing elements and their relative weights need to be known as well. The common way to determine the corresponding CT Hounsfield numbers to human tissues is to scan tissue phantom materials or animal tissues.¹⁰ Hounsfield numbers are then grouped according to their tissue types in order to assign elemental composition and weights [based on publications by the International Commission on Radiation units and Measurements (ICRU)¹¹ or International Commission on Radiological Protection (ICRP)¹²]. For any tissue, the measured Hounsfield number falls into an interval, where the intervals of different tissues can be overlapping. Tissue classification based on Hounsfield numbers for CT based Monte Carlo dose calculation was studied previously.^{1,3,6,13-15} In these studies, the number of materials adopted varies (up to 17).³ Also unlike the treatment planning programs in which a unique electron density or relative stopping power can be assigned to each Hounsfield number, in these

Monte Carlo studies, a single mass or electron density was used for the entire Hounsfield number group.

With potentially better critical organ sparing and less integral dose, radiotherapy with heavy charged particles has clear advantages over other modalities, like x rays and electrons. At MGH, clinical proton therapy with high-energy proton beams (up to 230 MeV) has been extensively practiced. The designated spread-out Bragg peak (SOBP) is achieved by passing the proton beams through rotating modulation wheels made of materials of different thicknesses in different sections. The proton beams are also appropriately shaped and intensity modulated with the aperture and compensator before impinging onto patients. We have clinically implemented Monte Carlo dose calculations as a quality assurance tool and are developing Monte Carlo based techniques to verify the accuracy of the pencil beam algorithm. The purpose of this work was to study the effects of three Hounsfield number conversion methods on Monte Carlo dose simulations of proton beams.

II. MATERIALS AND METHODS

A. Methods to convert Hounsfield number to material

A method suggested by Schneider, Bortfield and Schlegel,¹⁴ is based on a stoichiometric calibration of Hounsfield number (H) with mass density and elemental weights. Using experimentally determined parameters, Hounsfield numbers for 71 human tissues, whose characteristics were taken from literature, were calculated. Mass density and elemental weights of any Hounsfield number were obtained via linear interpolation. The Hounsfield numbers from -1000 to +1600 were binned into 24 groups by the authors: one group for air (-1000 to -950), one group for lung tissue (-949 to -120), seven groups for soft tissues (-119 to +120), and 15 groups for skeletal tissues (+121 to +1600). Within each group, the elemental composition and weights are preserved, whereas the mass density continuously increases with the Hounsfield number (except in the small range of 14–23 where a constant density of 1.03 g/cm³ is assigned). Although this method provides a robust division of soft tissues and skeletal tissues, relatively large discrepancies in elemental weights were reported, especially for soft tissues. The authors pointed out that this is because soft tissues in the Hounsfield number range between 0 and 100 can only be poorly distinguished by the means of Hounsfield numbers. Their data showed that more than 30 soft tissues with different elemental compositions are very close to each other in terms of Hounsfield number. We will refer to this method as the “Schneider method” throughout this paper.

A different conversion method is the one implemented in the BEAMnrc code.¹⁶ BEAMnrc includes a program called *ctcreate*, which is used to create the input file for the DOSXYZnrc¹⁵ user code for dose calculations. This program allows the user to specify a method for converting Hounsfield numbers to tissue materials, but also provides a default method. Four major types of tissues are adopted in the default method: air ($H \leq -950$), lung tissue ($-950 < H \leq -700$), soft tissue ($-700 < H \leq +125$) and bone ($H > +125$). The ICRU tissue compositions are used, and the mass density is obtained via linear interpolation of the values set at the boundaries of the Hounsfield number bins (0.001 g/cm³ at $H = -1000$, 0.044 g/cm³ at $H = -950$, 0.302 g/cm³ at $H = -700$, 1.101 g/cm³ at $H = 125$, and 2.088 g/cm³ at $H = 2000$). This method is very similar to others used in previous studies.^{1,3,6,13} This method is referred to as the “BEAM method” in this paper.

The third method we are considering is applied in FOCUS/XiO. In this method, no tissue composition is considered, and materials for Hounsfield numbers are specified as water of various densities. In FOCUS/XiO, the relationship between relative stopping power (relative to water) and Hounsfield number is established with four linear fits, and the five control points are 0 at $H = -1000$, 1.0 at $H = 0$, 1.04 at $H = 40$, 1.52 at $H = +1000$, and 2.02 at $H = +2000$. In this

method, no tissue composition is considered, and materials are specified as water of varying stopping power. The proton plans chosen in this study were designed with FOCUS/XiO, and hence the results with this method will serve as a reference. This method is referred to as the “Water equivalent method” in the following sections.

Figure 1 shows the curves of relative mass density versus H for the three conversion methods. The difference between the Schneider and Water equivalent methods is as much as 8% ($H = +1500$) in the range of skeletal tissues, and up to 4.7% (H around +100) for soft tissues. Between the BEAM and Water equivalent methods, for skeletal tissues ($H > +120$), the difference is up to 3% ($H = +1500$). For lung tissue, the Schneider method assigns up to 3.5% more mass density, whereas the BEAM method assigns up to 2% less, compared to the Water equivalent method. The difference between the Water equivalent method and either the Schneider or the BEAM method can be up to 10% in the air region ($H < -950$). However, due to its low density, the energy loss of protons in air is small, and hence the effect on dose distributions in tissues is insignificant. In the soft tissue region, the Schneider method assigns higher mass densities and the BEAM method adopts lower values than the Water equivalent method.

Figure 2 presents the weights of four major tissue elements (hydrogen, carbon, oxygen, and calcium) as a function of Hounsfield number for the three methods. The BEAM and Schneider methods use the same elemental weights for low-density air and lung tissues ($H < -120$). However, for soft and skeletal tissues, the difference in elemental weights is significant. In general, the Schneider method gives more weights to hydrogen and carbon, but less weights to oxygen and calcium. The largest difference appears at $H = -119$ for carbon and oxygen, where in the Schneider method, the elemental weight of carbon steps up from 0.105 (air) to 0.681 (adipose tissue), and that of oxygen steps down from 0.749 (air) to 0.198 (adipose tissue). The readings in the BEAM method are 0.111 and 0.762 for carbon and oxygen, respectively. For hydrogen and calcium, the greatest difference is located at $H = +126$, where in the BEAM method tissue type changes from soft tissue to bone.

B. Geant4 based Monte Carlo proton dose calculations

The Geant4 Monte Carlo code¹⁷ is based on object-oriented software design. Instead of being a stand alone executable, Geant4 is a developing toolkit of C++ class libraries, covering various needs for solving particle transport problems. As a general-purpose Monte Carlo code, some features in Geant4 are not fully optimized for radiotherapy dose calculations. For example, particle transport in CT voxels is unnecessarily slow. We have re-coded some parts of the Geant4 source code to improve its computational efficiency significantly.¹⁸ With a detailed model of the proton treatment nozzle¹⁹ and direct use of the same unsegmented CT datasets as in FOCUS/XiO to set up the patient geometry, any FOCUS/XiO proton treatment plan can be recalculated using Monte Carlo. This is currently done routinely at MGH. The Monte Carlo dose calculation acts on the original CT grid. No sampling on a bigger grid was done (as the planning system would do).

Within our Geant4 based user code, Hounsfield numbers are binned into groups according to their tissue types. Each type is characterized by elemental composition, mass density, and ionization potential. For the conversion, even though the number of different elemental compositions is small (same as the number of bins in the Hounsfield number conversion method), each Hounsfield number usually corresponds to a material with its own unique mass density. In defining a material, commonly used Monte Carlo codes such as MCNP, EGS and Geant4 all require the explicit specification of the elemental composition and mass density. Material properties can usually not be changed after the simulation initialization. Therefore, with direct use of the unsegmented CT datasets, one material has to be assigned to each Hounsfield number. Consequently, totally about 3000 ($-1000 \leq H < +2000$) materials have to be defined. This can cause problems in terms of computer memory or computational time because

it is time consuming to create the cross section, stopping power, and other associate tables for these materials. Instead of using the default method in Geant4 for material definition, we developed a new method that allows us to adjust the material density dynamically during particle transport.¹⁸ Thus, only one material needs to be defined (with a nominal mass density) for each Hounsfield number group.

C. Dose to water versus dose to tissue

Monte Carlo simulations that use water of various densities to represent human tissues report absorbed dose to water (analogous to most of the treatment planning programs), whereas simulations with the Schneider and BEAM conversion methods calculate absorbed dose to tissue directly. The discrepancy between dose to water and dose to tissue is tissue or Hounsfield number dependent. For photon beams, in soft tissues the difference is approximately 1%, while in cortical bone it exceeds 10%.²⁰ Similar effects exist for proton beams and accurate quantitative evaluation of this effect can be achieved only if dose to tissue can be converted to dose to water, or vice versa. This is more complicated for proton beams than for photon beams, due to the involvement of ionizations, multiple Coulomb scattering and nonelastic nuclear reactions with which protons can lose energy. Therefore, the proton energy-fluence varies strongly with the depth in the phantom or patient, unlike in the photon beam case where the primary electron energy fluence varies slowly with depth. In this study only a qualitative analysis based on mass stopping power ratios was done to address this issue. In the energy range of clinical proton beams, the dominant mechanism for protons to slow down is ionization. Without considering other energy loss mechanisms, the quotient of dose to water by dose to tissue is equated to the quotient of the mass stopping powers for water and tissue. Figure 3 presents this mass-stopping-power ratio, calculated using the stopping power values in Geant4, as a function of Hounsfield number ($-950 < H < 1500$) for the Schneider and BEAM methods at three proton energies (10, 100, and 200 MeV).

For Hounsfield numbers less than -120 (low density lung and soft tissues), the Schneider and BEAM methods use the same elemental composition, resulting in the same mass-stopping-power ratios for both methods. According to Fig. 3, at these energies, the mass stopping power for lung tissue can be up to 12% lower than that for water. In the soft tissue region, for the BEAM method, the ratio is well bound between 1.0 and 1.01; but for the Schneider method, a 2–3% lower value is observed for the adipose tissue (H close to -120). For cortical bone ($H > +1500$), the absorbed dose to bone is over 10% (10–14%) lower than the absorbed dose to water for both methods. For soft bone, with the Schneider conversion the mass-stopping-power ratio is 1.01 to 1.02. Because in the BEAM method the same elemental composition is used for all bony tissues, the mass-stopping-power ratio is about 1.1 for soft bone.

D. Case study

Two proton plans designed with FOCUS/XiO for patients treated at MGH were chosen for this study: one for lung tumor and one for paranasal sinus (PNS) malignancy. These plans were selected for their tissue variety. The lung tumor plan deals almost exclusively with soft and low-density lung tissues, whereas large air cavities and bony structures are involved in the PNS plan. The lung tumor plan is composed of three fields: left posterior (LP), right posterior (RP), and left anterior (LA). The PNS plan is composed of one anterior-posterior (AP), three left posterior (LP), and three right posterior (RP) fields, with paranasal sinus and lymph nodes as the treatment sites. Although only two proton plans and three treatment sites were studied, the major types of tissues (air, lung tissue, soft tissues, and bone) were all covered.

The Monte Carlo dose calculation for each field in the proton plans was done in two steps. In the first step, tracking protons through the treatment nozzle was proceeded. Variable nozzle components were appropriately positioned according to the treatment plan.¹⁹ The simulations

generated phase space files recording the spatial, energy and angular information of the particles (mainly protons and neutrons) at the exit of the treatment nozzle (downstream surface of the range compensator). In the second step, the patient geometry was constructed, and dose calculations were conducted with source particles sampled from the phase space files generated previously. The same phase space files were used in patient dose simulations with different Hounsfield number to material conversion methods. All relevant proton interactions with matter were included. At the energies of clinical proton beams (up to 230 MeV), nonelastic nuclear reactions can play a significant role,^{21,22} but ionization is the dominant process for protons to deposit energy in human tissues.

To ensure that the resultant differences in dose distributions calculated with the three conversion methods were statistically reliable, a sufficient number of source protons had to be tracked. The criterion to judge the sufficiency of the number of histories was based on the dose volume histograms (DVHs). Simulation results were deemed to be statistically acceptable if no further changes in integral DVH curves for the gross tumor volume (GTV) and clinical target volume (CTV) were observable after increasing the number of tracked source protons. Because GTV and CTV are normally located in high dose regions, such a criterion does not necessarily guarantee sufficient statistical resolution in low dose regions. However, while the statistical uncertainty may have a significant influence on the DVH of the target volume, its impact on the DVHs of organs at risk is only minor because the significance of the statistical effect depends on the steepness of the DVH.²³

As illustrated in Figs. 1 and 2, the conversion methods differ in both assignment of mass density and elemental composition. Therefore, any discrepancy in the final dose distribution arises from the combined effect of both. Consequently, this work focuses on

Study 1: Comparison of the Schneider and BEAM methods with respect to the dosimetric significance of choosing a particular method to convert Hounsfield numbers into mass density and elemental composition.

Study 2:

- (a) Comparison of all three conversion methods considering their differences in mass density but setting all materials to water. This study aims at differences due to elemental composition alone.
- (b) Comparison of the Schneider and BEAM methods considering the respective elemental composition and weights of each tissue material but adjusting the mass density to match the one used in the Water equivalent method. This study aims at differences due to mass density alone.

III. RESULTS AND DISCUSSION

A. Study 1: Dosimetric significance of differences in Hounsfield number conversion methods

Figure 4 illustrates the composite dose distributions in the CT iso-center slice for the lung tumor plan. The result based on the Water equivalent method is shown as a reference. The dose difference images, presenting the percent dose differences relative to the reference, are shown in Fig. 5 (first row, only the iso-center slice is shown). The analysis focuses on the target area (GTV and CTV). When comparing the two methods to the reference, the BEAM method shows larger discrepancy in the bone tissue (up to about 10%). Other than that, the dose distributions obtained using the Schneider and BEAM methods agree reasonably well in the entire CTV (the dose difference is within 5% for 97% of the CTV voxels for both methods, and the greatest difference is within 8%). The agreement is reasonable given the fact that the reference method reports dose to water instead of dose to tissue. The respective DVHs for the GTV and CTV are

also shown in Fig. 5 (second row). No discernible difference in the GTV DVHs is observed. For the CTV, the DVH curve with the BEAM conversion presents an extended dose falloff.

Figure 6 presents the iso-center slice dose distributions for the treatment site of paranasal sinuses of the PNS plan. The PNS fields have smaller field sizes, causing a larger spread of low dose deposition from scattered particles, compared to the fields in the lung tumor plan. The irradiated volume encompassing the paranasal sinuses contains large air cavities. The dose difference images in Fig. 7 (first row, only the iso-center slice is shown) reveal that the absorbed dose in air is significantly lower than in its surrounding soft or bony tissues in the target areas for the Schneider and BEAM methods. This is less pronounced with the Water equivalent method. In the paranasal sinus CTV, 4% and 9% of the nonair voxels (for the Schneider and BEAM conversion, respectively) show a larger than 5% dose difference compared to the reference method. The discrepancies in the DVHs for the PNS plan are pronounced (second row of Fig. 7). The CTV DVHs with the Schneider and BEAM methods are less smooth in the dose range around 50–60 Gy. This is due to the inclusion of air cavities in the contoured CTV volume and is eliminated if the DVHs are generated with the air regions in the CTV being excluded. Thus, the pencil-beam algorithm may have difficulties to accurately predict dose to air, which may cause misinterpretations of DVHs if air is included in a contour. The GTV DVH with the BEAM conversion shows appreciable deviation from those with the other two methods, indicating larger dose nonuniformity in the volume.

For both cases, the lung and paranasal sinus GTVs and CTVs, the Schneider conversion results in DVHs closer to the Water equivalent method compared with the BEAM method. Apparent is the difference between Schneider and BEAM for the paranasal sinus GTV. This shows that treatment planning decisions could be affected by the conversion method. The 95% coverage is achieved at 63.9 and 62.6 Gy for the Schneider and BEAM method, respectively.

The iso-center dose distributions, dose differences (only the iso-center slice is shown), and DVHs for lymph nodes of the PNS plan are shown in Figs. 8 and 9. In the lymph node CTVs, the BEAM method again shows larger difference in the bone tissue. Voxel by voxel comparison with the Water equivalent method indicated for both methods that over 10% of voxels show a larger than 5% dose difference. The DVHs obtained with the Schneider conversion for the left neck and right neck CTVs (for lymph nodes irradiation) present significant deviations from the reference DVHs, which is unlikely completely due to the fact that the Water equivalent method reports dose to water. Further study of dose difference images in the entire CTV slices showed that this discrepancy appears in inferior slices from the iso center, especially in the distal regions of the proton beams.

This effect seen in the distal region is presumably due to the fact that the Schneider method assigns up to 4.7% higher density to soft tissues than the Water equivalent method (Fig 1). Higher density in the beam path reduces the proton beam range. Because the distal penumbra region of SOBP is typically very steep, even a density change of a few percent within the penumbra region would presumably only result in a range change of less than a millimeter. However, if there is a significant difference along the beam path for many centimeters of tissue, the range could potentially change by a few millimeters, thus mainly affecting dose to organs at risk. Our investigation of H distribution in the entrance and CTV regions indicates that only two major soft tissues are present in the proton beam path: ~40% adipose ($H \sim -100$) and ~60% muscle ($H \sim +40$). (The H distribution in the entire entrance region is very similar to that in neck CTVs, which is presented in Fig. 13.) The density difference for muscle is ~0.1% between the Schneider and Water equivalent methods. However, the density of adipose is 3% higher in the Schneider conversion than the Water equivalent conversion. Such a difference is sufficient in terms of beam range, i.e., could cause deviations of up to a few millimeters. For the cases presented here, we studied range variation based on 95% dose level. It turned out that the

reduction in range varies significantly. In some regions, no apparent difference in range is discernible. On the other hand, there are regions where a few millimeter range difference is observed. Apparently, this is primarily due to the nonuniform distribution of adipose tissue. Regions where more adipose is in the proton path are more significantly affected, compared to regions with less or no adipose in the proton path.

B. Study 2: Dosimetric significance of assignments of elemental composition and mass density

The DVHs using the adjusted Schneider and BEAM methods are presented in Figs. 10 and 11 for the two plans, respectively. Comparison of Fig. 10 to Fig. 5 indicates that for the lung tumor plan the adjustment of mass density does not change the DVHs at all, whereas change of the tissue materials to water will generate DVHs indiscernible to those with the Water equivalent method. This confirms that the discrepancy in DVHs in Fig. 5 is mainly caused by differences in elemental composition. The same conclusion can be drawn for the PNS GTV and CTV. The less smooth area in CTV DVH in Fig. 7 is eliminated in Fig. 11 after air is replaced with water of equal mass density. According to Fig. 11, for the DVHs of the left and right neck, the Schneider method and the BEAM method show different results even by adjusting the parameters. In other words, neither the effect of mass density nor elemental composition could compensate for the DVH discrepancies. This is verified by the dose difference images.

C. Dose to water vs. dose to tissue

The significance of the elemental composition effect (dose to water vs. dose to tissue) is tissue-type dependent (Fig. 3). Figures 12 and 13 illustrate the distributions of Hounsfield numbers in the gross tumor and clinical target volumes. The contoured GTV of the lung tumor plan covers only the solid tumor (soft tissue). The Hounsfield numbers are largely confined in the range from -200 to $+200$, with a narrow peak at $H \sim 40$. Besides soft tissues, the CTV of the lung tumor plan contains a large fraction of lung tissue (47% in volume), and some bony tissue (about 12% in volume). The GTV of the PNS plan contains voxels of soft tissues with Hounsfield number ranging from zero to over 100, along with bony tissues (29% in volume). In addition to the GTV voxels, the CTV has air voxels (11% in volume). Hounsfield numbers in the left/right neck CTVs are strictly bound in the range of -150 to $+150$. The neck CTVs are mainly composed of two types of soft tissues (muscle and adipose), with CT numbers peaked at $+40$ (muscle) and -100 (adipose), respectively. The BEAM method uses the same elemental composition for all bony tissues, and the DVH deviations with the BEAM method for the lung CTV (Fig. 5) and PNS GTV (Fig. 7) come primarily from the soft bony tissue ($H > 120$). As already pointed out, the DVH falloff at around 50–60 Gy for PNS CTV with the Schneider and BEAM conversions is due to air gaps in the target volume. Dose to air is 20–70% lower than dose to water (Fig. 3 does not present the ratio of mass stopping powers for air.) Eleven percent of air volume is sufficient to cause the DVH deviation that appears in Fig. 7. The remaining elemental composition effect for PNS left/right neck CTVs with the mass-density adjusted Schneider conversion (Fig. 11) is due to the presence of a large amount of adipose tissue. In the Schneider conversion method, the elemental composition and weights of the adipose tissue are very different from other soft tissues. As illustrated in Fig. 3(a), the Schneider conversion reports a higher (2–3%) absorbed dose to adipose than dose to water.

Compared to photon and electron beams, energy loss cannot be scaled with mass or electron density in proton therapy. The influence from nonelastic nuclear reactions is more important for high-energy proton beams.²² In addition, multiple scattering effects affect beam broadening. In this study, although nonelastic nuclear interactions and multiple scattering effects were included in all simulations, no separate dose deposition from nuclear reactions was recorded. Simulations were conducted for the PNS case without tracking the secondary particles generated in simulations. The results show that in high dose regions (GTV or CTV),

the variation in absorbed dose from products of nuclear reactions due to the differences in elemental composition in these CT number conversion methods is insignificant. This is different for organs at risk, in particular within the beam path. Secondary particles due to nonelastic nuclear interactions significantly affect the dose in the entrance region of the beam entering the patient. Thus, in proton therapy in general, it is not possible to transform dose to tissue into dose to water or vice versa without taking into account the local linear energy transfer (LET) distribution.²⁴

IV. CONCLUSIONS

We compared three methods for converting Hounsfield numbers to tissue materials for Monte Carlo dose calculation in proton therapy. The results show that the method chosen to assign tissue materials affects the absorbed dose distributions as analyzed by dose-volume histograms. The discrepancy in dose originates from the different assignments of mass density and elemental composition from Hounsfield numbers.

We conclude that the differences are generally small but not insignificant and may affect the predicted proton beam range in particular for deep proton fields (study 1). Typically, slight differences in mass density assignments play only a minor role in the target region. Thus, the conversion methods by Schneider, Bortfield, and Schlegel¹⁴ and the BEAMnrc code^{15,16} result in differences predominantly because of their different assignments in elemental compositions (study 2). In the presence of large tissue inhomogeneities, for head and neck treatments, treatment planning decisions could be affected by these differences because of deviations in the predicted tumor coverage.

The difference is more pronounced for organs at risk because in regions within the SOBP, a slight change in mass density may only cause insignificant surge of dose deposition, whereas in the SOBP penumbra or distal regions, the influence can be much more significant. Thus, outside the target area, differences in elemental composition and mass density assignments both may play a role.

This study was conducted for proton beams under the realistic clinical situations, but it has general instructions for radiation therapy with all heavy charged particle beams. Proton or heavy ion beams present more conformal dose distributions and steep dose gradients, which can be near the critical structures. Radiation therapy with heavy charged particle beams requires more accurate knowledge in relative stopping power in order to convert Hounsfield numbers. For some cases (e.g., neck volumes in the PNS plan studied in this investigation), 2–3% of error in mass density, which is comparable to the accuracy of the calibrated Hounsfield number versus mass density curve, may cause appreciable dose deviations.

Acknowledgements

This work was supported in part by Grant No. P01 CA 21239 from the National Cancer Institute, DHHS. We would like to thank Judy A. Adams (MGH) for providing us with the treatment plans.

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FIG. 1.

Hounsfield number to mass density conversion (relative to $H=0$) as provided in three different methods: Schneider (Ref. 14) (solid line); BEAM (Ref. 16) (dashed line); Water equivalent (dashed-dotted line). The figure in the lower right corner shows a magnification of the region between $H=-50$ and $H=250$.

FIG. 2.

Elemental weights of four major tissue elements (H, C, O, and Ca) as a function of Hounsfield number for three different conversion methods: Schneider (Ref. 14) (solid line); BEAM (Ref. 16) (dashed line); Water equivalent (dashed-dotted line).

FIG. 3. Water-to-tissue mass stopping power ratio as a function of Hounsfield number for the Schneider method (a) and the BEAM method (b).

FIG. 4.
Dose distribution in the iso-center slice for the lung tumor plan. Top to bottom: Schneider, BEAM, and Water equivalent methods.

FIG. 5.

Dose difference images (Schneider: left; first row, BEAM: right; first row) and dose volume histograms for GTV (left; second row) and CTV (right; second row) for the lung tumor plan. Note the scale on the dose axis.

FIG. 6. Dose distribution in the iso-center slice for paranasal sinus of the PNS plan. Top to bottom: Schneider, BEAM, and Water equivalent methods.

FIG. 7.
Dose difference images (Schneider: left; first row, BEAM: right; first row) and dose volume histograms for GTV (left; second row) and CTV (right; second row) for paranasal sinus of the PNS plan. Note the scale on the dose axis.

FIG. 8.
Dose distribution in the iso-center slice for lymph nodes of the PNS plan. Top to bottom:
Schneider, BEAM, and Water equivalent methods.

FIG. 9.

Dose difference images (Schneider: left; first row, BEAM: right; first row) and dose volume histograms for GTV (left; second row) and CTV (right; second row) for paranasal sinus of the PNS plan. Note the scale on the dose axis.

FIG. 10. DVHs for the lung tumor plan. Shown are the curves for the Water equivalent method (solid lines), Schneider and BEAM method with the mass density adjusted to match the one in the Water equivalent method (original elemental composition) (long-dashed and dotted lines, respectively; “1”), and Schneider and BEAM method with the tissue set to water (original mass density assignments) (dashed-dotted and short-dashed lines, respectively; “2”).

FIG. 11.

DVHs for the PNS tumor plan. Shown are the curves for the Water equivalent method (solid lines), Schneider and BEAM method with the mass density adjusted to match the one in the Water equivalent method (original elemental composition) (long-dashed and dotted lines, respectively; “1”), and Schneider and BEAM method with the tissue set to water (original mass density assignments) (dashed-dotted and short-dashed lines, respectively; “2”).

FIG. 12.
Volume distributions of Hounsfield numbers in the GTV and the CTV for a lung cancer case.

FIG. 13.
Volume distributions of Hounsfield numbers for different structures for a paranasal sinus cancer case.