Virtual phantom studies of treatment planning in charged particle radiotherapy

An essential input in proton radiotherapy treatment planning is the calibration curve linking CT number with proton stopping power. This work considers the accuracy of the calibration process and its effect on the dose delivered.

1. Introduction
One of the necessary steps in commissioning a CT scanner for use in treatment planning is to link the image intensity recorded for tissues with their ability to slow protons. This enables the accurate calculation of dose-depth curves within the patient and so gives confidence that the dose received by a patient matches the planned distribution.

As the elemental composition of human tissues is somewhat variable, calibration is usually performed with tissue equivalent materials. The stoichiometric calibration method of characterising CT scanners using such materials was first described by Schneider et al. [1]. It produces coefficients related to the levels of photoelectric effect, coherent scattering and incoherent scattering observed in the scanner. These are typically used with the published recommendations of the ICRP and ICRU for tissue composition to estimate CT numbers in Hounsfield Units (HU), and the associated stopping power via the Bethe-Bloch formula.

The accuracy of CT to stopping power curve production can be explored in a system where all stopping powers can be explicitly calculated. ICRP Publication 110 [2] provides such a system in its voxelized reference male and female phantoms, which are based on CT scans of real patients with corrections that bring the anatomy in line with population averages. Each is segmented into more than 140 organs of specified elemental composition and mass density. The male version of this phantom is pictured in Figure 1.

2. Determining CT Number and Stopping Power
Typical CT scanners consist of a fan-beam x-ray source and detector array, mounted diametrically opposite on a ring gantry that rotates around the patient. A mathematical transformation of the detector signal received at many angles yields a measure of x-ray attenuation in each voxel (3D pixel), which is calibrated to the Hounsfield Units (HU) scale (air = 1000 and water = 0) to produce an axial slice of the final image. The patient is moved relative to the gantry to build up volumetric data.

X-ray intensity attenuates exponentially with depth and the coefficient may be calculated for materials of known composition using an additivity rule:

\[
\mu = \sum_{i} w_i \mu_i
\]

where \( i \) labels the elements, \( \mu \) is the linear attenuation coefficient, \( p \) is the mass density and \( w \) is the fraction by weight.

Elemental values of \( \mu \) are published by NIST [3]. A material’s HU value can thus be estimated by using the formula above for the material in question, and linearly scaling the result to the linear attenuation coefficients found for air and water. Unfortunately the process is somewhat complicated by the fact that \( \mu \) is energy dependent and so is a real CT scanner, whose tube produces a spectrum of x-rays, it will vary with depth. This is resolved here by considering an effective \( \mu = \frac{1}{\rho} \int d\mu \) at a particular reference depth.

The convention in treatment planning software is to input stopping power relative to water. The Bethe-Bloch equation is generally used to stoppings powers - corrections to it exist, but they generally only apply at lower energies where the residual range of protons is small. Here, the software package SRIM was used to calculate stopping powers at an energy of 100MeV. This package is capable of considering the effect of molecular bonds in materials, but as such information is not available for the ICRP 110 tissues the calculations assume simple elemental mixes.

The points plotted in Figure 2 show the results of HU and stopping power calculations for ICRP 110 tissues. The energy spectrum used was representative of a CT tube at 120kVp.

3. Simulation of CT and Stoichiometric Calibration
The simple method described for calculating Hounsfield Unit values in section 2 is useful, but a voxel-by-voxel replacement of the phantom’s organs with the respective CT number leads to a quantised CT scan with homogenous organs. Actual CT scans show variations within organs due to reconstruction artefacts and beam hardening, as well as for anatomical detail. In order to account for some of these, a simulation of a CT scanner was built in MATLAB. Figure 3 shows the approach this program takes. Rays are traced through transverse slices of the phantom at a range of angles, using the calculated attenuation coefficients for each voxel to attenuate each ray as appropriate. Once all data is collected, an inverse radon transform is performed to reconstruct the image. Simulating images containing water and air blocks allows the phantom scans to be scaled to HU. The inset of Figure 3 illustrates the difference between a homogenous and simulated scan.

The CT simulator could be characterised as a physical machine would be, and the approach followed was that of Schaffner [5]. The mean HU values found in tissue equivalent block tissues were used to find best-fit parameters for the stoichiometric calculation:

\[
HU = \mu_\text{rel}^\text{rel} (2\rho + c) - 1000 \quad \text{where } \mu_\text{rel} = \text{electron density relative to water}
\]

The best-fit parameters were found to be \( A = 0.0338, B = 2.1118, C = -866.5448 \). These could then be used to estimated HU values for all the ICRP 110 tissues, which were categorised into adipose, organs and muscle and bone categories. Individual linear fits for CT to stopping power were performed in each tissue category and then connected in the manner of Schaffner, which aims to set the stopping power to that of the most common tissue type at each CT value. The resulting calibration curve can be seen in Figure 2.

4. Effect on Delivered Dose
It was decided that a simple treatment plan for prostate cancer would be produced on the male phantom, as in this case the CT is a well-defined organ. The TPS used was Elekta XiO 4.6.2, and simulated scans were converted to DICOM using CERR [6], a MATLAB-based radiotherapy research platform. Following Meyer et al. [7], the plan employed parallel opposed lateral spot-scanned beams, with a target tumor dose of 76Gy to 100% of the CTV and 95% of the PTV (lateral margins 1.2cm, posterior 0.6cm, others 0.8cm). The resulting dose distribution can be seen in Figure 4.

Through manipulation of the XIO file system, it is possible to switch the CT scan which a plan is associated with. Using this technique, comparison could be made between the dose distribution with and without reconstruction artefacts, and with different CT number to stopping power curves. The plan was initially made on the CT with artefacts, using the stoichiometrically-derived CT number to stopping power curve.

Figure 4 shows the result of changing the scan to one in which the CT number within each organ is replaced by a single value, equal to the median HU value within that organ. It can be seen that the dose volume histograms within the STV and bladder change only minutely, to the degree that the two lines are barely distinguishable. One proposed reason for this is that the STV margins are sufficient to retain almost complete coverage of the CTV despite the differing variation in HU. It could also be that the HU distributions within many organs are approximately symmetrical about the median, so any increase in stopping power within one voxel is likely to be counteracted by a decrease in another further down the beam’s path.

5. Conclusion
It has been shown that, within the phantom system detailed in ICRP 110, the stoichiometric calibration approach taken by Schaffner can produce a good approximation of the actual CT and stopping power values for tissues. However, beam hardening effects measured HU values can only truly be performed for points where the incident x-ray spectrum is identical - this is not explored in detail here.

The planning study has revealed that the presence of reconstruction artefacts in the scan result in no significant dose variation within the CTV or bladder in the prostate treatment plan considered. It is likely, however, that any differences would be more pronounced just outside the PTV border where dose gradients are steepest. To identify such regions, future studies will consider the differences between calculated dose matrices.

Proposed further work will involve input of the exact calculated stopping power for each of the phantom’s organs into the treatment planning system. This will involve replacing the CT number with a label of tissue type and the use of a label to stopping power calibration curve instead. It is hoped this process will bring a better understanding of the precise differences in dose distribution seen during the CT calibration process.