DETERMINATION OF CT-TO-DENSITY CONVERSION RELATIONSHIP
FOR IMAGE-BASED TREATMENT PLANNING SYSTEMS

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(Received 18 May 2004; accepted 23 May 2005)

Abstract—The implementation of tissue inhomogeneity correction in image-based treatment planning will improve the accuracy of radiation dose calculations for patients undergoing external-beam radiotherapy. Before the tissue inhomogeneity correction can be applied, the relationship between the computed tomography (CT) value and density must be established. This tissue characterization relationship allows the conversion of CT value in each voxel of the CT images into density for use in the dose calculations. This paper describes the proper procedure of establishing the CT value to density conversion relationship. A tissue characterization phantom with 17 inserts made of different materials was scanned using a GE Lightspeed Plus CT scanner (120 kVp). These images were then downloaded into the Eclipse and Pinnacle treatment planning systems. At the treatment planning workstation, the axial images were retrieved to determine the CT value of the inserts. A region of interest was drawn on the central portion of the insert and the mean CT value and its standard deviation were determined. The mean CT value was plotted against the density of the tissue inserts and fitted with bilinear equations. A new set of CT values vs. densities was generated from the bilinear equations and then entered into the treatment planning systems. The need to obtain CT values through the treatment planning system is very clear. The 2 treatment planning systems use different CT value ranges, one from −1024 to 3071 and the other from 0 to 4096. If the range is correct, it would result in inappropriate use of the conversion curve. In addition to the difference in the range of CT values, one treatment planning system uses physical density, while the other uses relative electron density. © 2005 American Association of Medical Dosimetrists.

Key Words: CT scanning, Tissue characterization, Electron density, Treatment planning system, Tissue inhomogeneity.

INTRODUCTION

Image-based treatment planning has become the standard of practice in radiotherapy.1–7 Patient data are now acquired within a short time using the new generation fast helical computed tomography (CT) scanners. The CT images are then downloaded into a treatment planning system. At the treatment planning workstation, the CT image set is assigned as the primary set of patient data for treatment planning. Next, the tabletop is removed from the CT images and the external patient contours are extracted using edge-detection technique. In addition, the contours of the internal organs can be identified. Additional tools such as image fusion and co-registration are available to allow visualization of targets seen on other tomographic modalities superimposed onto the CT image set for delineation. Based on this information, a virtual patient can be constructed for 3-dimensional treatment planning. Inverse treatment planning for intensity-modulated radiation therapy can be implemented if the target volumes are drawn, and dose goal and dose constraint objectives are defined for these targets and organs at risk.

Tissue inhomogeneity correction should be applied to have accurate radiation dose calculations, in particular, to the lung. Studies have shown that uncorrected treatment plans can produce radiation dose errors exceeding 30% the prescribed dose.8,9 The implementation of tissue inhomogeneity correction is facilitated by the availability of image-based treatment planning systems. Here, the tissue inhomogeneity is derived by converting the CT value in each voxel into radiological parameters such as relative electron density or physical density. The CT value and radiological parameter relationship is typically established empirically by scanning a tissue characterization phantom. This paper reports the proper procedure of relating CT values to radiological parameters and implementing them in the treatment planning systems.

METHODS AND MATERIALS

The physical features of the tissue characterization phantom, CIRS Model 062 (CIRS Tissue Simulation Technology, Norfolk, VA), are shown in Fig. 1. It is designed with a height of 270 mm and a width of 330 mm for an abdominal scan. However, it can be separated into a smaller size head phantom with a radius of 90 mm. The holes to accommodate the inserts are arranged in 2 concentric rings with radii of 60 mm and 115.3 mm. Each ring will have 8 holes equally spaced around the...
There is also a hole at the center of the phantom; hence, it can accommodate a total of 17 inserts. The commercially available tissue characterization phantom comes with 17 inserts and a titanium insert as an option as listed in Table 1. The relative electron density ranges from 0.190 to 1.512 or correspondingly from 0.195 g/ml to 1.609 g/ml for physical density. Inserts that can accommodate thermoluminescence dosimeters (TLDs) or films are also available upon request. The arrangement of the insert materials is shown in Fig. 2.

The GE Lightspeed Plus scanner is a third-generation (the tube and detector array are attached to the same rotating frame and move simultaneously) multislice scanner with a 70-cm bore size aperture and 496.9-mm scanning field of view (SFOV). It has a 16-row detector system and an 8-row data acquisition system (DAS) to allow for simultaneous acquisition at different longitudinal locations as opposed to single-row detector array of conventional single-slice CT scanners. With this system, the scanner provides 8 data rows of output, allowing for simultaneous acquisition of as many as 8 axial images per gantry rotation. The 16-row detector is segmented into 16 physically separated cells in the longitudinal direction to provide post-patient collimation of the x-ray beam. The actual detector cell size is 2.19 mm in the longitudinal direction. The effective cell size is 1.25 mm in the longitudinal direction at the isocenter. These cells in the longitudinal direction can be summed into a “macro-cell” giving the thickness of 1.25 mm (1 cell), 2.50 mm (2 cells), 3.75 mm (3 cells), and 5.0 mm (4 cells). The number of macro-cell in the longitudinal direction forms the macro-row. The maximum scanned width for a 16-row detector would be 20 mm (4 × 5.0 mm). The scanner has a shorter focal spot to imaging isocenter distance of 541 mm compared to 630 mm in a single-slice scanner. The focal spot to detector distance is 949 mm. The system is equipped with large and small focal spots with dimensions of 1.2 mm × 1.2 mm and 0.9 mm × 0.7 mm, respectively.

The scanner has the ability to generate images of various thickness, made possible with the helical scanning mode. However, the image thickness is dependent on the pitch, which is defined as the ratio of the table travel per gantry revolution to the beam collimation and number of detectors in a macro-row. For example, if a scanning of 4 slices is obtained using 4 detector macro-row × 2 pitches, then reconstruction of images of thickness from 1× to 4× the detector macro-cell size is possible.

Before performing the CT scan, the tissue characterization phantom was set up such that the laser passed through the center of the phantom. Next, the scanning table was arranged such that the internal CT laser passed through the center of the phantom; the CT scanner was then set to zero. A typical scanning parameter of 120 kVp used in standard scanning protocols with 5-mm slice thickness was used. After the scan, CT images of the tissue characterization phantom were downloaded into the network and forwarded to the Varian Eclipse and the Philip Pinnacle treatment planning systems.

At each treatment planning system, these CT images were converted into treatment planning images. The

Table 1. Tissue descriptions and densities

<table>
<thead>
<tr>
<th>Materials</th>
<th>Location in Figure 2</th>
<th>Physical Density (g/ml)</th>
<th>Relative Electron Density</th>
</tr>
</thead>
<tbody>
<tr>
<td>Syringe water</td>
<td>1</td>
<td>1.000</td>
<td>1.000</td>
</tr>
<tr>
<td>Lung (inhale)</td>
<td>2</td>
<td>0.195</td>
<td>0.190</td>
</tr>
<tr>
<td>Lung (exhale)</td>
<td>3</td>
<td>0.495</td>
<td>0.489</td>
</tr>
<tr>
<td>Breast (50/50)</td>
<td>4</td>
<td>0.991</td>
<td>0.976</td>
</tr>
<tr>
<td>Dense bone 800 mg/ml HA</td>
<td>5</td>
<td>1.609</td>
<td>1.512</td>
</tr>
<tr>
<td>Trabecular bone</td>
<td>6</td>
<td>1.161</td>
<td>1.117</td>
</tr>
<tr>
<td>Liver</td>
<td>7</td>
<td>1.071</td>
<td>1.052</td>
</tr>
<tr>
<td>Muscle</td>
<td>8</td>
<td>1.062</td>
<td>1.043</td>
</tr>
<tr>
<td>Adipose</td>
<td>9</td>
<td>0.967</td>
<td>0.952</td>
</tr>
<tr>
<td>Titanium</td>
<td>4.507</td>
<td>3.735</td>
<td></td>
</tr>
</tbody>
</table>
CT images were then retrieved and a region of interest (approximately 300 mm$^2$ for large inserts and 30 mm$^2$ for small inserts) was drawn on the central portion of each insert to determine the mean CT value and its standard deviation. The CT values were then plotted against the relative electron density or physical density of the tissue. The plot was fitted with 2 linear equations for CT values from the $-1000$ to $0$--$50$ range and the above $0$--$50$ range.$^{13}$

RESULTS

Figure 3 shows the axial CT image of the tissue characterization phantom. Because the axial CT image was generated through the filtered backprojection reconstruction technique, streaking artifact may be present, due to its neighboring high attenuation insert. However, as seen in the figure, there are clearly no artifacts, indicating that the CT value of each insert is minimally influenced by its neighboring value. Regions of interest were drawn on the central portion of the inserts to determine the mean CT values and their standard deviations.

Figure 4 shows the CT-to-density conversion or tissue characterization curve for the Varian Eclipse treatment planning system. The relationship is plotted as the mean CT values vs. relative electron densities as required by the dose calculation algorithm. In addition, the range of CT values were found to be from $-1000$ to $3071$. The plot shows scattered data points but its behavior can be predicted to consist of 2 linear relationships for CT value ranges from $-1000$ to $0$ and above $0$. In addition, it should be recognized that the slope of these straight lines and the point of inflection change somewhat from CT scanner to CT scanner.$^8$ The fitted equations has correlation coefficients, $R^2$, of greater than 0.99.

On the other hand, the tissue characterization curve for the Philips Pinnacle treatment planning system (Philips Medical, Andover, MA) is totally different compared to the Varian Eclipse treatment planning system (Varian, Andover, MA). As shown in Fig. 5, the mean CT values are plotted against the physical densities of the tissues. The use of physical densities instead of relative electron densities should be noted for the Philips Pinnacle treatment planning system. In addition, the range of CT values was found to be from $0$ to $4096$. This CT value range is different from the Varian Eclipse treatment planning system. The CT value of the image set for the Philips Pinnacle treatment planning system is derived by adding a numerical value of $1000$ to the CT value of the image set from the CT scanner. As a standard definition for CT scanner, the CT value is defined to be $-1000$ for air and $0$ for water. Again, Fig. 5 shows scattered data points but can be fitted with bilinear relationships, one relationship from CT values of $0$ to $1000$ and the other from over $1000$. 

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**Fig. 3.** Axial image of the tissue characterization phantom showing little to no streaking artifacts.

**Fig. 4.** Tissue characterization curve for the Varian Eclipse treatment planning system using relative electron densities.

**Fig. 5.** Tissue characterization curve for the Philips Pinnacle treatment planning system using physical densities.
DISCUSSION

The human body is composed of various tissues and cavities that have radiological properties that are different from water. In some instances, foreign materials such as metallic and non-metallic prostheses may be implanted into patients to improve their quality of life.14–18 The radiation dose and its distribution are affected by these inhomogeneities, and because treatments are becoming more conformal, there are opportunities for geometrical misses of the isodose coverage of the target. To maximize the benefit of radiation therapy, the dose delivered to all irradiated tissues in the presence of these inhomogeneities should be predicted accurately. AAPM Report No. 85 has alluded to the need for the implementation of tissue inhomogeneity corrections.14

The implementation of tissue inhomogeneity corrections in image-based treatment planning systems requires prior knowledge of the radiological properties of these tissues, cavities, and prosthetic materials. With the advent of CT, it is possible to derive the density information in vivo and incorporate it into the dose calculation process. A number of published articles have shown a correlation between the radiological properties and CT values.19,20 As such, radiological properties of tissues, cavities, and prosthetic materials for treatment planning systems are derived in vivo and exclusively through the CT values.

It is necessary to determine the CT-to-density conversion curve empirically because the CT values are dependent on the individual scanner parameters such as kVp/filtration and reconstruction algorithm. The tissue characterization phantom is solely designed to establish the CT-to-density conversion curve. Appropriate scanning parameters must be chosen to minimize the effect of kVp and slice thickness on the CT-to-density conversion curve. The kVp chosen should be the typical values used in scanning protocols for radiotherapy patients. After the scan, the CT images must be downloaded into the individual treatment planning system before determining the mean CT value and its standard deviation. This is important because the CT values used may be different for different treatment planning systems. Furthermore, the type of radiological parameters required can be different, such as relative electron density vs. physical density of tissues, again depending on the dose calculation algorithms. After the mean CT values are determined, the data must be fitted with the bilinear equations. After the fit, these equations will be used to generate the CT value vs. the radiological parameter data for input into the conversion file in the treatment planning system. This last step is important otherwise the dose algorithm will interpolate between measured scattered data points.

REFERENCES