Optimum size of a calibration phantom for x-ray CT to convert the Hounsfield units to stopping power ratios in charged particle therapy treatment planning

T. Inaniwa¹*, H. Tashima² and N. Kanematsu³

¹Department of Accelerator and Medical Physics, National Institute of Radiological Sciences, QST, 4-9-1 Anagawa, Inage-ku, Chiba 263-8555, Japan
²Department of Radiation Measurement and Dose Assessment, National Institute of Radiological Sciences, QST, 4-9-1 Anagawa, Inage-ku, Chiba 263-8555, Japan
³Medical Physics Section, National Institute of Radiological Sciences Hospital, QST, 4-9-1 Anagawa, Inage-ku, Chiba 263-8555, Japan

*Corresponding author. Department of Accelerator and Medical Physics, National Institute of Radiological Sciences, QST, 4-9-1 Anagawa, Inage-ku, Chiba 263-8555, Japan. Tel: +81-43-251-3170; Fax: +81-43-251-1840; Email: inaniwa.taku@qst.go.jp

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ABSTRACT

In charged-particle therapy treatment planning, the volumetric distribution of stopping power ratios (SPRs) of body tissues relative to water is used for patient dose calculation. The distribution is conventionally obtained from computed tomography (CT) images of a patient using predetermined conversion functions from the CT numbers to the SPRs. One of the biggest uncertainty sources of patient SPR estimation is insufficient correction of beam hardening arising from the mismatch between the size of the patient cross section and the calibration phantom for producing the conversion functions. The uncertainty would be minimized by selecting a suitable size for the cylindrical water calibration phantom, referred to as an ‘effective size’ of the patient cross section, $L_{\text{effective}}$. We investigated the $L_{\text{effective}}$ for pelvis, abdomen, thorax, and head and neck regions by simulating an ideal CT system using volumetric models of the reference male and female phantoms. The $L_{\text{effective}}$ values were 23.3, 20.3, 22.7 and 18.8 cm for the pelvis, abdomen, thorax, and head and neck regions, respectively, and the $L_{\text{effective}}$ for whole body was 21.0 cm. Using the conversion function for a 21.0-cm-diameter cylindrical water phantom, we could reduce the root mean square deviation of the SPRs and their mean deviation to $\leq 0.011$ and $\leq 0.001$, respectively, in the whole body. Accordingly, for simplicity, the effective size of 21.0 cm can be used for the whole body, irrespective of body-part regions for treatment planning in clinical practice.

Keywords: beam hardening effect; computed tomography simulation; standard body tissues; computational human phantom; charged particle therapy; treatment planning

INTRODUCTION

In charged-particle therapy treatment planning, the accurate prediction of particle range in patients is essential for conformal dose delivery to the target. Particle range is determined by integrating the stopping power ratios (SPRs) of body tissues relative to water along the beam path in a patient. The volumetric distribution of the SPRs in a patient is conventionally obtained from the x-ray computed tomography (CT) data, using a predetermined polylne relationship between CT number and SPR of the body tissues, referred to as the CT number-to-SPR conversion function [1–3]. Uncertainties of SPR estimation can induce range uncertainties of up to 3.5% [4, 5], which in current clinical practice are considered by adjusting the corresponding distal and proximal margins to the target. Yang et al. [6] grouped the uncertainties in the SPR estimation into several categories according to their sources and estimated their degrees for (lung tissues, soft tissues, bone tissues):...
uncertainties in patient CT imaging (3.3%, 0.6%, 1.5%), uncertainties related to the CT-number-to-SPR conversion functions (3.8%, 1.4%, 1.7%), uncertainties in mean excitation energies (0.2%, 0.2%, 0.6%) and uncertainties due to the energy dependence of SPR not commonly accounted for by a dose algorithm (0.2%, 0.2%, 0.4%). To fully exploit the advantages of charged-particle therapy, these uncertainties should ultimately all be minimized, no matter how large or small they actually are.

In this study, we focused on the uncertainties in patient CT imaging mainly caused by the acquisition of the CT numbers themselves. The CT number is directly related to the linear attenuation coefficient of the object for x-rays, and is usually calibrated to 0 for water and −1000 for air. The CT number is strongly affected by the x-ray energy spectrum at the point of measurement. The initial energy spectrum depends on the scanner properties, such as tube voltage, target material, filter, and detector sensitivity. The spectrum varies in the materials traversed by the x-rays up to the point of interest due to energy-dependent attenuations, namely beam hardening. The variation in CT number between scanners can be handled by creating a CT-number-to-SPR conversion function specifically for each scanning condition of each scanner. The variation in CT number due to the beam hardening is considered to be minimized by creating the CT-number-to-SPR conversion functions using x-ray calibration phantoms with typical body sizes, e.g. 10–40 cm [6–8]. However, the mismatch between the size of the object and the calibration phantom induces non-negligible effects in SPR of body tissues. Schaffner and Pedroni [9] reported that CT-number variation of bone tissues between calibration phantoms of 15 cm and 30 cm diameters leads to an uncertainty in SPR of 1.5%. Yang et al. [6] reported that CT-number variations in bone and lung tissues between phantoms of 16 cm and 32 cm diameters lead to uncertainties of 1.9% and 2.6%, respectively. These studies, however, were solely based on the CT-number measurements of the tissue substitutes in cylindrical calibration phantoms. The impact of the beam hardening on the uncertainty of SPR estimation in a patient has never been investigated using a realistic patient geometry. In addition, the optimum size of the x-ray calibration phantom for minimizing the uncertainty has not been determined or proposed. The optimum phantom size, in sum, represents an ‘effective size’ of the patient cross section for beam hardening, and investigated the effective patient-cross-section size for patient SPR estimation in charged-particle therapy using an ideal CT system and realistic human tissue computational phantoms.

MATERIALS AND METHODS

X-ray CT

We modeled an ideal CT scanner consisting of an x-ray source generating parallel broad beams and a detector-array with a 100% detection efficiency with perfect anticscatter grids. In addition, CT scanning was simulated based on a theoretical x-ray spectrum with an infinite imaging dose. The reconstructed CT image was not affected by factors such as statistical noise, electronic noise or scatter, so we focused our investigation on the error in SPR estimation due to the beam-hardening effects.

The x-ray energy spectrum of a CT scanner $\Phi(E)$ was generated using the SpecCalc x-ray generator program [10]. For the generation, we used the following parameters: a normal tube voltage of 120 kVp with variations of ±10 kVp, tungsten as the target material, a 7° anode angle and a 7.4-mm-thickness aluminum filter. The generated energy spectra of the CT scanner are shown in Fig. 1.

The simulated detector signal of the incident x-rays without an object was calculated as:

$$Q_0 = \int_{0}^{E_{\text{max}}} E \cdot \Phi(E) \, dE,$$

where $E_{\text{max}}$ is the maximum energy of the x-rays, i.e. 120 ± 10 keV. The measured signal of the x-rays transmitted through an object that enter the detector was calculated as:

$$Q = \int_{0}^{E_{\text{max}}} E \cdot \Phi(E) \cdot \exp\left(-\int_{0}^{t} \mu(E, t) \, dt\right) \, dE,$$

where $\mu(E, t)$ is the linear attenuation coefficient of the object at a position $t$ along the projection line with length $l$. Projection of the object by the x-rays in a given direction was defined as:

$$\lambda_p = \ln \frac{Q_0}{Q}.$$

To eliminate the effect of beam hardening by the polychromatic x-rays, $\lambda_p$ of each projection line was corrected to the projection by a monochromatic x-ray with equivalent attenuation at the incident, $\lambda_{\text{MT}}$, using a look-up table describing water-equivalent thickness [11]. Tomographic image reconstruction from a sinogram of scanned and corrected projections gave the mean attenuation coefficient of the object over the incident energy spectrum, $\mu = \int_{0}^{E_{\text{max}}} \mu(E) \Phi(E) \, dE / \int_{0}^{E_{\text{max}}} \Phi(E) \, dE$. In this study, projections $\lambda_{\text{MT}}$ were generated for every 1° step, and a 2D filtered backprojection algorithm with a ramp filter was used for the reconstruction.

The values of $\langle \mu \rangle$ were converted into CT number $H$ in Hounsfield units (HUs):

$$H = -1000 \frac{Q_0}{Q}.$$

![Fig. 1. The x-ray energy spectra $\Phi(E)$ of 120 kVp (solid curve), 110 kVp (dashed-curve) and 130 kVp (dotted curve) beams from the simulated CT scanner.](https://academic.oup.com/jrr/article-abstract/59/2/216/4582868)
H-to-(S/S_w) conversion functions

To produce the CT number to SPR conversion functions, H-to-(S/S_w), we followed the procedure reported by Kanematsu et al. [1], based on the standard tissue data in the International Commission on Radiological Protection Publication 110 [12]. They defined 11 representative tissues of the human body with mass density \( \rho \) and elemental weights \( w \) and compiled their SPRs. When the CT numbers \( H \) of the 11 tissues are known, based on the hypothesis that an arbitrary tissue is a binary mixture of the representative tissues adjacent with higher and lower \( H \), the SPR of the arbitrary tissue can be derived from its CT number \( H \) by polyline interpolation.

To determine the CT number \( H \) of the 11 tissues, we simulated cylindrical water phantoms of diameter \( L \) with a 2.5-cm diameter insert of one of these tissue materials. The energy-dependent attenuation coefficient \( \mu \) of water, air, and the tissue materials were determined by interpolation from XCOM, a database of photon cross sections provided by the National Institute of Standards and Technology (NIST) at http://physics.nist.gov/PhysRefData/Xcom/html/xcom1.html. The inserts at the center of the water phantoms were scanned by the CT scanner individually, and their CT numbers \( H \) were scanned by the CT scanner individually, and their CT numbers \( H \) were produced by interpolating the CT number \( H \) with polyline interpolation.

To confirm the accuracy of the produced H-to-(S/S_w) conversion functions, 53 standard body tissues with \( \rho \) and \( w \) listed in the ICRP report were applied to the simulation of CT scanning as inserts of water phantoms with \( L = 10, 30 \) and 50 cm. The H-to-(S/S_w) relations of the standard body tissues determined for respective phantom sizes \( L \) were compared with the corresponding H-to-(S/S_w) conversion functions.

ICRP computational phantom

We used volumetric models of the reference male and female phantoms provided in the ICRP Publication 110 [12]. The height and mass of the models were (176 cm, 70 kg) and (167 cm, 59 kg), respectively. Their slice thickness (voxel height) and voxel in-plane resolution were (8.0 mm, 2.137 mm) and (4.84 mm, 1.775 mm). For better spatial resolution, in this study, the voxel in-plane resolutions were reduced to one-third of the original resolutions, i.e. 0.7123 mm and 0.5917 mm, respectively. Since charged-particle therapy is applied to a variety of tumors [13], the phantom data for pelvis, abdomen, thorax, and head and neck regions were included in the analysis, as shown in Fig. 4a. The phantom arms were removed to simulate the arm abduction used in treatment.

Reference SPR

The volumetric models of the male and female phantoms were converted to the SPR distributions by the following steps. The electron density \( n_e \) of body tissues with a given \( \rho \) and \( w \) was derived by:

\[
n_e = \frac{\rho}{\mu} \sum_i \frac{w_iZ_i}{A_i},
\]

where \( \mu \) is the electron mean free path in water and air, respectively.

Predicted SPR

The volumetric models of male and female phantoms were scanned by the CT scanner slice-by-slice. The energy-dependent attenuation coefficients of 53 tissue materials \( \mu \) were determined with XCOM. The reconstructed CT images represented in HUs were then converted to the SPR distributions by the H-to-(S/S_w) conversion function for a phantom size \( L \). We refer to such a derived SPR as the ‘reference SPR’, and we symbolize it as \((S/S_w)_{ref}\) hereafter.

SPR error analysis

From the reference and predicted SPR distributions of the male and female phantoms, voxel-by-voxel absolute deviation of the SPR, \( \delta_S = (S/S_w)_{ref} - (S/S_w)_{prd} \) was calculated. The root mean square deviation of the SPR, \( \sqrt{\delta_S^2} \), was derived to investigate the absolute error in the SPR estimation with the H-to-(S/S_w) conversion functions. The mean deviation of the SPR, \( \bar{\delta_S} \) was also derived to estimate the range error of proton beams in a patient.

Effective size of patient cross section for H-to-(S/S_w) conversion functions

We varied the diameter of the x-ray calibration phantom \( L \) from 6 cm to 50 cm in 1 cm steps, and 45 H-to-(S/S_w) conversion functions were produced. For each conversion function, the \( \delta_S \) map was obtained on each slice. We derived the root mean square deviation of the SPR, \( \sqrt{\delta_S^2} \) in each body-part region (pelvis, abdomen, thorax, and head and neck) of the male and female phantoms. The
average value of $\sqrt{\delta S^2}$ between the two phantoms was used to determine the effective size of the patient cross section for SPR estimation for each body-part region, $L_{\text{effective}}$. The effective size $L_{\text{effective}}$ was also determined for respective slice positions of the male and female phantoms. To investigate the slice-by-slice variation of $L_{\text{effective}}$ across the body-part regions, the standard deviation of $L_{\text{effective}}, \sigma_{\text{effective}}$, was determined for the respective regions.

The uncertainty in the determined $L_{\text{effective}}$ in each body-part region due to variations in body size was evaluated by one-half of the difference between the effective size of the male phantom $L_{\text{male}}$ and that of the female phantom $L_{\text{female}}$, i.e. $\Delta L \equiv (L_{\text{male}} - L_{\text{female}})/2$. In addition, the uncertainty in the determined $L_{\text{effective}}$ due to variations in initial energy spectrum of the CT scanner was evaluated by the deviations in the effective phantom sizes for the energy spectra with tube voltages of 110 kVp and 130 kVp from that with 120 kVp, $\Delta L_{110}$ and $\Delta L_{130}$.

**RESULTS**

**H-to-(S/S_w) conversion functions**

Figure 2 shows the H-to-(S/S_w) conversion functions produced by the x-ray calibration phantoms of $L = 10$, 30 and 50 cm. Although there was a small discrepancy around $-150$ HU for adipose tissues, three conversion functions practically coincided with each other up to $+100$ HU for soft tissues. Subsequently, these functions deviated gradually with HU. Since beam hardening up to the point of the insert was milder in the smaller phantom, the slope of the conversion function for $L = 10$ cm was gentler than the slopes for $L = 30$ or 50 cm. The SPRs at 1400 HU, i.e. for the bone tissues, derived by the functions for $L = 10$, 30 and 50 cm were 1.61, 1.70 and 1.76, respectively. This implied that the error in the predicted SPR of bone tissues could be $\pm 4\%$ due to beam hardening, even if a moderate size of $L = 30$ cm was used for the conversion function. Further, the error in the predicted SPR could be $\pm 6\%$ for the teeth at $\approx 2700$ HU.

Correlations between $H$ and $S/S_w$ of 53 ICRP body tissues determined with the x-ray calibration phantoms of $L = 10$, 30 and 50 cm are plotted by colored plus symbols in Fig. 2. The ICRP tissues distribute around the polylines with symbols in Fig. 2. The ICRP tissues distribute around the polylines with symbols in Fig. 2. The ICRP tissues distribute around the polylines with symbols in Fig. 2. The ICRP tissues distribute around the polylines with symbols in Fig. 2. The ICRP tissues distribute around the polylines with symbols in Fig. 2. The ICRP tissues distribute...
Fig. 4. (a) Water-equivalent thickness distribution of the ICRP female phantom in the AP (PA) direction. White solid lines indicate the boundaries of anatomical regions, i.e. pelvis, abdomen, thorax, and head and neck. (b) The root mean square deviation $\delta_S^2$ and (c) the mean deviation of the SPR $\delta_S$, at each slice position by the $H$-to-$(S/S_w)$ conversion functions with $L = 10$ cm (black curves), 30 cm (red curves), 50 cm (green curves), 21 cm (blue curves), the region-specific $L_{\text{effective}}$ (light blue curves) and the slice-specific $L_{\text{effective}}$ (pink curves). (d) The effective patient-cross-section size $L_{\text{effective}}$ determined at each slice position.

Fig. 3. Axial $(S/S_w)_{\text{ref}}$ distributions (upper row) on the head (left column), thorax (2nd left column), abdominal (2nd right column) and pelvic (right column) planes. The corresponding $\delta_S$ distributions by the $H$-to-$(S/S_w)$ conversion functions with $L = 10$ cm (2nd upper row), 30 cm (middle row) and 50 cm (2nd lower row), and the effective size $L_{\text{effective}}$ of the slices (lower row). The root mean square deviation and the mean deviation of the $\delta_S$ distribution are shown in square brackets $\left[ \sqrt{\delta_S^2} : \delta_S \right]$. 
Fig. 5. Root mean square deviation of SPR, $\sqrt{\delta S^2}$, for different phantom sizes $L$ in (a) the whole body, (b) head and neck, (c) thorax, (d) abdominal and (e) pelvic regions of the male phantom (blue curves), female phantom (red curves), and their average (black curves). The solid, dashed, and dotted curves are $\sqrt{\delta S^2}$ for an x-ray tube voltage of 120 kVp, 110 kVp and 130 kVp, respectively.
Table 1. Effective sizes of the patient cross section, $L_{\text{effective}}$, in the whole body, head and neck, thorax, abdominal and pelvic regions

| Region          | $L_{\text{effective}}$ [cm] | $\sigma_{\text{effective}}$ [cm] | $\Delta L$ [cm] | $\Delta L_{110}$ [cm] | $\Delta L_{130}$ [cm] |
|-----------------|-------------------------------|----------------------------------|-----------------|----------------------|----------------------|
| Whole body      | 21.0                          | 4.8                              | 2.5             | -0.1                 | +0.1                 |
| Head/neck       | 18.8                          | 4.7                              | 1.5             | +0.2                 | -0.2                 |
| Thorax          | 22.7                          | 6.2                              | 4.1             | -0.2                 | +0.1                 |
| Abdomen         | 20.3                          | 2.3                              | 1.7             | -0.3                 | +0.3                 |
| Pelvis          | 23.3                          | 4.5                              | 2.5             | -0.3                 | +0.2                 |

The standard deviation of $L_{\text{effective}}$ across each body-part region, $\sigma_{\text{effective}}$. The deviation in $L_{\text{effective}}$ due to variation in body size, $\Delta L$, and the deviations due to the variation in initial energy spectrum of the CT scanner, $\Delta L_{110}$ and $\Delta L_{130}$.

were slightly different for different body-part regions, while their variations were $<\pm 2.5$ cm around $L_{\text{effective}} = 21.0$ cm of the whole body. The effective size $L_{\text{effective}}$ determined for each slice position of the female phantom is shown in Fig. 4d. The variations of $\sqrt{\delta_S^2}$ and $\delta_S$ with slice position of the female phantom by the conversion functions with $L = 21.0$ cm, with the $L_{\text{effective}}$ of the respective body-part regions, and with the $L_{\text{effective}}$ of the respective slice positions are shown in Fig. 4b and c, respectively. The conversion functions by the effective sizes considerably reduced the values of $\sqrt{\delta_S^2}$ and $\delta_S$, especially in the head and neck region. Figure 3(1e)–(4e) (the bottom row images) show axial $\delta_S$ distributions by the $H$-$\langle S/S_t \rangle$ conversion functions with $L_{\text{effective}}$ determined for the slice positions. The $\sigma_{\text{effective}}$ quantifying the slice-by-slice variation of $L_{\text{effective}}$ is also shown in Table 1. The x-ray attenuation in the lung tissues was insignificant due to their low mass density, while the attenuation in the blade bones was significant. The thorax region contained both of these tissues, inducing a large $\sigma_{\text{effective}}$, i.e. $\geq 6$ cm, in the region as shown in Fig. 4d. Figure 6 shows the $\delta_S$ within the whole body and the body-part regions of the male and female phantoms, and their average. The variations of the average $\delta_S$ for 6 cm $< L < 50$ cm were $\leq 0.004$ for all of the body-part regions except for the head and neck region. The fraction of the bone tissues was high in the head and neck region, which resulted in the overestimation of SPR for large $L$, e.g. $\delta_S = 0.011$ for $L = 50$ cm. In other regions, the overestimation of SPR in bone tissues was compensated for by the underestimation in adipose tissues, resulting in a moderate value of $\delta_S$. The effective size deviations due to variation in body size, $\Delta L$, and due to the variation in initial energy spectrum of the CT scanner, $\Delta L_{110}$ and $\Delta L_{130}$ are shown in Table 1. The deviation $\Delta L$ in the thorax region was $> 4$ cm due to the significant differences in shoulder width and volume of the breasts between male and female models. In contrast, the deviations $\Delta L_{110}$ and $\Delta L_{130}$ were insignificant compared with $\Delta L$.

DISCUSSION

The effective size of the patient cross section for minimizing the uncertainty in patient SPR estimation was determined for charged-particle therapy treatment planning using human-tissue computational phantoms provided by ICRP. The effective size was 21.0 cm for the whole body. Using the determined effective size, we were able to reduce the root mean square deviation of SPR $\delta S$ to $\leq 0.011$, and the mean deviation $\overline{\delta_S}$ to $\leq 0.001$ in the whole body.

In the human body, we found that the overestimation (or underestimation) of the SPR in bone tissues is often compensated for by the underestimation (or overestimation) in adipose tissues. The absolute value of $\overline{\delta_S}$ realized by $L_{\text{effective}} = 21.0$ cm was in fact one order of magnitude smaller than the corresponding value of $\delta_S$. The absolute value of $\overline{\delta_S}$ was $< 0.001$ in the pelvic region, while it was $< 0.004$ in the head and neck region, with $L_{\text{effective}} = 21.0$ cm. These results indicate, for instance, that for a proton beam with 25-cm water equivalent length (WEL) range in the pelvic region and a proton beam with 12-cm WEL range in the head and neck region, the expected mean range errors are $< 0.03$ and $< 0.05$ cm WEL in their regions, respectively. Schaffner and Pedroni [9] reported larger range errors of 0.33 cm WEL in a prostate patient case and 0.14 cm WEL in a brain patient case for proton beams with similar ranges, i.e. 25 and 12 cm WEL, where they derived the range errors by linearly adding the expected range uncertainties of soft tissue and bone, corresponding to the calculation of $\sqrt{\delta_S^2}$, for their patient cases.

The slice-by-slice variation in $L_{\text{effective}}$ in each body-part region quantified by $\sigma_{\text{effective}}$ was larger than the deviations $\Delta L$, $\Delta L_{110}$ and $\Delta L_{130}$ as shown in Table 1. Further, $\sigma_{\text{effective}}$ was larger than the variation in $L_{\text{effective}}$ among the body-part regions. These results suggest that the preparation of the $H$-$\langle S/S_t \rangle$ conversion functions according to the body-part regions is unnecessary for practical purposes. For simplicity, an x-ray calibration phantom with a fixed size of $L = 21.0$ cm should rather be used to produce the $H$-$\langle S/S_t \rangle$ conversion function, irrespective of the body-part regions, by which $\sqrt{\delta_S^2}$ and $\delta_S$ are reasonably reduced in all body-part regions, as shown in Fig. 4. At the beginning of this study, we expected that the impact of the beam hardening on the uncertainty of SPR estimation in a patient can be minimized by selecting the optimum phantom sizes for calibrating $H$-$\langle S/S_t \rangle$ conversion functions. However, even with the conversion function derived from the optimum calibration for each slice, the predicted SPRs deviated from the reference SPRs. This may be the intrinsic limitation of a single-energy CT with polychromatic x-rays. The $H$-$\langle S/S_t \rangle$ conversion functions constructed with the selected representative tissues specifically for the body-part regions may potentially reduce the deviations. The beam-hardening correction based on the phantoms with more realistic compositions and configurations may also reduce the deviations. A dual-energy CT for patient SPR estimation may be another method for reducing the deviations [17–19].

There are several sources of uncertainty in patient SPR estimation, and one of them is insufficient correction of beam hardening arising from the mismatch between the size of the object and the calibration phantoms investigated in this study. To fully exploit the advantages of charged particle therapy, the remaining uncertainty sources should also be minimized.

CONCLUSION

One of the uncertainty sources in patient SPR estimation in charged particle therapy treatment planning is insufficient correction of the
beam-hardening effect arising from the mismatch between the size of the object and the calibration phantom used. The uncertainty would be minimized by selecting a suitably sized x-ray calibration phantom in construction of the $H$-to-$\langle S/S_w \rangle$ conversion function, namely the effective size of the patient cross section. We determined the effective size using an ideal CT system and realistic

Fig. 6. Mean deviation of SPR, $\delta S$, for different phantom sizes $L$ in (a) the whole body, (b) head and neck, (c) thorax, (d) abdominal and (e) pelvic regions of the male phantom (blue curves), female phantom (red curves), and their average (black curves). The solid, dashed and dotted curves are $\delta S$ for an x-ray tube voltage of 120 kVp, 110 kVp and 130 kVp, respectively.
human tissue computational phantoms provided by ICRP. The effective size was 21.0 cm, with which the root mean square deviation of SPRs $\delta_S^2$ and the mean deviation of SPRs $\delta_S$ could be reduced to $\leq 0.011$ and $\leq 0.001$, respectively, in the whole body. The effective patient-cross-section size of 21.0 cm can be used for the whole patient body, irrespective of body-part regions, for treatment planning in clinical practice.

CONFLICT OF INTEREST
The authors declare that there are no conflicts of interest.

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REFERENCES
1. Kanematsu N, Inaniwa T, Nakao M. Modeling of body tissues for Monte Carlo simulation of radiotherapy treatment planned with conventional x-ray CT systems. Phys Med Biol 2016;61:5037–50.
2. Schneider U, Pedroni E, Lomax A. The calibration of CT Hounsfield units for radiotherapy treatment planning. Phys Med Biol 1996;41:111–24.
3. Kanematsu N, Matsufuji N, Kohno R et al. A CT calibration method based on the polybinary tissue model for radiotherapy treatment planning. Phys Med Biol 2003;48:1053–64.
4. Moyers MF, Miller DW, Bush DA et al. Methodologies and tools for proton beam design for lung tumors. Int J Radiat Oncol Biol Phys 2001;49:1429–38.
5. Moyers MF, Sardesai M, Sun S et al. Ion stopping powers and CT numbers. Med Dosim 2009;35:179–94.
6. Yang M, Zhu RX, Park PC et al. Comprehensive analysis of proton range uncertainties related to patient stopping-power-ratio estimation using the stoichiometric calibration. Phys Med Biol 2012;57:4095–115.
7. Constantinou C, Harrington JC, Dewerd LA. An electron density calibration phantom for CT-based treatment planning computers. Med Phys 1992;19:325–7.
8. Jakel O, Jacob C, Schardt D et al. Relation between carbon ion ranges and x-ray CT numbers. Med Phys 2001;28:701–3.
9. Schaffner B, Pedroni E. The precision of proton range calculations in proton radiotherapy treatment planning: experimental verification of the relation between CT-HU and proton stopping power. Phys Med Biol 1998;43:1579–92.
10. Poludniowski G, Landry G, DeBlois F et al. Spekcalc: a program to calculate photon spectra from tungsten anode x-ray tubes. Phys Med Biol 2009;54:N433–8.
11. Brooks RA, Di Chiro G. Beam hardening in x-ray reconstructive tomography. Phys Med Biol 1976;21:390–8.
12. ICRP. Adult reference computational phantoms. Publication 110. International Commission on Radiological Protection, Ottawa, 2009.
13. Tsujii H, Kamada T. A review of update clinical results of carbon ion radiotherapy. Jpn J Clin Oncol 2012;42:670–85.
14. ICRU. Stopping powers of electrons and positrons. ICRU Report 37. ICRU, Bethesda, MD, 1984.
15. Kanematsu N, Inaniwa T, Koba Y. Relationship between electron density and effective densities of body tissues for stopping, scattering, and nuclear interactions of proton and ion beams. Med Phys 2012;39:1016–20.
16. Inaniwa T, Kanematsu N. Effective particle energies for stopping power calculation in radiotherapy treatment planning with protons and helium, carbon, and oxygen ions. Phys Med Biol 2016;61:N542–50.
17. Saito M, Tsukihara M. Technical note: exploring the limit for the conversion of energy-subtracted CT number to electron density for high-atomic-number materials. Med Phys 2014;41:071701.
18. Yang M, Virshup G, Clayton J et al. Theoretical variance analysis of single- and double-energy computed tomography methods for measuring proton stopping power ratios of biological tissues. Phys Med Biol 2010;55:1343–62.
19. Hümemohr N, Krauss B, Tremmel C et al. Experimental verification of ion stopping power prediction from dual energy CT data in tissue surrogates. Phys Med Biol 2014;59:83–96.