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Study on Methods for Obtaining Relative Stopping Power of Charged Particles from Dual-Energy CT Images

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Abstract

In proton and heavy ion radiotherapy, acquiring the relative stopping power of each voxel in a patient through CT scanning is critical for the treatment planning system to accurately calculate the dose distribution within the body. This study performed two CT scans on tissue-equivalent phantoms using conventional CT at different tube voltages. The CT values from the two image sets were utilized to calculate the phantom's relative electron density and effective atomic number, which were then used to determine the phantom's relative stopping power. The results indicate that the relative stopping power of tissue-equivalent phantoms predicted using dual CT values outperforms that obtained through conventional methods based on a single image set. Consequently, the proposed method plays a significant role in further enhancing the accuracy of dose calculation in proton and heavy ion radiotherapy treatment planning.

Full Text

Preamble

Acquisition of Charged Particle Relative Stopping Power Based on Dual-Tube Voltage CT Images

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Abstract

In proton and heavy ion radiotherapy, accurately calculating the dose distribution within the body using a treatment planning system depends critically on obtaining the relative stopping power for each voxel through CT scanning of the patient. This study employed a conventional CT scanner set to different tube voltages to perform two CT scans of a tissue-equivalent phantom. The relative electron density and effective atomic number of the phantom were calculated from the CT values derived from the two image sets, which were then used to compute the phantom's relative stopping power. The results demonstrate that predicting the relative stopping power of tissue-equivalent phantoms using dual CT values yields superior accuracy compared to conventional methods based on a single set of images. Consequently, the method established in this study holds significant potential for further improving the accuracy of dose calculations in proton and heavy ion radiotherapy treatment planning.

Keywords: radiotherapy; relative electron density; effective atomic number; CT scanner; relative stopping power

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Introduction

Radiotherapy has evolved for over a century as a treatment modality for malignant tumors. The advent of proton and heavy ion therapy has ushered modern radiotherapy into a new era of development [**?**]. When proton or heavy ion beams are extracted from an accelerator and enter the human body, they release minimal energy before reaching the tumor lesion. However, upon arrival at the lesion, the beams deposit a massive amount of energy instantaneously, after which the beam energy decays rapidly. This sharp peak formed during the process is known as the Bragg Peak. During treatment, appropriate beam energy can be selected based on tumor depth, and the Bragg peak can be spread out (Spread-out Bragg Peak; SOBP) [**?**] to cover the tumor volume. Consequently,

proton and heavy ion therapy can increase the dose to the tumor region while reducing the dose to normal tissues along the entrance path, thereby decreasing radiation toxicity and maximizing therapeutic efficacy.

Proton and heavy ion treatment planning systems utilize computed tomography (CT) as fundamental input data, converting CT values (Hounsfield Units, HU) into the particle's stopping power ratio relative to water (Stopping Power Ratio, SPR) or water-equivalent path lengths (WEPL) to calculate particle range, energy deposition, and dose distribution within the medium [**?**, **?**]. This conversion process relies on CT number-to-SPR or WEPL calibration curves. After converting each pixel of the CT image to WEPL, the dose calculated in water can be mapped onto the CT image. The dose for each pixel is obtained by integrating the dose distribution curve in water using each pixel's WEPL as the integration interval and then dividing by the corresponding pixel WEPL [**?**, **?**]. The accuracy of the calibration curve directly affects the precision of stopping power determination for human tissues and organs, thereby influencing the dose distribution [**?**]. Two common methods exist for obtaining CT number-to-relative stopping power calibration curves: direct measurement using animal organs or tissues [**?**], or measurement using tissue-equivalent phantoms [**?**, **?**]. These methods introduce range uncertainties of 3%–5%, necessitating target margin expansion in treatment planning to mitigate their effects [**?**].

The energy stopping power of charged particles passing through a medium can be described by the Bethe-Bloch formula. In Schneider et al.'s study [**?**], the medium's SPR was approximated by Equation (1), with higher-order infinitesimals omitted:

$$SPR \approx \rho_e \cdot \frac{2m_{ec}^2\beta^2 I(1-\beta^2)}{2m_{ec}^2\beta^2 I_w(1-\beta^2)}$$

where ρ_e is the electron density of the medium relative to water, β is the relativistic velocity ratio of the charged particle, m_{ec}^2 is the electron rest energy, I is the mean excitation energy of the medium material, and the mean excitation energy of water is $I_w = 75\text{eV}$ [**?**].

As shown in Equation (1), SPR is not only proportional to ρ_e but also correlated with the medium's I value. The dependence of SPR on β is very small, below 0.4% [**?**], and this effect is neglected in the present study. Based on the energy range of the carbon ion therapy system developed by the Institute of Modern Physics, Chinese Academy of Sciences (120 MeV/u–400 MeV/u) [**?**], the median energy of 260 MeV/u was uniformly adopted for calculations to ensure minimal deviation of the final results within the therapeutic energy range.

In Hünemohr's research [**?**], dual-energy CT (DECT) enables extraction of an additional parameter, thereby resolving uncertainties in SPR arising from the I value. CT values of the same phantom obtained at two different energies allow definition of the effective atomic number Z_{eff} for each voxel in addition to ρ_e

[**?**], providing more accurate information to distinguish tissues with similar ρ_e and thus enabling more precise SPR calculation.

This study utilized nine tissue-equivalent phantom inserts with known parameters from a CT electron density phantom. Based on the DECT dual-parameter extraction method reported by Hünemohr et al. [**?**], a conventional single-energy CT (SECT) was employed with different tube voltages to scan tissue-equivalent phantoms, extracting ρ_e and Z_{eff} and calculating SPR for comparison with traditional methods, aiming to obtain more accurate SPR values through this approach.

2.1 Equipment and Materials

A German Siemens positioning CT scanner (Model: SOMATOM Sensation Open) was used, along with an electron density phantom (Model: CIRS 062M) manufactured by CIRS Inc. (www.cirsinc.com) and its nine tissue-equivalent phantom inserts. The parameters are listed in Table 1 .

Table 1: CIRS 062M Material Parameter List
Tissue-equivalent material parameters for bone trabeculae (200 mg/cc), bone cortical (800 mg/cc), and bone cortical (1250 mg/cc)

2.2 Phantom Scanning

The tissue-equivalent phantom inserts were placed in their designated slots and scanned using the positioning CT scanner. Four scan sequences were established with tube voltages set to 80 kVp, 100 kVp, 120 kVp, and 140 kVp, respectively. Tube current was uniformly set to 320 mAs, slice thickness to 3 mm, and image reconstruction thickness to 1.5 mm. The scanning schematic is shown in Figure 1 [Figure 1: see original paper].

Using the circular selection tool in the RadiAnt DICOM Viewer medical imaging software, the mean CT values and standard deviations of the middle layer of each tissue-equivalent phantom were measured across the four image sequences. The results are presented in Table 2 . The spatial positions of the measured regions were identical across all four sequences.

Figure 1. Phantom scanning schematic diagram

Table 2. Mean CT values (HU) and standard deviations of tissue-equivalent phantoms

Tissue-equivalent material	80 kVp	100 kVp	120 kVp	140 kVp
Lung (inhale)	-			
	816.49$±6.81|−			
	817.69±6.15|−			
	818.12±5.81|−			
	817.19±5.68||*Lung(exhale)*|−			
	499.01±6.80|−			
	500.03±3.78|−			
	500.33±3.28|−			
	500.53±3.29||*Adipose*|−			
	97.53±5.99|−			
	85.12±3.77|−			
	76.68±3.06|−			
	72.72±2.74||*Breast*|−			
	47.89±5.34|−			
	40.09±3.46|−			
	34.33±2.86|−			
	31.62±2.42||*Solidwater*|45.32±7.16|44.59±5.18|45.16±4.26|43.57±4.01||*Brain*			

2.3 Parameter Extraction Method

In Hünemohr's study [**?**], the X-ray linear attenuation coefficient μ_E of human tissues was approximated as a decomposition into a virtual pure photoelectric absorption material and a virtual pure Compton scattering material:

$$\mu_E \approx A\rho_e f_E + B\rho_e E^3 \quad (2)$$

where n is the exponent correlating photoelectric effect cross-section with atomic number, f_E is a function approximately flat with respect to X-ray energy and can be treated as constant in this study, A and B are scaling factors nearly independent of material, Z is material atomic number, and E is X-ray energy. Equation (2) is also applicable to the continuous X-ray spectrum of CT.

For CT's continuous X-ray spectrum, the average linear attenuation coefficient $\bar{\mu_E}$ of a medium relates to CT image HU values as follows:

$$HU = \left(\frac{\bar{\mu_E} - \bar{\mu_{E,w}}}{\bar{\mu_{E,w}}}\right) \times 1000 \quad (3)$$

where $\bar{\mu_{E,w}}$ is the average linear attenuation coefficient of water. In subsequent calculations, the ratio of material linear attenuation coefficient to water's $\bar{\mu_E}/\bar{\mu_{E,w}}$ is denoted as μ.

Scanning tissue-equivalent materials with high (H) and low (L) energy yields the following expressions:

$$\mu_{E_L} \approx A\rho_e f_{E_L} + B\rho_e$$
$$\mu_{E_H} \approx A\rho_e f_{E_H} + B\rho_e \quad (4)$$

Combining with Equation (3) and determining parameters A and B, Equations (3) and (4) can be solved simultaneously for ρ_e and Z_{eff}. Substituting known data yields the following results:

$$\rho_e = C \times (+1) + (1 - C) \times ($$
$$Z_{eff} = \left(\frac{1}{n} - D\right) \times (+1) + (Z_w \quad (5)$$

As shown in Equation (4), parameters C and D can be determined through dual-energy scanning of one known tissue-equivalent phantom. For the atomic number range $Z = 1 - 20$ and the CT spectra used, the optimal coefficient is $n = 3.1$ [**?**].

In this study, CT image data from all nine materials were utilized, with high energy uniformly set to 140 kVp and low energy set to 80 kVp and 100 kVp, respectively, denoted as 80/140 and 100/140 groups. Parameters C and D in Equation (5) were fitted, with results shown in Table 3 .

Table 3. Fitted parameters C and D

Group	C	D
80/140	[value]	[value]
100/140	[value]	[value]

Note: The original table values were not provided in the text.

Substituting the measurement results from Table 2 into Equation (5) yields ρ_e and Z_{eff} values, which were compared with known reference data from Table 1. The results are presented in Table 4 .

Table 4. Extracted ρ_e and Z_{eff} results and deviations for tissue-equivalent phantoms

Tissue-equivalent material	ρ_e calculated	Z_{eff} calculated	ρ_e deviation	Z_{eff} deviation
Lung (inhale)	0.182$±0.012(80/140), 0.184±0.018(100/140)|7.557±1.166(80/140), 7.295±2.438(100/140)| -0.77% 0.88% 9.00±0.008(80/140), 0.999±0.01(100/140)|7.535±0.376(80/140), 7.506±0.497(100/140)|0.4 (80/140), (80/140), 0.11±0.006(80/140), 0.983±0.009(100/140)|6.948±0.174(80/140), 6.932±0.250(100/140)|0.6 -1.06% 0.14% 0.10±0.009(80/140), 1.051±0.012(100/140)|7.541±0.196(80/140), 7.549±0.265(100/140)|− (100/140) (100/140) 0.10±0.017(80/140), 1.141±0.025(100/140)|9.665±0.195(80/140), 9.527±0.353(100/140)|1.5 (100/140)			

2.4 Calculation of ln(I) Values

The $\ln(I)$ value in Equation (1) is estimated using the Z_{eff} extracted through the above method [**?**], with the following relationship to Z_{eff}:

$$\ln(I) = 0.125Z_{eff} + 3.379 \quad (Z_{eff} \leq 8.5)$$
$$\ln(I) = 0.098Z_{eff} + 3.376 \quad (Z_{eff} > 8.5)$$

3.1 SPR Calculation

Based on the calculation results from Table 4 and Equation (1), the relative stopping power of materials was computed and compared with reference values, as shown in Table 5 . The SPR reference values were not directly provided in the material specifications; the values in the table were calculated from the ρ_e and Z_{eff} values supplied in Table 1 using Equation (1).

Table 5. SPR measurement results and errors

Tissue-equivalent material	SPR reference	SPR measured (DECT80/140)	SPR measured (DECT100/140)	DECT 80/140 error	DECT 100/140 error
Lung (inhale)	0.182$±0.068|0.185±0.103|0.79%			-1.13%	
	| –				
	9.49±0.017|0.499±0.023|–				
	|0.19±0.008|0.962±0.011|–				
	| –				
	0.37±0.007|0.991±0.010|–				
	|0.32±0.010|1.042±0.014|–				
	| –				
	0.24±0.009|1.051±0.012|–				
	| –				
	0.30±0.015|1.142±0.022|–				
	|2.59±0.015|1.433±0.022|–				
	|1.37±0.016|1.602±$0.023				

3.2 Comparison with SECT Single-Parameter Method

Using the 120 kVp scanning parameter and the traditional SECT single-parameter extraction method [**?**], the HU-SPR curve was fitted (Figure 2 [Figure 2: see original paper]). The SPR values and errors of tissue-equivalent phantoms were then calculated through this curve and compared with the errors from the dual-parameter extraction method in Table 5 (Table 6 , Figure 3 [Figure 3: see original paper]).

Figure 2. Fitted HU-SPR curve

Table 6. Comparison of SPR measurement errors between DECT and SECT methods

Tissue-equivalent material	DECT80/140	DECT100/140	SECT120
Lung (inhale)	-9.49%	-8.15%	-0.50%
Lung (exhale)	0.19%	0.43%	4.02%
Adipose	-0.37%	-0.31%	-2.49%
Breast	0.32%	0.45%	-0.41%
Solid water	-0.24%	-0.25%	1.72%
Brain	-0.30%	-0.31%	-1.23%
Bone trabeculae (200 mg/cc)	2.59%	3.38%	1.45%
Bone cortical (800 mg/cc)	1.37%	1.94%	0.50%
Bone cortical (1250 mg/cc)	-0.79%	-1.13%	0.19%

Figure 3. Comparison of SPR measurement deviations between DECT and SECT methods

Discussion

Proton and heavy ion radiotherapy, as a precision radiotherapy modality, has attracted widespread attention in the radiotherapy community in recent years. The CT number-to-relative stopping power conversion curve plays a crucial role in the accuracy of proton and heavy ion radiotherapy treatment planning [**?**]. This study employed a conventional positioning CT scanner with different tube voltages to generate X-rays of different energies for scanning tissue-equivalent phantoms. Dual parameters ρ_e and Z_{eff} were extracted from two sets of CT values obtained from high- and low-energy scans, and the medium's relative stopping power was derived from these dual parameters.

The parameters extracted using the method in this study (Table 4) show that for ρ_e deviation, lung (inhale) reached -9.00% and -8.00% in the DECT80/140 and DECT100/140 groups, respectively, while deviations for other tissues were around 1%. For Z_{eff} deviation, bone trabeculae (200 mg/cc) showed errors of -8.39% and -9.70% in the two groups, respectively, while other deviations were less than 3%. Witt et al. [**?**] compared theoretically calculated data with actual measured data and found differences of 5%–15% in lung-equivalent models, attributing this primarily to the significant uncertainty in the manufacturing process of porous materials [**?**]. Therefore, the large deviations observed for lung (inhale) and bone trabeculae in this study may also result from uncertainties in the manufacturing process of porous materials. In Schneider's study [**?**], a 10% error in I value caused less than 1.5% deviation in SPR prediction. Since $\ln(I)$ has a linear relationship with Z_{eff} [**?**], the deviation in Z_{eff} also results in relatively small SPR prediction errors.

The SPR calculation results in Table 5 also show larger deviations for lung (inhale) and bone trabeculae. Excluding these anomalous data, the average deviation for the remaining data is 0.03% in the 80/140 group and 0.12% in the 100/140 group. These results are superior to those obtained using the conventional SECT single-image method (Figure 3). Therefore, based on these results, the method established in this study can predict SPR more accurately and is clinically feasible, though further correction of relative stopping power for low-density or porous tissues is necessary.

Moreover, this study used a fixed phantom for two CT scans, ensuring absolute spatial matching between the two image sets. However, for human patients, organ motion during the two scans introduces spatial mismatch between images. Consequently, this method is only applicable for proton and heavy ion radiotherapy of sites with high positional reproducibility over short periods (between the two CT scans), such as the head or extremities fixed with high-precision positioning devices. For other sites, either DECT scanning or deformable registration [**?**] must be employed to achieve spatial matching between the two image sets.

Analysis of Equation (5) in this paper reveals that differences in calculated ρ_e and Z_{eff} between energy groups are determined first by the high-energy

HU_H value, which establishes the baseline magnitude of ρ_e and Z_{eff}, and then adjusted by the difference between high- and low-energy CT values ($HU_L - HU_H$). Since HU_H values in different group calculations all adopt the HU value at the highest energy of 140 kVp, the primary factor determining final result accuracy is the difference between high- and low-energy CT values. Therefore, when selecting this method for SPR prediction, it is essential to maximize the CT value difference ($HU_L - HU_H$) to improve prediction accuracy, making it more appropriate to select tube voltage settings with larger energy differences.

The SPR reference data in this study were derived from the CIRS 062M product specifications, which themselves have certain deviations due to manufacturing uncertainties. Therefore, the SPR data obtained in this study require further calibration through actual clinical measurements.

Furthermore, the fitted parameter data in this study are only applicable to the CT scanner used in the experiments. For the same tissue-equivalent phantom, different CT devices, scanning parameters, phantom position, and surrounding materials can all affect CT values, leading to differences of up to 3% in SPR predicted by SECT methods [**?**]. Consequently, it is necessary to develop device-specific and parameter-specific dual-parameter extraction equations for the method proposed in this paper.

In summary, this study established a method to improve the accuracy of medium relative stopping power by obtaining dual CT values through setting different tube voltages on a conventional CT scanner. This approach provides an important reference value for further enhancing the accuracy of dose calculations in proton and heavy ion radiotherapy treatment planning.

References

[1] WANG Lan, DAI Xiaoya. Status and Prospect of Global Proton Heavy Ion Hospital [J]. *Hospital Construction and Equipment in China*, 2016 (1): 27-31. (in Chinese) (Wang Lan, Dai Xiaoya. Status and Prospect of Global Proton Heavy Ion Hospital[J]. China Hospital Construction and Equipment, 2016(1): 27–31.)

[2] CHUANG Y-C, WU P-H, SHEN Y-A, et al. Recent Advances in Metal-Based NanoEnhancers for Particle Therapy[J]. *Nanomaterials*, 2023, 13(6): 1011. doi: 10.3390/nano13061011

[3] LIU Xinguo, LI Qiang, DU Xiaogang, et al. Preliminary IMP Heavy Ion Therapy Planning System [J]. *Nuclear Physics Review*, 2010, 27(4): 480-487. (in Chinese) (Liu Xinguo, Li Qiang, Du Xiaogang, et al. Preliminary IMP Heavy Ion Therapy Planning System[J]. Nuclear Physics Review, 2010, 27(4): 480–487.)

[4] 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[J]. *Physics in Medicine and Biology*, 1998, 43(6): 1579–1592. doi: 10.1088/0031-9155/43/6/016

[5] LIU Xinguo, LI Qiang, DAI Zhongying. Dose calculation method in IMP heavy ion therapy [J]. *Nuclear Physics Review*, 2009, 26(1): 69-75. (in Chinese) (Liu Xinguo, Li Qiang, Dai Zhongying. Dose calculation method in IMP heavy ion therapy[J]. Nuclear Physics Review, 2009, 26(1): 69–75.)

[6] MUSTAFA A A, JACKSON D F. The relation between X-ray CT numbers and charged particle stopping powers and its significance for radiotherapy treatment Planning[J]. *Physics in Medicine and Biology*, 1983, 28(2): 169–176. doi: 10.1088/0031-9155/28/2/006

[7] RIETZEL E, SCHARDT D, HABERER T. Range accuracy in carbon ion treatment planning based on CT-calibration with real tissue Samples[J]. *Radiation Oncology*, 2007, 2(1): 14. doi: 10.1186/1748-717X-2-14

[8] SZYMANOWSKI H, OELFKE U. CT calibration for Two-dimensional scaling of proton pencil Beams[J]. *Physics in Medicine and Biology*, 2003, 48(7): 861–874. doi: 10.1088/0031-9155/48/7/304

[9] COOLENS C, CHILDS P J. Calibration of CT Hounsfield units for radiotherapy treatment planning of patients with metallic hip prostheses: The use of the extended CT-Scale[J]. *Physics in Medicine and Biology*, 2003, 48(11): 1591–1603. doi: 10.1088/0031-9155/48/11/308

[10] YANG M, WOHLFAHRT P, SHEN C, et al. Dual- and multi-energy CT for particle stopping-power estimation: Current state, challenges and Potential[J]. *Physics in Medicine & Biology*, 2023, 68(4): 04TR01. doi: 10.1088/1361-6560/acabfa

[11] SCHNEIDER W, BORTFELD T, SCHLEGEL W. Correlation between CT numbers and tissue parameters needed for Monte Carlo simulations of clinical dose Distributions[J]. *Physics in Medicine and Biology*, 2000, 45(2): 459–478. doi: 10.1088/0031-9155/45/2/314

[12] DEASY J. ICRU Report 49, Stopping Powers and Ranges for Protons and Alph Particles[J]. *Medical Physics*, 1994, 21(5): 709–710. doi: 10.1118/1.597176

[13] YANG M, ZHU X R, PARK P C, et al. Comprehensive analysis of proton range uncertainties related to patient Stopping-power-ratio estimation using the stoichiometric Calibration[J]. *Physics in Medicine and Biology*, 2012, 57(13): 4095–4115. doi: 10.1088/0031-9155/57/13/4095

[14] MENG W-B, LI Q, YE Y-C, et al. Determination of virtual source position using back projecting zero and convergent arcTAN method for Scanning-passive scatter beam in carbon ion Therapy[J]. *Journal of X-Ray Science and Technology*, 2023, 31(1): 181–197. doi: 10.3233/XST-221274

[15] HÜNEMOHR N, KRAUSS B, TREMMEL C, et al. Experimental verification of ion stopping power prediction from dual energy CT data in tissue Surrogates[J]. *Physics in Medicine and Biology*, 2014, 59(1): 83–96. doi: 10.1088/0031-9155/59/1/83

[16] MURAISHI H, NISHIMURA K, ABE S, et al. Evaluation of Spatial Resolution for Heavy Ion CT System Based on the Measurement of Residual Range Distribution With HIMAC[J]. *IEEE Transactions on Nuclear Science*, 2009, 56(5): 2714–2721. doi: 10.1109/TNS.2009.2023520

[17] MENG Wanbin, MA Xiaoyun, ZHANG Yanshan, et al. Measurement of the CT-WEPL calibration curve in the carbon ion treatment planning system [J]. *Modern Medicine and Health*, 2022, 38(6): 920-924. (in Chinese) (Meng Wanbin, Ma Xiaoyun, Zhang Yanshan, et al. Measurement of the CT-WEPL calibration curve in the carbon ion treatment planning system[J]. Modern Medicine and Health, 2022, 38(6): 920–924.)

[18] WITT M, WEBER U, KELLNER D, et al. Optimization of the Stopping-power-ratio to Hounsfield-value calibration curve in proton and heavy ion Therapy[J]. *Zeitschrift für Medizinische Physik*, 2015, 25(3): 251–263. doi: 10.1016/j.zemedi.2014.11.001

[19] URIE M, GOITEIN M, HOLLEY W R, et al. Degradation of the Bragg peak due to Inhomogeneities[J]. *Physics in Medicine and Biology*, 1986, 31(1): 1–15. doi: 10.1088/0031-9155/31/1/001

[20] LAO Zheng, FAN Wenhui, BI Fang, et al. Comparative study on deformation registration accuracy of medical image processing MIM software and Velocity software in elastic phantom [J]. *China Medical Equipment*, 2021, 18(6): 47-51. (in Chinese) (Lao Zheng, Fan Wenhui, Bi Fang, et al. Comparative study on deformation registration accuracy of medical image processing MIM software and Velocity software in elastic phantom[J]. China Medical Equipment, 2021, 18(6): 47–51.)

[21] SCHNEIDER U, PEDRONI E, LOMAX A. The calibration of CT Hounsfield units for radiotherapy treatment Planning[J]. *Physics in Medicine and Biology*, 1996, 41(1): 111–124. doi: 10.1088/0031-9155/41/1/009

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