Fusion Visualization Technique to Improve a Three-Dimensional Isotope-Selective CT Image Based on Nuclear Resonance Fluorescence with a Gamma-CT Image

Khaled Ali 1,2,* , Heishun Zen 1, Hideaki Ohgaki 1, Toshiteru Kii 1, Takehito Hayakawa 3,4, Yoshitaka Taïra 5, Masahiro Katoh 6, Toshiteru Kii 7, Masaki Fujimoto 5 and Hiroyuki Toyokawa 7

1 Institute of Advanced Energy (IAE), Kyoto University, Gokasho, Uji, Kyoto 611-0011, Japan; zen@iae.kyoto-u.ac.jp (H.Z.); ohgaki.hideaki.2w@kyoto-u.ac.jp (H.O.); kii@iae.kyoto-u.ac.jp (T.K.)
2 Physics Department, Faculty of Science, South Valley University, Qena 83523, Egypt
3 Tokai Quantum Beam Science Center, National Institutes for Quantum Science and Technology (QST), Ibaraki 319-1106, Japan; hayakawa.takehito@qst.go.jp (T.H.); shizuma.toshiyuki@qst.go.jp (T.S.)
4 Institute of Laser Engineering, Osaka University, Suita, Osaka 565-0871, Japan
5 UVSOR Synchrotron Facility, Institute for Molecular Science, National Institutes of Natural Sciences, Okazaki 444-8585, Japan; mkatoh@ims.ac.jp (M.K.); yostaira@ims.ac.jp (Y.T.); mfamoto@ims.ac.jp (M.F.)
6 Hiroshima Synchrotron Radiation Center, Hiroshima University, Higashi-Hiroshima 739-0046, Japan
7 Institute of Advanced Industrial Science and Technology (AIST), Tsukuba Central 2-4, Ibaraki 305-8568, Japan; h.toyokawa@aist.go.jp
* Correspondence: khaled.ali.28e@st.kyoto-u.ac.jp or khaled.ali@sci.svu.edu.eg; Tel.: +81-774-38-3425; Fax: +81-774-38-3426

Abstract: One of the most noteworthy aspects of computed tomography (CT) based on the nuclear resonance fluorescence (NRF) transmission method is the isotope selectivity that makes it possible to discern an isotope of interest from other isotopes within a sample. We experimentally obtained a three-dimensional (3D) isotope-selective CT image based on the NRF transmission method (3D NRF-CT) for the enriched lead isotope distribution of 208Pb in a cylindrical holder in a previous study. The cylindrical holder’s diameter and height are 25 mm and 20 mm, respectively. The NRF-CT imaging technique requires a considerable data accumulation time. It took 48 h to obtain an image with a resolution of 4 mm/pixel in the horizontal plane and 8 mm/pixel in the vertical plane using a laser Compton scattering (LCS) gamma-ray beam with a beam size of 2 mm and a flux density of 10 photons/s/eV. Improving the NRF-CT image resolution with the existing hardware is challenging. Therefore, we proposed an alternative method to improve the NRF-CT image resolution using the fusion visualization (FV) technique by combining the NRF-CT image including isotopic information with a gamma-CT image, which provides better pixel resolution. The 3D gamma-CT image for the same sample was measured at the same beamline BL1U in the ultraviolet synchrotron orbital radiation-III (UVSOR-III) synchrotron radiation facility at the Institute of Molecular Science at the National Institutes of Natural Sciences in Japan under similar experimental conditions except for the LCS gamma-ray beam flux and beam size. Obtaining a 3D gamma-CT image with a resolution of 1 mm/pixel took 5 h using an LCS gamma-ray beam with a beam size of 1 mm and a flux density of 0.7 photons/s/eV. The data processing of the FV technique has been developed, and the 3D NRF-CT image quality was improved.

Keywords: computed tomography (CT); fusion visualization (FV); isotope selectivity; laser Compton scattering gamma-ray beam (LCS); nuclear resonance fluorescence (NRF)

1. Introduction

Nuclear safeguards have a critical role in the non-proliferation of nuclear materials for reprocessing nuclear fuels and the control of nuclear materials. Nuclear resonance fluorescence (NRF) analysis [1] is a promising technology for nuclear safeguards and...
other nuclear applications [2–4]. Non-destructive inspection (NDI) for isotopes based upon NRF has recently attracted a lot of attention because of its capacity to identify shielded isotope substances such as fissile material content in spent fuel and commercial cargo [2–7]. Because the NRF energy is specific to each nuclide, a nuclide of interest can be characterized by measuring the absorption of the incident gamma rays at the NRF energy or of the scattered gamma rays with an energy identical to the NRF energy. Energy-tunable quasi-monochromatic gamma rays generated by laser Compton scattering (LCS) [8,9] are suitable for the NRF measurements. Furthermore, computed tomography (CT) based on the NRF transmission method [10,11] with LCS gamma rays has been proposed as a novel technology for imaging an isotope of interest within massive materials [11]. The NRF-CT imaging technique makes it possible to distinguish an isotope of interest from other isotopes within the same element in a sample. Therefore, the isotope selectivity is one of its strengths. Zen et al. [12] experimentally demonstrated the NRF-CT using an LCS gamma-ray beam. They obtained a two-dimensional (2D) NRF-CT image with a resolution of 5 mm/pixel for the $^{208}$Pb isotope in a natural lead rod hidden inside an iron cylinder holder, together with a set of different materials (aluminum, iron, and a vacant area).

In our previous study [13], we obtained a 2D isotope-selective CT image based on the NRF transmission method for the distribution of an enriched lead isotope ($^{208}$Pb) and of another enriched lead isotope ($^{206}$Pb) implied inside an aluminum cylinder holder. The 2D NRF-CT image had a resolution of 2 mm/pixel with a data acquisition time of 60 h [13]. As a result, $^{208}$Pb and $^{206}$Pb were clearly distinguished. In a subsequent study [14], we added another one-dimensional scan to obtain a three-dimensional (3D) NRF-CT image with horizontal and vertical resolutions of 4 and 8 mm/pixel, respectively. The data acquisition time was 48 h [14]. It is obvious that improving the resolution of NRF-CT images is crucial for realistic applications. To improve the image resolution while keeping a reasonable data acquisition time, upgrading the detection efficiency of the NRF measurement system and/or increasing the gamma-ray beam intensity are possible approaches. However, in practice, it is difficult to upgrade the detection system and/or the gamma-ray beams system of existing LCS facilities such as the ultraviolet synchrotron orbital radiation-III (UVSOR-III) synchrotron radiation facility at the Institute of Molecular Science at the National Institutes of Natural Sciences in Japan.

In the current study, we proposed an alternative approach that applied a numerical treatment, called the fusion visualization (FV) technique [15–18], to improve the NRF-CT image quality. In the case of the NDI for the hidden isotopes within an assembly volume, scientific visualization is representing complex raw data as 2D or 3D images to understand the shape of the hidden isotopes that may be overlooked by standard methods alone. The term “data fusion” refers to a numerical data processing technique that may give more consistent, accurate, and meaningful information by integrating multiple data sources. In the current study, we applied the FV technique to improve the quality of the 3D NRF-CT image that was obtained in our previous study [14]. The FV technique is based on the integration of various data sources. One of the sources is the primary 3D NRF-CT image [14], which supplies the desired isotope distribution, but its image quality needs improvement. Another data source is a high-resolution gamma-CT image for the same sample under similar experimental conditions measured at the beamline BL1U in the UVSOR-III facility but with some alterations in the LCS gamma-ray beam parameters. In this article, we describe the numerical treatment procedures of the FV technique for integrating the primary images of the 3D NRF-CT and the 3D gamma-CT to improve the quality of an isotope-selective 3D NRF-CT image. We compared a few different numerical treatments of the FV technique and discussed the quality of the obtained images.

In Section 2, we outline the experimental setup at the UVSOR synchrotron radiation facility’s beamline BL1U. We present a full description of the sample under investigation in Section 2.1. The experimental setup to measure the gamma-CT images is described in full in Section 2.2. We provide a brief explanation of the 3D NRF-CT image in Section 2.3, which was previously presented [14]. Section 3 provides the obtained results and the FV
numerical treatments. Section 3.1 illustrates the obtained 2D gamma-CT images in detail and presents how they were visualized to one 3D image. Additionally, a reference to the original 3D NRF-CT image that needs to be improved is provided. The post-multiply FV method between the original NRF-CT and gamma-CT images and its outcomes are shown in Section 3.2. The other possible methods of the FV between the original images followed by our recommendation are also presented. In Section 4, we give the conclusions of our research. This article incorporates sub-elementary Materials (Movie S1 and Movie S2).

2. Experimental Setup

The primary images of the 3D gamma-CT and the 3D NRF-CT were acquired using LCS gamma-ray beams at the beamline BL1U in the UVSOR-III synchrotron radiation facility. First, we measured the 3D gamma-CT image based on atomic absorption using a relatively low flux LCS gamma-ray beam. Second, high flux LCS gamma rays were used to measure the 3D NRF-CT image. An electron beam with an energy of 746 ± 1 MeV and an average current of 300 mA was stored in the top-up mode of the UVSOR-III electron storage ring. We generated LCS gamma-ray beams with a maximum energy of 5528 MeV by a head-on collision between the electrons and a laser beam provided by a Tm-fiber laser system (TLR-50-AC-Y14, IPG Laser GmbH, Köttinger Weg 188 Wissen, Germany), which has a lasing wavelength of 1.896 ± 2 µm (0.6539 eV), a spectral linewidth of 0.7 nm, a beam quality M² of 1.05, and a maximum average power of 50 W. The laser beam was randomly polarized and was operated in the continuous wave (CW) operation mode.

2.1. CT Sample Description

A cylindrical aluminum container with 20 mm in height and 25 mm in diameter was used as a specimen holder for the CT sample in both experiments. Three circular holes with identical diameters of 6.1 mm were drilled into the container body at a 120° pitch angle. A set of various cylindrical rods with diameters of 6 mm and heights of 6 mm were used to fill the holes as shown in Figure 1a with the following arrangements. (i) An iron (Fe) rod was inserted in the bottom hollow of the first hole (1), followed by an enriched lead isotope (208 Pb) rod, leaving the third hollow unfilled. (ii) The bottom of the second hole (2) was filled with an enriched lead isotope (206 Pb) rod, followed by an aluminum (Al) rod and a 208 Pb rod. (iii) A 208 Pb rod was placed in the bottom of the third hole (3), followed by an Fe rod and a 206 Pb rod. Figure 1b displays a picture of the CT sample. An automated scanning system implemented with LabVIEW software was built to scan the CT sample in three dimensions. The scanning system consisted of three components: (i) a three-axis moving stage that can move in three dimensions (vertical (y), horizontal (x), and rotational (θ)), (ii) a data analysis system, and (iii) a detector spectrum recording system which includes a multichannel analyzer (APG7400, Techno AP Co., Ltd., Mawatari, Hitachinaka-shi, Ibaraki, Japan).
2.2. 3D Gamma-CT Image Measurement

Figure 1 shows a schematic diagram of the experimental setup at the beamline BL1U in the UVSOR-III facility used to measure the 3D gamma-CT image. A lead collimator with a quadrangular prism shape that was 200 mm in length, 100 mm in width, and 100 mm in height, which had a circular hole with a diameter of 1 mm, was positioned in the LCS gamma-ray beam’s path to define the diameter and the energy spectrum of the LCS gamma-ray beam on the CT sample. We operated the laser system at a typical average power of 2.4 W during measurement of the gamma-CT images to generate an LCS gamma-ray beam with an intensity of $7 \times 10^6$ photons/s with 100% energy bandwidth at the full width at half maximum (FWHM) before collimation. The EGS5 Monte Carlo simulation code [19] was used to estimate the LCS gamma-ray beam’s properties after the collimation [13]. The estimated flux of the LCS gamma-ray beam after collimation was 0.7 photons/s/eV at a maximum energy of 5528 MeV with 1.1% energy bandwidth at the FWHM. Since the LCS gamma-ray beam’s divergence was small (on the order of $10^{-3}$), the diameter of the LCS beam on the CT sample was almost the same as the size of the hole in the collimator: approximately 1 mm. The flux of the incident LCS gamma-ray beam was measured using a 5-mm-thick plastic scintillator (PL) located 180 cm downstream from the collimator. After passing through the PL, the transmitted LCS gamma-ray beam was injected into the CT sample, which was installed on the three-axis traveling stage located 213 cm downstream from the collimator. The flux of the transmitted gamma rays was measured using a $3.5'' \times 4''$ LaBr$_3$(Ce) scintillation detector positioned 82 cm downstream from the CT sample. This transmitted flux was used for evaluating the atomic attenuation of the sample. We chose the LaBr$_3$(Ce) scintillation detector due to a better energy resolution, 2.7% at 662 keV, than the NaI(Tl) scintillation detector whose typical energy resolution is approximately 6% to 8% at 662 keV. In addition, the availability of a high counting rate of the LaBr$_3$(Ce) scintillation detector due to the short decay time is suitable for our measurement. The intrinsic radiation of the LaBr$_3$(Ce) scintillation detector contributes significantly to the background radiation in the region below 1.6 MeV [20], which is far below our energy of interest (around 5.5 MeV). To avoid signal pile-up on the LaBr$_3$(Ce) detector, a 5-cm-thick bismuth (Bi) absorber was placed in front of the detector. We used the bismuth as an absorber because of its high density. In addition, in the NRF-CT imaging technique, we measure the scattered NRF gamma rays from a witness target made from the isotope of interest such as the lead isotope $^{208}$Pb in our experiments. If we used lead material as an

![Figure 1](https://doi.org/10.3390/xxxxx)
absorber, the NRF gamma rays would be contaminated by the scattered NRF gamma rays from the absorber material. Therefore, it is preferable to avoid using the isotope of interest material as an absorber. Other high-density materials, such as tungsten, can be used as absorbers as well; however, bismuth is less expensive than tungsten.

The CT sample was scanned along three axes as follows: (i) the vertical position \((y)\) was fixed at a set position; (ii) 2D gamma-CT images were scanned in the horizontal direction \((x)\) across the range from \(-12\) to \(+12\) mm with a step size of \(1\) mm. The sample was scanned in each horizontal position in the direction of rotation \((\theta)\) across the range from \(0\) to \(150\) degrees with a step size of \(30\) degrees; (iii) (i) and (ii) were repeated for different vertical positions \((y)\) from \(1\) to \(22\) mm with a step size of \(1\) mm. We gathered 3300 data points in the presence of the CT sample by measuring 25 locations along the \(x\)-axis, 22 layers in the \(y\) direction, and 6 rotational angles \((\theta)\). In addition, we measured one data point for each scan in the \(x\) direction in the absence of the CT sample. Therefore, a total of 3432 data points were measured with an average measurement time of \(5\) s per data point. The overall time needed to obtain the gamma-ray spectra at all data points was approximately \(5\) h.

![Figure 2. Schematic diagram of the experimental setup to measure the 3D gamma-CT image at the beamline BL1U in the UVSOR-III synchrotron radiation facility.](image)

### 2.3. 3D NRF-CT Image Measurement

We used a high-flux LCS gamma-ray beam to measure the 3D NRF-CT image. Since the total gamma-ray flux was proportional to the average power of the laser with a proportional constant of \(3.5 \times 10^8\) photons/s/W, we used the Tm-fiber laser system with a typical average power of \(36\) W to generate the LCS gamma-ray beam. When the laser system was operated at \(36\) W, a total flux higher than \(10^8\) photons/s with a 100% energy bandwidth could be generated. We used a collimator with a hole diameter of \(2\) mm, which is twice as wide as that of the collimator used for 3D gamma-CT image measurement. The LCS gamma-ray beam flux passing through the collimator was numerically estimated using the EGS5 Monte Carlo simulation code [19], and thus we found that the flux density was 10 photons/s/eV and that the maximum energy was \(5528\) MeV with 2.9% energy bandwidth at the FWHM, which was able to excite the \(^{1}\) \text{NRF}\ level at \(5512\) MeV in \(^{208}\text{Pb}\). Although the NRF-CT imaging technique is time-consuming, we were able to shorten the image acquisition time for each measurement point, which had been measured in our previous research [13] by adjusting the CT sample scanning pattern and increasing the LCS gamma-ray beam’s intensity flux. The overall time needed to obtain a 3D NRF-CT image was approximately \(48\) h. Reference [14] gives more details on the experimental setup, the CT sample scanning plan, and a schematic diagram of the 3D NRF-CT image.

### 3. Results and Discussion

#### 3.1. 3D-CT Primary Images

CT imaging can be performed by measuring the decrease in the gamma-ray intensity passing through a CT sample along a series of linear paths and various angles. The amount of decrease depends on the gamma-ray energy, the path lengths, and the material’s linear attenuation coefficients. Therefore, one can obtain a gamma-CT image for a sample using
the gamma-ray transmission factor measurements of the off-resonance energy, known as off-resonance ($\epsilon_{\text{off}}$) attenuation, which originates from the atomic effect. The procedures of the 2D gamma-CT images calculation are detailed in references [13,14,21]. Since the height of the CT sample holder was 20 mm, we divided it into 20 layers and added two extra layers; one of the two extra layers was located below the bottom of the CT sample and the other above it. These layers were separated by a vertical distance of 1 mm. Therefore, the total number of scanned layers was 22. We assigned the labels L1 to L22 to the layers at the vertical distances of $y = -1$ mm to $y = 21$ mm, respectively. Consequently, we obtained one 2D gamma-CT image ($25 \times 25$) with a resolution of 1 mm/pixel for each row of rods in the $y$ direction of the CT sample scan. The pixel size was determined by the scanning step in the $x$ direction. In total, 22 reconstructed 2D gamma-CT images were obtained, which we named $L_{\text{off}1}$ to $L_{\text{off}22}$. Since Layer L1 was located beneath the CT sample holder and Layer L2 was located at the bottom edge of the holder, the LCS gamma-ray beam traveled underneath the rods. Therefore, the reconstructed images for the first two layers $L_{\text{off}1}$ and $L_{\text{off}2}$ did not show any characterization of the rods within the CT sample. Figure 3a shows the cross-sectional slices of the CT sample for the horizontal layers from L3 to L8. The total height of these layers was 6 mm, which was identical to the height of each rod within these layers. Two rods of enriched lead isotopes ($^{208}$Pb and $^{206}$Pb) were inserted into two of the holes in the CT sample, and an Fe rod was inserted into the third hole. Figure 3b shows the reconstructed CT images from $L_{\text{off}3}$ to $L_{\text{off}8}$ due to the atomic attenuation measured by the LaBr$_3$(Ce) detector. Because the lead isotope ($^{206}$Pb and $^{208}$Pb) rods had the same atomic attenuation, the two high-attenuation areas in the white-colored area caused by the atomic process corresponding to the locations of these lead rods were plainly evident in all reconstructed images. However, it was difficult to discriminate between the two isotopes. Furthermore, the atomic attenuation caused by the Fe rod was smaller than that caused by the lead rods, making the Fe rod appears as a faint shadow.

Figure 3. (a) Cross-sectional slice images of the CT sample in the layers from L3 to L8. (b) Reconstructed images from $L_{\text{off}3}$ to $L_{\text{off}8}$ of the off-resonance attenuation (atomic effect) measured by the LaBr$_3$(Ce) detector with a resolution of 1 mm/pixel.

Figure 4a shows the cross-sectional slices of the CT sample for the horizontal layers from L9 to L14. The total height of these layers was also 6 mm. Rods of $^{208}$Pb, Fe, and Al were inserted into the first, second, and third holes, respectively. The reconstructed CT images from $L_{\text{off}9}$ to $L_{\text{off}14}$ are shown in Figure 4b. A high-attenuation area caused by the atomic process corresponding to the $^{208}$Pb rod is clearly visible in the reconstructed images.
The atomic attenuation originating from the Fe rods was less than that caused by the $^{208}\text{Pb}$ rod, making the Fe rod discernible but not as apparent. Since the CT sample holder was Al, the Al rod was not visible.

**Figure 4.** (a) Cross-sectional slice images of the CT sample in the layers from L9 to L14. (b) Reconstructed CT images from $L_{\text{off}9}$ to $L_{\text{off}14}$ of the off-resonance attenuation (atomic effect) with a resolution of 1 mm/pixel.

The results for the horizontal layers from L15 to L20 are shown in **Figure 5**. Two lead isotope rods ($^{206}\text{Pb}$ and $^{208}\text{Pb}$) were inserted into the two holes, while the third hole remained unfilled as shown in **Figure 5a**. **Figure 5b** shows the reconstructed images from $L_{\text{off}15}$ to $L_{\text{off}20}$. The two high-attenuation areas corresponding to the $^{206}\text{Pb}$ and $^{208}\text{Pb}$ rods can be clearly observed but they cannot be distinguished. The low-intensity area at the third hole corresponds to the vacant space.

**Figure 5.** (a) Cross-sectional slice images of the CT sample in the layers from L15 to L20. (b) Reconstructed CT images from $L_{\text{off}15}$ to $L_{\text{off}20}$ of the off-resonance attenuation (atomic effect) at a resolution of 1 mm/pixel.
Since Layers L21 and L22 were located above the inserted rods within the CT sample, the LCS gamma-ray beam did not pass through these rods. Therefore, there was no characterization for the rods in the two reconstructed images, \( L_{\text{off}}21 \) and \( L_{\text{off}}22 \). The 3D-CT image was, in general, primarily made up of the 2D-CT images taken at various positions. Thus, the 22 measured 2D gamma-CT images were visualized together to create one 3D gamma-CT image. We used the MicroAVS data visualization tool [22] to create the 3D gamma-CT image with a resolution of 1 mm/pixel. Movie S1 shows the visualized 3D gamma-CT image. Figure 6 shows a shot of the 3D gamma-CT image, which was captured from the visualized 3D movie (see Movie S1 in the Supplementary Materials). We adjust a value of approximately 60% of the maximum intensity of the 2D gamma-CT images as a threshold limit for visualizing the 3D surface. Since the Fe or Al rods induced less atomic attenuation than the lead rods, this threshold limit ensured that only the lead rods appeared on the 3D surface. Intensity values greater (less) than the limit appeared (disappeared) on the 3D image. The five high-attenuation areas caused by the atomic process, which correspond to the positions of the five enriched lead isotope (\(^{208}\text{Pb} \) and \(^{206}\text{Pb} \)) rods within the CT sample, are clearly visible. The gamma-CT imaging technique is capable of clearly identifying the presence of the materials to be examined. In addition, we obtained the 3D gamma-CT image with a higher resolution and shorter acquisition time than the NRF-CT imaging technique. However, it lacked the isotope-selective capability needed to distinguish between different isotopes of the same element. Therefore, we utilized the 3D gamma-CT image as an additional information source to improve the quality of the 3D isotope-selective CT imaging based on the NRF transmission method.

![Figure 6](image-url)

**Figure 6.** A shot of the three-dimensional visualization of the atomic attenuation (3D gamma-CT image). See corresponding Movie S1 in the Supplementary Materials.

For the primary 3D NRF-CT image, we chose the layers at the vertical distances of \( z = 3, 11, \) and 17 mm, measured from the holder’s bottom, to reconstruct the 2D NRF-CT images of \( L_{\text{NRF}}4, L_{\text{NRF}}12, \) and \( L_{\text{NRF}}18, \) respectively. One 2D NRF-CT \((7 \times 7)\) image was obtained for each horizontal row of rods (see Figure 1a). In our previous study [14], we reported the visualization of the reconstructed 3D images with a resolution of 4 or 8 mm/pixel for the horizontal or vertical plane, respectively, which were based on the atomic attenuation and the pure NRF attenuation. The measurement procedure, numerical equations, and 2D NRF-CT images were reported in [14]. If a 3D NRF-CT image were measured under the
We scanned the CT sample in the $-x$ direction with a 1 mm step size across the range from $-12$ to $+12$ (25 positions). Furthermore, we scanned the CT sample in the $\theta$ direction with an angle step of $30^\circ$ across the range from $0^\circ$ to $150^\circ$ (six angles). Therefore, we obtained sinograms of the $\varepsilon_{\text{off}}$ transmission factor $M_{\varepsilon_{\text{off}}}(x_i, \theta_j)$ in a matrix with a size of $25 \times 6$. Moreover, the reconstructed 2D gamma-CT images had a matrix size of $25 \times 25$. For the 2D NRF-CT images, we scanned the CT sample in the $x$ direction with a 4 mm step size across the range from $-12$ to $+12$ (seven positions). We scanned the CT sample in the $\theta$ direction with an angle step of $30^\circ$ across the range from $0^\circ$ to $150^\circ$ (six angles). Therefore, we obtained sinograms of the $\varepsilon_{\text{NRF}}$ transmission factor $M_{\varepsilon_{\text{NRF}}}(x_i, \theta_j)$ with a size of $7 \times 6$ [14]. The reconstructed 2D NRF-CT images had a matrix size of $7 \times 7$. For overlapping the two 2D-CT images resulting from the sinograms of the reconstructed transmission factors $\varepsilon_{\text{NRF}}$ and $\varepsilon_{\text{off}}$, the images needed to have the same size in both directions ($x$ and $z$). In order to ensure the two images had the same size in both directions, we introduced a sinogram adjustment process to numerically divide each $\varepsilon_{\text{NRF}}$ sinogram from $M_{\varepsilon_{\text{NRF}}}(x_i, \theta_j)$ with a dimension of $7 \times 6$ to $M_{\varepsilon_{\text{NRF}}}(x_i, \theta_j)$ with a dimension of $25 \times 6$ as follows: we divided each value in the $x$ direction equally into 25 values to create a sinogram $(x, \theta)$ with a dimension of 175, 6. We then combined each seven consecutive values together into one value to create a new sinogram in a dimension of $25 \times 25$. Therefore, the newly reconstructed images (2D NRF~–~CT) had a size of $25 \times 25$.

3.2. Post-Multiply FV Method

One of the FV numerical approaches is the so-called “post-multiply FV method”, which analyzes a 3D gamma-CT image with a resolution of 1 mm/pixel as an additional data source to improve the resolution of the 3D NRF-CT image [14]. The procedure of the post-multiply FV method is as follows: (i) sinogram adjustment, (ii) layer alignment ($x, \theta$), (iii) 2D gamma-CT image segmentation, (iv) 2D-CT image overlapping, and (v) image post-multiply FV method.

(I) Sinogram Adjustment

The transmission factors of the off-resonance attenuation ($\varepsilon_{\text{off}}$) and the on-resonance attenuation ($\varepsilon_{\text{NRF}}$) [14] were gathered into a 2D matrix (sinogram). The size of the two-dimensional sonogram was determined by the scanning steps in the $x$ direction and the $\theta$ angle. We obtained the 2D gamma-CT images by scanning the CT sample in the $x$ direction with a 1 mm step size across the range from $-12$ to $+12$ (25 positions). Furthermore, we scanned the CT sample in the $\theta$ direction with an angle step of $30^\circ$ across the range from $0^\circ$ to $150^\circ$ (six angles). Therefore, we obtained sinograms of the $\varepsilon_{\text{off}}$ transmission factor $M_{\varepsilon_{\text{off}}}(x_i, \theta_j)$ in a matrix with a size of $25 \times 6$. Moreover, the reconstructed 2D gamma-CT images had a matrix size of $25 \times 25$. For the 2D NRF-CT images, we scanned the CT sample in the $x$ direction with a 4 mm step size across the range from $-12$ to $+12$ (seven positions). We scanned the CT sample in the $\theta$ direction with an angle step of $30^\circ$ across the range from $0^\circ$ to $150^\circ$ (six angles). Therefore, we obtained sinograms of the $\varepsilon_{\text{NRF}}$ transmission factor $M_{\varepsilon_{\text{NRF}}}(x_i, \theta_j)$ with a size of $7 \times 6$ [14]. The reconstructed 2D NRF-CT images had a matrix size of $7 \times 7$. For overlapping the two 2D-CT images resulting from the sinograms of the reconstructed transmission factors $\varepsilon_{\text{NRF}}$ and $\varepsilon_{\text{off}}$, the images needed to have the same size in both directions ($x$ and $z$). In order to ensure the two images had the same size in both directions, we introduced a sinogram adjustment process to numerically divide each $\varepsilon_{\text{NRF}}$ sinogram from $M_{\varepsilon_{\text{NRF}}}(x_i, \theta_j)$ with a dimension of $7 \times 6$ to $M_{\varepsilon_{\text{NRF}}}(x_i, \theta_j)$ with a dimension of $25 \times 6$ as follows: we divided each value in the $x$ direction equally into 25 values to create a sinogram $(x, \theta)$ with a dimension of 175, 6. We then combined each seven consecutive values together into one value to create a new sinogram in a dimension of $25 \times 25$. Therefore, the newly reconstructed images (2D NRF~–~CT) had a size of $25 \times 25$.

(II) Layer Alignment ($x, \theta$)

We measured the 22 layers of the gamma-CT measurement ($L_{\text{off}}1$ to $L_{\text{off}}22$) in addition to the three layers of the NRF-CT measurement ($L_{\text{NRF}}4$, $L_{\text{NRF}}12$, $L_{\text{NRF}}18$), and we aligned every 2D NRF~–~CT image to a group with six 2D gamma-CT images measured for six consecutive layers. Figure 7 illustrates the aligned layers in the vertical direction for both kinds of images. We aligned the 2D gamma-CT images from $L_{\text{off}}3$ to $L_{\text{off}}8$, which included the rods of $^{208}$Pb, $^{206}$Pb and Fe, with the 2D NRF~–~CT image of $L_{\text{NRF}}4$. We also aligned the 2D gamma-CT images from $L_{\text{off}}9$ to $L_{\text{off}}14$, which included the rods of $^{208}$Pb, Fe, and Al, with the 2D NRF~–~CT image of $L_{\text{NRF}}12$. The 2D gamma-CT images from $L_{\text{off}}15$ to $L_{\text{off}}20$, including the $^{206}$Pb and $^{208}$Pb rods in addition to the vacant area, were aligned with the 2D NRF~–~CT image of $L_{\text{NRF}}18$. 
The arrangement of the inserted rods within the CT sample holder and the scanned layers in the \( z \) direction used for the 2D gamma-CT and NRF-CT images.

(III) 2D Gamma-CT Image Segmentation

Image segmentation is a process of partitioning an image into numerous segments or a process of placing pixels such that they have non-overlapping areas [23]. This segmentation procedure is the first stage of image analysis [24], object representation, visualization, and other image processing strategies used in a variety of fields [25]. The main goal of image segmentation is to simplify and/or transform an image into one that can be readily analyzed [24]. In the case of segmentation of a 2D gamma-CT image, we assumed that the atomic absorption intensities varied in some regions of the CT sample and that the intensity at each pixel in a region where a rod existed would be almost constant. The first step of 2D gamma-CT image segmentation was primary image scaling. We scaled the primary gamma-CT images to a value between 0 and 255. The range of 0 to 255 was chosen to provide for an 8-bit representation of each pixel. The black, gray, and white colors were represented by values of 0, 128, and 255, respectively. One of the most commonly used approaches used for image segmentation is the threshold method [24,26]. Thresholding plays a crucial role in various algorithms used for image analysis, object representation, and visualization [27]. In this method, we chose a threshold value \( (T_r) \) for transforming an image from a grayscale into a binary image, in which each pixel has a value of 0 or 1, to distinguish the foreground and background of the image [28]. A pixel with an intensity greater (lower) than \( T_r \) is shown by a white-colored (black-colored) area. Thresholding methods are grouped into two types: local thresholding and global thresholding. The local thresholding methods apply different threshold values to different regions of the image. Each \( T_r \) value is determined by the neighborhood of the pixel to which the threshold is applied [26,29]. On the other hand, in global thresholding, a single \( T_r \) value is used to separate the foreground and the background for all pixels in an image [24,30]. In the present analysis, we performed global thresholding with a common threshold \( T_r \) value of 128 as the second step of gamma-CT image segmentation. The black and white 2D segmented gamma-CT images of the scanned layers are shown in Figure 8, with the layers labeled from \( L_{sg}3 \) to \( L_{sg}20 \). The positions of the inserted rod turned entirely white, while the residual areas turned black. The image segmentation procedure resulted in a set of segments that collectively covered the whole image or a set of contours as the object’s edge. Each pixel in the inserted rods’ positions had a common property such as color or intensity. Furthermore, the neighboring areas differed substantially in terms of the same characteristic. When 2D segmented gamma-CT
images are created for the employment of the FV technique with the NRF-CT images after the gamma-CT image segmentation, the resulting contours can be used to precisely preserve the $^{208}\text{Pb}$ rod locations within the gamma-ray images of the CT sample and cut the surrounding noise and distortions.

Figure 8. Cont.
(IV) 2D-CT Image Overlapping

We overlapped the primary 2D-CT images to obtain a 2D FV NRF-CT image as follows: the 2D NRF∼−CT images of Layers L_{NRF4}, L_{NRF12}, and L_{NRF18} were overlapped with the 2D segmented gamma-CT images from L_{sg3} to L_{sg8}, L_{sg9} to L_{sg14}, and L_{sg15} to L_{sg20}, respectively, according to the following equation:

\[ [2D\ FV\ NRF-CT(x_i, z_j)] = [2D\ NRF∼−CT(x_i, z_j)] \times [2D\ segmented\ gamma−CT(x_i, z_j)] \]  

We labeled the resulting 2D FV NRF-CT images as L_{FV3} to L_{FV20}. While the \( ^{208}\text{Pb} \) rod is clearly visible in the fused images from L_{FV3} to L_{FV8}, as shown in Figure 9a, the Fe and \( ^{206}\text{Pb} \) rods are almost invisible. Figure 9b shows the fused images from L_{FV9} to L_{FV14}. Only the \( ^{208}\text{Pb} \) rod can be seen clearly, but neither the Fe nor the Al rods are visible. Moreover, we can clearly pinpoint the location of the \( ^{208}\text{Pb} \) rod in the reconstructed images from L_{FV15} to L_{FV20}, as shown in Figure 9c. The \( ^{206}\text{Pb} \) rod and the vacant area disappeared. Obviously, the noise in all 2D FV NRF-CT image backgrounds caused by using the post-multiply FV method for the primary data sources was totally eliminated from the reconstructed image background. Since the overlapping process of the post-multiply FV method was performed by multiplying the intensity value at each pixel in the 2D NRF∼−CT by a value of 1 within the \( ^{208}\text{Pb} \) locations or by a value of 0 in the surrounding regions, the method preserved the locations of the isotopes of interest and completely eliminated the surrounding distortion and background noise. Furthermore, it kept the intensity value at each pixel constant, making this approach useful for isotope quantification.
from $L_{FV}^{15}$ to $L_{FV}^{20}$, as shown in Figure 9c. The $^{206}\text{Pb}$ rod and the vacant area disappeared. Obviously, the noise in all 2D FV NRF-CT images caused by using the post-multiply FV method for the primary data sources was totally eliminated from the reconstructed image background. Since the overlapping process of the post-multiply FV method was performed by multiplying the intensity value at each pixel in the 2D NRF-CT by a value of 1 within the $^{208}\text{Pb}$ locations or by a value of 0 in the surrounding regions, the method preserved the locations of the isotopes of interest and completely eliminated the surrounding distortion and background noise. Furthermore, it kept the intensity value at each pixel constant, making this approach useful for isotope quantification.

(III) Image Visualization in 3D

The 22 2D FV NRF-CT images were combined together using the MicroAVS data visualization tool to create the 3D FV NRF-CT image. We also adjusted a value within the intensity range of the 2D FV NRF-CT images as a threshold limit, so that only the intensity values greater than the threshold limit ($^{208}\text{Pb}$ rods) appeared on the 3D surface. Movie S2 shows a visualization of the fused CT-image of the NRF attenuation caused by the $^{208}\text{Pb}$ isotope rods within the CT sample (pure NRF) in three dimensions. Figure 10 shows a shot of the 3D FV NRF-CT image, which was captured from the visualized 3D movie (see Movie S2 in the Supplementary Materials). The visualization clearly shows the locations of the enriched lead isotope ($^{208}\text{Pb}$) rods. In contrast, the rods of $^{206}\text{Pb}$, Fe, and Al and the empty areas are not visible.

Figure 9. Reconstructed 2D FV NRF-CT images from (a) $L_{FV}^{3}$ to $L_{FV}^{8}$, (b) $L_{FV}^{9}$ to $L_{FV}^{14}$, and (c) $L_{FV}^{15}$ to $L_{FV}^{20}$ for the post-multiply FV method with the CT sample’s geometry for each layer group.
(V) Image Visualization in 3D

The 22 2D FV NRF-CT images were combined together using the MicroAVS data visualization tool to create the 3D FV NRF-CT image. We also adjusted a value within the intensity range of the 2D FV NRF-CT images as a threshold limit, so that only the intensity values greater than the threshold limit \( (^{208}\text{Pb} \text{ rods}) \) appeared on the 3D surface. Movie S2 shows a visualization of the fused CT-image of the NRF attenuation caused by the \( ^{208}\text{Pb} \) isotope rods within the CT sample (pure NRF) in three dimensions. Figure 10 shows a shot of the 3D FV NRF-CT image, which was captured from the visualized 3D movie (see Movie S2 in the Supplementary Materials). The visualization clearly shows the locations of the enriched lead isotope \( (^{208}\text{Pb}) \) rods. In contrast, the rods of \( ^{206}\text{Pb}, \text{Fe}, \text{and Al} \) and the empty areas are not visible.

![Figure 10](image-url)

Figure 10. A shot of the three-dimensional visualization of the fused CT-image of the NRF attenuation caused by the distribution of \( ^{208}\text{Pb} \) (3D FV NRF-CT image). See corresponding Movie S2 in the Supplementary Materials.

We obtained an isotope-selective 3D FV NRF-CT image with a resolution of 1 mm/pixel for the distribution of an enriched lead isotope \( (^{208}\text{Pb}) \) inserted with rods of different materials within a cylindrical sample 20 mm in height and 25 mm in diameter. Since the fused image has the same pixel resolution as the primary gamma-CT image, the image distortion and the background noise vanished, and the locations of the \( ^{208}\text{Pb} \) rods were clearly visible. In contrast, the other rods completely disappeared, since their intensity values were less than the selected threshold limit of the 3D surface. Therefore, the combination of primary 3D NRF-CT images with high-resolution 3D gamma-CT images obtained by the FV technique provided a beneficial improvement in the image resolution and the isotope-selective capability in three dimensions.

The numerical treatment of the FV technique can be used in different approaches for primary 3D-CT images. Most of these approaches are similar to the post-multiply FV method, except for a few minor differences. One of the approaches is to directly overlap a 2D NRF-CT image with a 2D gamma-CT image without the 2D gamma-CT image segmentation process according to the following equation:

\[
[2D \text{ FV NRF} - \text{CT} (x_i, z_j)] = [2D \text{ NRF} - \text{CT} (x_i, z_j)] \times [2D \text{ gamma} - \text{CT} (x_i, z_j)]
\]  \hspace{1cm} (2)
The 2D FV NRF-CT image obtained by this approach in the current study had the required quality. However, it could not be used for isotope quantification because the intensity at each pixel of the NRF-CT image was not conserved. Furthermore, even if the background noise were significantly reduced, it would be impossible to eliminate all noise. Another approach is the post-sum FV method, for which the overlapping is as follows:

\[
[2D \text{ FV NRF} - \text{CT} (x_i, z_j)] = [2D \text{ NRF} - \text{CT} (x_i, z_j)] + [2D \text{ gamma} - \text{CT} (x_i, z_j)]
\]  

(3)

The fused image obtained using the post-sum FV method shows the change in the intensity at each pixel but retains some of the background noise, so this method is useless for isotope quantification. Figure 11 shows a comparison between the 2D FV NRF-CT images for the layer L4 obtained by (a) the post-multiply FV method, (b) the post-multiply FV method without 2D gamma-CT image segmentation, and (c) the post-sum FV method.

![Figure 11](image)

Figure 11. (a) 2D NRF-CT images obtained via the post-multiply FV method, (b) the post-multiply FV method (without 2D gamma-CT image segmentation), and (c) the post-sum FV method for layer L4.

These results show that the alternative approaches may change the intensity of each pixel, so that they are unable to maintain the quantity of the isotope of interest. Furthermore, the noise cannot be completely removed even if it can be greatly decreased. Therefore, we recommend the post-multiply FV method as the most beneficial method for NRF-CT imaging that is likely to be effectively applicable in a variety of fields, including nuclear engineering and nuclear safety.

4. Conclusions

The fusion visualization (FV) technique can be used as an effective numerical treatment to improve computed tomography (CT) imaging based on the nuclear resonance fluorescence (NRF) transmission method. We obtained two three-dimensional (3D) CT images for the distribution of an enriched lead isotope (208Pb) inserted into a cylindrical holder together with rods of various materials (another enriched lead isotope of 208Pb, iron, and aluminum). One of the images was the isotope-selective 3D CT image based on the NRF transmission method, which had a relatively low resolution. The other was a significantly higher-resolution 3D gamma-CT image. The two images were measured at the beamline BL1U in the ultraviolet synchrotron orbital radiation-III (UVSOR-III) synchrotron radiation facility at the Institute of Molecular Science at the National Institutes of Natural Sciences in Japan. We generated a laser Compton scattering (LCS) gamma-ray beam with a maximum energy of 5528 MeV, which was able to excite the \(J^π = 1−\) state at 5512 MeV in 208Pb. We used two LCS beams with different intensities and beam diameters for two experiments. First, we generated an LCS gamma-ray beam with a diameter of 1 mm and a flux of 0.7 photons/s/eV to obtain a 3D gamma-CT image with a resolution of 1 mm/pixel. Next, we generated an LCS gamma-ray beam with a diameter of 2 mm and a flux of 10 photons/s/eV to measure the 3D NRF-CT image with a resolution of 4 and 8 mm/pixel in the horizontal and vertical planes, respectively. We applied the FV technique using the post-multiply FV method between the two primary images to create a 3D isotope-selective fused CT image based on the NRF transmission method (3D FV.
NRF-CT) for the distribution of $^{208}$Pb in the CT sample. The background noise in the fused 3D image was removed and the quantity of the isotope of interest was conserved. Although we also examined the direct overlapping method and the post-sum FV method, we concluded that the post-multiply FV method was the most suitable for isotope-selective CT imaging to be used in nuclear engineering and various nuclear applications.

**Supplementary Materials:** The following are available online at https://www.mdpi.com/article/10.3390/app112411866/s1, Movie S1: Three-dimensional visualization of the atomic attenuation (3D gamma-CT image), Movie S2: Three-dimensional visualization of the fused CT image of the NRF attenuation caused by the distribution of $^{208}$Pb (3D FV NRF-CT image).

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