Dose reduction and image quality in CT angiography for cerebral aneurysm with various tube potentials and current settings

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Objectives: The purpose of this study is to investigate the image quality on both axial and three-dimensional CT angiograms of the brain at various tube potentials and currents, and to propose the use of descriptors for evaluating the image quality of three-dimensional CT angiograms using entropy analysis.

Methods: A head phantom was used as a target object. Axial CT and three-dimensional CT angiograms were obtained at various effective milliamperere-second values (49–350 mAs) and tube potentials (80–140 kVp) with a 64-row detector CT scanner. Lens doses were measured using a planar silicon pin-photodiode system. The signal-to-noise ratio (SNR) and streak artefacts on the axial CT angiograms were evaluated and the image quality of the three-dimensional CT angiograms was assessed using entropy analysis.

Results: Lens doses increased with tube potential and effective milliamperere-seconds. From the evaluation of SNR and streak artefacts on axial CT angiograms, we found that the image quality was improved by setting the tube potential at 100 kVp. However, there was little visual difference in the image quality for 100 kVp between 252 (effective value recommended by the manufacturer) and 350 mAs (maximum effective value). In the entropy analysis of the image quality of three-dimensional CT angiograms, the mutual information (information gain) per lens dose was largest at 80 kVp and 252 mAs.

Conclusion: Our results suggested that the suitable tube potentials for axial CT and three-dimensional CT angiograms were 100 and 80 kVp, respectively, and the effective milliamperere-second value recommended by the manufacturer was appropriate.

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Aneurysms are one of the most important causes of subarachnoid haemorrhage (SAH). Currently, intra-arterial digital subtraction angiography (IADSA) is the most reliable method for detecting aneurysms because of its high resolution and large field of view; however, it is invasive and time consuming, and carries a risk of morbidity and mortality. Therefore, as an alternative method of IADSA, an accurate non-invasive examination is desirable in emergency screening for SAH.

In the last few decades, CT scanners have progressed remarkably through the introduction of helical scanning algorithms and the development of multirow X-ray detectors. These improvements in CT scanners have made it possible to detect smaller cerebral vascular lesions such as small aneurysms. Moreover, since CT images with thin slices (0.5 mm thickness) can be consecutively acquired by using these technologies, it is easy to reconstruct isotropic images, and three-dimensional (3D) CT angiography (CTA) can be obtained. 3D CTA examination has shown potential in the low-invasive detection of

aneurysms and has made it possible to observe intracranial aneurysms from all directions. Recent literature has demonstrated high accuracy in detecting aneurysms more than 3 mm in diameter using CTA technique [1]. Thus, CTA examination is expected to play an important role in the detection of aneurysms. On the other hand, there is a trend towards an increased radiation dose with multi-detector row CT (MDCT), compared with single-detector CT [2]. Therefore, the evaluation of the radiation dose and image quality for CTA images is important.

Recently, many studies on the optimisation of tube potential and tube current settings in CTA examinations have been performed from the viewpoint of dose reduction [3–8]. Waaijer et al [6] reported that, in a phantom study with identical CT dose index volume, a tube potential of 90 kVp resulted in a 45–52% increase in the squared signal-to-noise ratio (SNR²) compared with the SNR² at 120 kVp. By contrast, Ertl-Wagner et al [8] reported that a higher potential resulted in better image quality. However, most of these investigations have shown the ability of axial CT images to detect aneurysms, whereas few studies have evaluated this ability in both axial and 3D CTA images. In addition, there is no definitive quantitative descriptor of image quality for 3D CTA images. If a quantitative method of evaluating image quality on 3D CT images is developed, it will surely prove very useful.

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In the present study, we have assessed the image quality of axial and 3D CT angiograms of the brain at various tube potentials and current settings using an anthropomorphic vascular phantom with simulated intracranial aneurysms and cranial bone structure and three physical indices devised on the basis of information theory.

Methods and materials

Head phantom design

We designed the head phantom used in this study, and subcontracted the phantom manufacture to R-TEC, Hamamatsu, Japan. The head phantom was used as a target object. The anthropomorphic phantom was composed of simulated brain parenchyma, intracranial arteries with various intracranial aneurysms and skull bone. The simulated brain parenchyma was made of polyurethane polymer which contained 2.5% calcium phosphate to achieve a similar CT number to brain parenchyma (CT number: 30–40 HU). The simulated intracranial arteries were bored through the polyurethane. Three types of simulated aneurysms (*i.e.* four aneurysms 3 mm in diameter, three aneurysms 4 mm in diameter and three aneurysms 5 mm in diameter) were placed on the simulated cerebral artery. The simulated cranium was made of gypsum, which had almost the same CT number as the cranial bone. Moreover, the simulated cranium was enveloped by polyurethane polymer, which had almost the same CT number as the soft tissue. The anthropomorphic phantom is shown in Figure 1.

CT image acquisition

The anthropomorphic phantom was scanned using a 64-row helical MDCT scanner (LightSpeed VCTTM; GE Medical Systems Milwaukee, WI). To determine the concentration of non-ionic contrast medium [OmnipaqueTM

300 (300 mg ml⁻¹ iodine); Daiichi Sankyo, Tokyo Japan], mean CT numbers for cerebral arteries in clinically enhanced CT images were measured. The mean CT number was estimated to be approximately 250 HU. From this measurement, we decided to adjust the concentration of the contrast medium to 10 mg ml⁻¹. The simulated arteries were filled with contrast medium at a concentration of 10 mg ml⁻¹, and the anthropomorphic phantom was scanned. The parameters for the CTA examination were as follows: beam width of 0.625×32 mm; beam pitch of 0.531; gantry rotation period of 0.4 s; slice thickness of 0.5 mm; field of view of 230 mm; tube potentials of 80, 100, 120 and 1401 kVp; tube currents of 65, 200, 335 and 465 mA; effective mAs of 49, 151, 252 and 350 (effective mAs was calculated using (tubecurrent)×(rotationperiod)/(beampitch)); and reconstruction kernels of “Standard” for soft tissue and brain. All CT images were transferred from the CT scanner to a personal computer using the standard file format with a matrix size of 512×512 and a grey level of 16 bits found in the Digital Imaging and Communications in Medicine (DICOM) standard.

The axial CT images were also transferred to a workstation for post-processing. The slice images were converted into multiplanar reconstruction images and the 3D CTA images were obtained using volume-rendering software (zio TermTM 2009; ziosoft, Tokyo, Japan). In order to obtain 3D intracranial artery images, skull bone was automatically eliminated from the 3D CTA images using this software. This reconstruction processing was performed by a radiological technologist with 15 years' experience in 3D reconstruction. The 3D reformatted images (3D CTA images) were transferred to a personal computer by DICOM format with a matrix size of 512×512 and a grey level of 16 bits for further analysis.

Dose measurement on head phantom

Radiation dose to the anthropomorphic phantom was evaluated using a dosimetry system with pin photodiode

Figure 1. Anthropomorphic head phantom. (a) External view; (b) three-dimensional CT image.

sensors, which was developed in a previous study [9]. Commercially available planar silicon pin photodiodes (HamamatsuTM S8385-04; Hamamatsu Photonics, Hamamatsu, Japan) possessing a sensitive area of 2.0×2.0 mm² were used as the dosimeters. Since the sensitivity of the photodiode to diagnostic X-rays differed with the incident direction of X-rays between front and back, two photodiodes were glued together back to back with epoxy resin, and then used as a single dosimeter with a parallel connection to obtain the isotropic sensitivity to incident X-rays [9]. Dose calibrations were performed against a RadcalTM 9015 dosimeter (Radcal Corporation, Monrovia, CA) with a 6 cm³ ion chamber attached, which was placed adjacent to a photodiode dosimeter at the same distance from the X-ray tube in an irradiation field. The ion chamber dosimeter is a tertiary standard, calibrated at the Japan Quality Assurance Organization in February 2007, where dosimeter readings were calibrated to exposure doses at six points of effective or equivalent photon energies from 20 to 72 keV. The pin photodiode dosimeter readings calibrated to the exposure dose were converted to the absorbed dose in soft tissue using the mass energy-absorption coefficient for soft tissue at each effective energy. The X-ray energy dependence of the sensitivity of the photodiodes is approximately flat within 8% in an effective energy range between 25 and 45 keV, but decreased at a rate of 6%/10 keV with increasing energy between 45 and 70 keV. Further, the photodiodes have a flat angular response within 2%.

In the head CT examination, the eye lenses were exposed to primary X-ray beams. It is well known that lenses are highly sensitive to radiation [10]. Thus, the two pin photodiode devices were placed on the surface of both eyelids of the anthropomorphic phantom, and the radiation doses for the eyes were measured five times under the same exposure condition as the CT image acquisition. The mean absorbed dose was then calculated using the conversion factor estimated for each dosimeter separately at the effective energy of X-rays used. The average dose of both sides was taken as the lens dose.

Evaluation of image quality for axial CTA image

In this study, signal-to-noise ratio (SNR) and streak artefacts on the axial CT images (source images for 3D CTA) were quantitatively assessed. The axial CT images at the level of the circle of Willis were employed as the target images for evaluating SNR and streak artefacts.

Evaluation of image noise using Gauss evaluation method and SNR

As shown in Figure 2a, to evaluate SNR, a rectangular region of interest (ROI) with the size of 40×40 pixels (Area A) was placed on the simulated right middle cerebral artery in the axial CT image; a simulated aneurysm of diameter 5 mm was located in the middle of Area A. A small rectangular area 5×5 pixels (Area B) was also placed on the simulated aneurysm in Area A. The mean CT number in Area B, μ_{aneurysm} , was regarded as the signal (S) of SNR. The standard deviation (SD) of CT numbers (so-called noise SD) of the simulated parenchyma around the simulated aneurysm was defined as

the noise (N) and was measured using our devised evaluation method—Gauss evaluation method [11]. This evaluation method makes it possible to evaluate the noise SD of the surrounding parenchyma, even if the aneurysm was included in the ROI. Therefore, the image noise on the parenchyma image which has an influence on the simulated aneurysm image can be assessed with high accuracy.

The following is a brief description of the Gauss evaluation method, adapted from our earlier study [11]. It is known that the image noise can be statistically characterised by the Gaussian distribution. Accordingly, the CT numbers attributed to the simulated parenchyma will be plotted so that the points form a straight line when they are plotted on normal probability paper. On the other hand, the CT numbers attributed to simulated aneurysms and arteries will not fall on the straight line on normal probability paper. Therefore, the noise SD of the simulated parenchyma in Area A can be estimated by the line component near the origin on the normal probability plot. That is, the image noise SD in the simulated parenchyma area, $\sigma_{\text{parenchyma}}$, can be estimated by:

$$\sigma_{\text{parenchyma}} = \frac{1}{m} \quad (1)$$

because the line component on the normal probability paper can be expressed by:

$$\Phi^{-1}[F(x)] = \frac{x - \mu}{\sigma_{\text{parenchyma}}} = mx + l \quad (2)$$

where $\Phi^{-1}[F(x)]$, $F(x)$, μ , m and l are the inverse standard normal distribution function for the CT number (random variable, x), unknown cumulative probability function, mean value, slope and y -intercept of the straight line, respectively. Here, the unknown cumulative probability function, $F(x)$, was estimated using the mean rank method with “order statistics” [12]. We used this estimation method to achieve high accuracy in calculating cumulative probabilities as recommended by Gumbel [12]. The CT numbers on ROI were arranged in ascending order, and $(\hat{F}_{x_{(i)}})$ was computed as:

$$(\hat{F}_{x_{(i)}}) = \frac{i}{n+1}, \text{ for } i=1, \dots, n \quad (3)$$

where n is a sampling size (in this study, $n=40 \times 40 = 1600$), and $x_{(1)} \leq x_{(2)} \leq \dots \leq x_{(1600)}$ are those 1600 arranged CT numbers. According to the calculation process mentioned above, the mean CT number of the aneurysm and the noise SD of the simulated parenchyma around it were estimated, and the SNR was calculated using the following equation:

$$\text{SNR} = \frac{\mu_{\text{aneurysm}}}{\sigma_{\text{parenchyma}}} \quad (4)$$

Evaluation of streak artefacts and their location parameter using Gumbel evaluation method

To evaluate streak artefacts on the axial CT images, an ROI with the size of 60×100 pixels (Area C) was placed on the image area at Willis circle level where streak

Figure 2. Example of axial CT angiography images of the phantom at the level of the circle of Willis. Figures (a) and (b) were acquired at 252 and 49 effective mAs, respectively, under a tube potential of 120 kVp. Figures (a) and (b) were used to evaluate the signal-to-noise ratio and streak artefact on axial CT images, respectively.

artefacts were noticeable, as shown in Figure 2b. Because there is no quantitative descriptor of streak artefacts on CT images, we analysed the streak artefacts on CT images using the extreme value theory from our previous studies [11, 13–15]. These results showed that the CT number variations caused by the streak artefacts can be statistically modelled by a Gumbel distribution, which is one of the generalised extreme value distributions; based on these results, we devised a new method of detecting the streak artefacts on CT images, which we called the Gumbel evaluation method. The following is a brief description of this new evaluation method, adapted from our earlier study [11, 13–15]. The CT numbers along each line segment with a length of 60 pixels in the rectangular region measuring 60×100 pixels in the axial CTA images were measured at 1-pixel intervals (*i.e.* along the white arrow in Figure 2b), and a total of 100 line profiles of the CT numbers were acquired.

In our earlier investigations, the largest difference between adjacent CT numbers of one CT number profile was attributed to streak artefacts [11, 13, 14]. Thus, the largest difference between adjacent CT numbers was employed as a feature variable of streak artefacts. Here, the largest difference between adjacent CT numbers was obtained from each of the CT number profiles. The 100 largest differences between adjacent CT numbers can be considered to be the largest values among a large set of independent and identically distributed random values, and can be modelled by a Gumbel distribution [11, 13–15]. The cumulative probability function of the Gumbel distribution, $F(x)$, is expressed as:

$$F(x) = \exp \left[- \exp \left(- \frac{x - \beta}{\gamma} \right) \right] \quad (5)$$

where β and γ are the location and scale parameters, respectively. The location and scale parameters correspond to the respective mode and variance of the probability density distribution [12]. Here, the unknown cumulative probability function can also be estimated by the mean rank method [*i.e.* Equation (3)], as well as that for the Gauss distribution. In the present study, the location parameter was estimated by line-fitting on the Gumbel plot, and was used as the physical index for evaluating streak artefacts on the axial CTA images.

When the fitting line for the Gumbel plot is expressed as $ax+b$, the location parameter β is given by $\beta=b/a$ because

$$- \ln \{ - \ln [F(x)] \} = \frac{1}{\gamma} x - \frac{\beta}{\gamma} \quad (6)$$

Therefore, we estimated the location parameter from the parameters, a and b , obtained by line-fitting on the Gumbel plot [11, 13, 14].

Evaluation of image quality for 3D CTA image

To date, the image quality of 3D CTA images has to be assessed by a human observer [16]. In order to obtain an accurate result using this method, many observers must participate in image-reading experiments, which are time consuming and tedious. Therefore, we have devised an alternative analysis method for assessing the image quality of 3D CTA images quantitatively.

The target image for assessing the image quality of a 3D CTA image is shown in Figure 3; most of the simulated aneurysms were included in this image. As shown in this figure, an ROI with a size of 200×100 pixels was placed on the 3D CTA image so as to include the detectable simulated aneurysms.

In CTA examinations of the head, it is important to understand the positional relationship between cerebral arteries with aneurysms and skull bone. However, the skull bone in 3D CTA images may obstruct the detection of aneurysms, depending on the situation [17, 18]. The more conspicuous the skull bone in CTA images, the more difficult the detection of aneurysms; consequently, the uncertainty of aneurysms in a head CTA image would increase. Thus, the skull bone images can be considered to play the role of anatomical noise in the detection of aneurysms [19, 20]. Signals buried in noise can be analysed using information theory. Therefore, we devised an analytical method for evaluating the image quality of 3D CTA images on the basis of information theory. In our devised analytical approach, “equivocation” and “information gain” are employed as the descriptors of image quality of 3D CTA images. The equivocation (or conditional entropy) of two random variables x and y taking possible values i and j , respectively, $H(x/y)$, is defined as:

Figure 3. Example of 3D CT image of cerebral arteries in the phantom. This figure was acquired at 120 kVp and 252 effective mAs.

$$H(x/y) = - \sum_{ij} p(j)p(i/j) \log p(i/j) \quad (7)$$

where $p(j)$ is a probability for possible value j and $p(i/j)$ is a conditional probability for possible value i [21, 22].

Now, the 3D CTA images of the simulated arteries with the simulated aneurysms and skull bone (*i.e.* 3D reformatted CTA image without the skull bone and 3D skull bone image) were regarded as signal and noise, respectively. Here, we consider the entropy of the pixel values of the 3D reformatted CTA image with the skull bone conditional on the corresponding image without the skull bone taking a certain pixel value. Thus, in Equation (7), $p(j)$ means the probability that the pixel value on the 3D reformatted CTA image without the skull bone will have a value within the j th bin, and the conditional probability, $p(i/j)$, means the probability that the pixel value on the 3D reformatted CTA image with the skull bone will fall within the i th bin, when the pixel value on the corresponding 3D reformatted CTA image without the skull bone takes a value within the j th bin. The probabilities in the calculation of this conditional entropy were estimated from the histogram with a discrete interval (bin) of eight pixel values. Intuitively, the equivocation estimated from this calculation process will give the quantity correlating with the uncertainty of the simulated artery with the simulated aneurysms in the 3D reformatted CTA image with skull bone; the smaller the equivocation, the higher the image quality of 3D CTA image.

Let us consider the mutual information (information gain), $I(x;y)$, defined as the difference between the entropy of the 3D reformatted CTA image with the skull bone, $H(x)$, and the equivocation, $H(x/y)$ [21, 22]:

$$I(x;y) = H(x) - H(x/y) \quad (8)$$

In contrast with the equivocation, this information gain will give the quantity correlating with the certainty of the 3D reformatted CTA image without the skull bone. Further, in order to consider the trade-off between radiation exposure and image quality, we consider the information gain per unit organ dose, and call it the "benefit-to-risk ratio" (BRR):

$$\text{BRR} = \frac{I(x;y)}{\text{organdose}} \quad (9)$$

In this study, the lens dose was regarded as the organ dose, because eye lenses are highly sensitive to radiation [10].

Results

Lens dose measurement

Figure 4 shows the relationship between the effective milliamperere-second value and lens dose. For each tube potential, the lens doses showed a linear relation to the effective milliamperere-second values, and their linear correlation coefficients (Pearson's r) were ≥ 0.999 . At a given effective milliamperere-second value, the lens dose increased as the tube potential increased.

Evaluation of SNR and streak artefacts on axial CT image

One example of the relationship between the CT numbers in Area A in Figure 2a and their cumulative probabilities is shown in Figure 5. For <50 HU, the CT numbers were distributed linearly on normal probability paper. Similar results were also obtained for the other effective milliamperere-second values. We confirmed that the CT numbers deviating from a linear distribution on the normal probability plot corresponded to those attributed to the simulated artery and aneurysms. Thus, the noise SD of the simulated brain tissue in Area A can be estimated by the Gauss evaluation method. Here, the

Figure 4. Effective milliamperere-second dependence of lens dose.

Figure 5. Noise analysis in Area A in Figure 2a using our devised Gauss evaluation method.

noise SD for the simulated brain tissue around the simulated aneurysm was estimated from the slopes of the line on this normal probability paper, and the SNR was calculated. Figure 6 illustrates the relationship between the effective milliamperere-second values and SNR. For each tube potential, the SNR increased with the effective milliamperere-second values. For >151 effective mAs, the SNRs were higher at 100 kVp than at 80, 120 or 140 kVp. However, at 49 effective mAs, the tube potential of 120 kVp gave the highest SNR; this effective milliamperere-second value is not used in clinical practice.

Figure 7 shows one example of the Gumbel plot of the estimated cumulative probability function *vs* the largest difference between the adjacent CT numbers. In the Gumbel plot, the largest difference between the adjacent CT numbers was distributed linearly. Here, the location parameters were estimated from the slopes and *y*-intercepts of the fitting lines for the Gumbel plots. As shown in Figure 8, the location parameter for each tube potential decreased as the effective milliamperere-second value increased. However, for >252 effective mAs, there was little difference among the location parameters at 100, 120 and 140 kVp, whereas the location parameter at 80 kVp was about twice as large at the other tube

Figure 6. Relationship between effective milliamperere-second value and signal-to-noise ratio (SNR) for various tube potentials.

Figure 7. Streak artefact analysis in the region of interest in Figure 2b using our devised Gumbel evaluation method.

potentials. This meant that there were many noticeable streak artefacts in the axial CTA image obtained at 80 kVp. In fact, we subjectively recognised that the streak artefacts in the axial CTA image obtained at 80 kVp were most noticeable and the streak artefacts at 100 kVp were almost the same level as 120 and 140 kVp.

Evaluation of image quality of 3D CTA image using equivocation, information gain and BRR

Figure 9 illustrates the relationship between effective milliamperere-second value, equivocation and information gain at 120 kVp. The equivocation reduced with the increase in the effective milliamperere-second value, whereas the information gain increased. Further, as illustrated in Figure 10, the simulated cerebral artery in the 3D CTA image obtained at 350 effective mAs has a smooth contour, but the simulated cerebral artery at 49 effective mAs has a notched one. Thus, the image quality of the 3D CTA image was improved by increasing the effective milliamperere-second value.

Figure 8. Relationship between the effective milliamperere-second value and location parameter in Gumbel distribution for various tube potentials.

Figure 9. Relationship among effective milliampere-second value, equivocation and information gain for three-dimensional CT angiography image obtained at 120 kVp.

The tube potential dependence of BRR at 252 and 350 effective mAs is shown in Figure 11; for reference, the information gains corresponding to the BRR are also included in this figure. The information gains decreased as the tube potential increased. As illustrated in Figure 12, the 3D CTA image obtained at 80 kVp shows higher contrast features than the image obtained at 120 kVp. At a given tube potential, the information gain at 350 effective mAs was slightly larger, at 252 effective mAs. On the other hand, the BRR also reduced as the tube potential increased. However, at a given tube potential, the BRR for 350 effective mAs was always smaller, than for 252 effective mAs. For example, at 120 kVp, the BRR was $0.034 \text{ bit mGy}^{-1}$ for 252 effective mAs and $0.025 \text{ bit mGy}^{-1}$ for 350 effective mAs, respectively. These results showed that the information gain per radiation dose was larger at 252 than at 350 effective mAs. Further, there was little visual difference in image quality between 252 and 350 effective mAs, as shown in Figure 12. From these results, the image quality of 3D CTA image can be considered to be improved by the reduction in tube potential. However, the image quality for the 3D CTA image obtained with the effective milliampere-second value recommended by the manufacturer was found to be better than in the other conditions.

Discussion

To our knowledge, there is no reliable method for quantitatively assessing the image quality of 3D CTA images. In this study we adopted equivocation and information gain to evaluate the image quality of the 3D CTA image. In the relationship between effective mAs

Figure 11. Relationship between tube potential and "benefit-to-risk ratio" at 252 and 350 effective mAs. Effective values of 252 and 350 mAs were manufacturer recommended and maximum values, respectively.

and these indices, the equivocation decreased as the effective milliampere-second value increased. By contrast, the information gain increased with the effective milliampere-second value. It is widely known that the image quality of CT image is improved by increasing the radiation dose, and that the effective milliampere-second value has a linear relation to radiation dose. In this study, the effective milliampere-second value was confirmed to show a linear relation to the radiation dose (Figure 4). From this result, we consider that the equivocation and information gain are useful indices for evaluating the image quality of 3D CTA images.

As for the axial CTA images, the SNR at 100 kVp was higher than with the other tube potentials. Further, for >252 effective mAs, streak artefacts at 100 kVp were almost the same level as at 120 and 140 kVp, and streak artefacts on the axial CTA image obtained at 80 kVp were most noticeable. From these results, the image quality of the axial CTA images was better at 100 kVp than at 80, 120 or 140 kVp. Additionally, the lens dose was lower at 100 kVp than at 120 or 140 kVp, the tube potentials used in clinical practice. For example, at 252 effective mAs, the lens doses for 100, 120 and 140 kVp were estimated to be 28.4, 40.9 and 54.2 mGy, respectively, and the lens dose reduction of 30–48% was achieved at 100 kVp. Our results were quite similar to those of Waaier et al [6]. Bahner et al [7] also reported that, in the quantitative assessment of radiation dose and contrast-to-noise ratio, CTA of the brain obtained by using 80 kVp not only is superior to 120 kVp in that it leads to higher contrast and contrast-to-noise ratio but also reduces patient radiation exposure by approximately 40%. However, Bahner et al performed the evaluation of image noise using a phantom made of homogeneous material (*e.g.* water

Figure 10. Examples of three-dimensional CT angiography images obtained at 49, 151, 251 and 350 effective mAs and 120 kVp.

Figure 12. Examples of three-dimensional CT angiography images acquired at 80, 100, 120 and 140 kVp. Upper and lower images show cerebral artery images obtained at 252 and 350 effective mAs, respectively.

phantom) to avoid the influence of beam-hardening artefacts caused by skull bone on the measurement of the noise SD. Thus, the noise SD reported by Bahner et al did not include streak artefacts on axial CTA images. Here, our results showed that the streak artefacts on the axial CT image obtained at 80 kVp were more noticeable than any others. Therefore, we suspect that the difference in the components involved in the noise SD causes the difference between these results.

From the above considerations, the suitable tube potential for an axial CTA image can be considered to be 100 kVp. Further, at a tube potential of 100 kVp, the SNRs were 20.57 for 252 effective mAs and 24.17 for 350 effective mAs, and an 18% increase in the SNR was achieved at 350 effective mAs. By contrast, a 38% increase of lens dose was confirmed under an exposure condition of 100 kVp and 350 effective mAs (39.3 mGy). We subjectively recognised that the image quality of the axial CTA image obtained at 252 effective mAs was the same as that at 350 effective mAs, as illustrated in Figure 12. Thus, the suitable value can be considered to be 252 effective mAs, which is the same as the value recommended by the manufacturer.

The image quality of the 3D CTA image was also quantitatively assessed using the equivocation, information gain and BRR. From these indices, the image quality of the 3D CTA image was indicated to be better at 80 kVp than at 100, 120 or 140 kVp. Further, from BRR, the appropriate effective value was 252 mAs; namely, the BRR for 80 kVp and 252 effective mAs was 0.090 bit mGy⁻¹, the highest of all exposure conditions. There are few studies on the image quality of 3D CTA images. As far as we are aware, Murakami et al [16], in a phantom study using 64-slice MDCT, are the only authors to report the superiority of 3D CTA images at 100 kVp to those at 120 kVp. Their report is in agreement with our finding that the image quality of a 3D CTA image was improved by reducing the tube potential. However, these authors suggested that it is desirable to use 100 kVp for 3D CTA imaging, which differs from our results. Here, whereas a reduction in the tube potential leads to an increase in the attenuation of iodinated contrast material, the usage of 80 kVp may cause an increase in image noise in addition to the accentuation of the beam-hardening artefact. Because 3D CTA images are reconstructed by experienced radiologists and radiological technicians using volume-rendering software, streak artefacts may have little effect on the image quality of

3D CTA compared with axial CTA. The k-edge of iodine is 33 keV, and it corresponds to the spectrum peak value for tube potentials between 60 and 70 kVp [23, 24]. Therefore, it would be reasonable to use 80 kVp for obtaining 3D CTA images.

As mentioned above, suitable tube potentials for axial CTA and 3D CTA were estimated to be 100 and 80 kVp, respectively. These results suggested that 80 kVp is suitable for the detection of aneurysms using only 3D CTA images and that 100 kVp is appropriate for protocols using both axial CTA and 3D CTA images. In addition, suitable effective milliamperes-second values for both axial CTA and 3D CTA images would be those recommended by the manufacturer (e.g. in this study, this value is 252 effective mAs). However, in order to validate these results, the image quality of CTA images must also be evaluated using image reading studies, a task left for future work.

Conclusion

We have investigated the image quality on both axial and 3D CTA images of cerebral arteries at various tube potentials and current settings using an anthropomorphic vascular phantom with simulated intracranial aneurysms and cranial bone structure and the three physical indices based on the information theory; these three indices were the equivocation, information gain and BRR. Our results suggested that the suitable tube potentials for axial CTA and 3D CTA images were 100 and 80 kVp, respectively, and that the appropriate effective milliamperes-second value was the one recommended by the manufacturer.

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