Article

Modified Contrast-Detail Phantom for Determination of the CT Scanners Abilities for Low-Contrast Detection

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Abstract: Computerised tomography (CT) continues to be a corner stone medical and radiologic imaging modalities in radiology and radiotherapy departments. Its importance lies in its efficiency in low contrast detectability (LCD). The assessment of such capabilities requires rigorous image quality analysis using special designed phantoms with different densities as well as variation in atomic mass numbers (A) of the material. Absence of such ranges of densities and atomic mass numbers, limits the dynamic range of assessment. An example is Catphan phantom which represents only three subject contrast levels 0.3, 0.5 and 1 per cent. This project aims to present a phantom with extended range of available subject contrast to include very low-level values and to increase its dynamic scale. With this design, a relatively large number of different contrast objects (holes) can be presented for imaging by a CT scanner to assess its LCD ability. We shall thus introduce another LCD phantom to complement the existing ones, such as Catphan. The cylindrical phantom is constructed using Poly (methyl methacrylate) (PMMA), with craters (holes) having dimensions that gradually increase from 1.0 to 12.5 mm penetrated in configuration that extend from the centre to the corner. Each line of the drilled holes in the phantom is filled with contrast material of specific concentrations. As opposed to the phantom of low detail contrast used in planar imaging, the iodine (contrast material) in this phantom replaces the depth of the phantom holes. The iodine could be reduced to 0.21 milli-Molar (mM) and can be varied for the next line of holes by a small increment depending on the required level of contrast detectability assessment required.

Keywords: CT; contrast detail phantom; radiology; IQF; low contrast detectability
1. Introduction

Computed tomography (CT) is a powerful imaging modality extensively used in diagnostic facilities universally since its invention in 1971 [1,2]. CT scanners nowadays are used comprehensively for the diagnosis of a wide range of pathologies. Recent advances in CT technology enabled a more comprehensive anatomical coverage in shorter scan time [3]. As a result, the CT scan becomes the prime source of ionising radiation in medical imaging and constitutes up to 70% of collective radiation doses [4,5]. The effective dose from vascular CT procedures may exceed 100 mSv [2]. The main superiority of CT imaging systems matched to conventional radiography (planar radiography) is its capacity to provide low contrast resolution radiographs, which enables the detection of low contrast pathologies [6]. CT vendors are working to design facilities that reduce high doses associated with CT utilisation. Concerns about high radiation doses, especially for vulnerable patients such as infants and pregnant women [7], continuously arise without compromising the sensitivity of diagnostic information [8,9]. In recent years, many authors reported that patients are receiving unnecessary radiation doses during imaging. Previous studies showed that more than 1% of the patients receiving an effective dose exceeding 100 mSv within five years, and cancer incidence due to CT procedures might reach 2% [10,11]. Therefore, proper optimization of radiation doses while retaining the image quality is recommended. Optimization could be achieved by choosing exposure parameters and imaging protocol to provide the necessary diagnostic findings. Since the introduction of hybrid imaging combination of transmission imaging (CT) with emission imaging (single-photon emission computed tomography (SPECT) and positron emission tomography (PET)) was developed to provide accurate physiological and morphological findings. A recent study showed that CT constitutes up to 80% of patients’ doses during hybrid imaging techniques [12]. Therefore, an extensive study of the CT image quality parameters to optimize CT image quality is required to assure that patients’ benefit from the procedure outweighs the projected cancer risk.

Recent studies showed that adopting proper imaging protocols may reduce the patients’ doses up to 40% while maintaining the diagnostic findings [13–15]. Mechanisms for optimization for different patient conditions include selecting the appropriate tube potential (kVp), which impacts the contrast of the imaged subject since the tissue linear and mass attenuation coefficients are reliant on the radiation energy and contrast materials. This may increase the Hounsfield unit (CT number) at lesser tube voltage (kVp) [16–19]. The effect of choosing a low X-ray tube voltage (kVp) for low-contrast (LC) objects detection has been deliberated in the literature [20–23]. LCD can be defined as the capacity of the CT machine to discriminate materials with analogous attenuation characteristics such as soft tissues. It is thus a measure of resolution ability which can, in turn, be influenced by the noise of the image, particularly if the substance (object) contrast is small. Low contrast detection is usually quantified by radiographs acquired from the specific phantom. This is followed by assessing images, usually performed by human observers, for instance, radiologists, to determine the most miniature visually differentiable objects with the lowest contrast. The Catphan phantoms are usually made of Polymethyl Methacrylate (PMMA) together with materials of varied densities are inserted in them. It comprises three sequences of nine rod-shaped each, with diameters extending from 2.0 to 15.0 mm, and with only three levels of contrast (1%, 0.5% and 0.3%), limiting its utilization. These contrast levels are expressed as a percentage according to the manufacturer’s specifications. They are defined as the disparity in CT number between the average pixel values quantified on a region of interest (ROI) positioned on a 15.0 mm substance (object) and a sealed background area of equal dimension, devised by ten [24]. Traditionally, phantoms such as Catphan help in assessing only small body parts with wide contrast variations, such as the head. The proposed new design aims to extend detectability by using various contrast media concentrations and thus subjecting contrasts in the cylindrical holes. This, in turn, offers a more significant number of objects of various subject contrasts. Moreover, the levels of tested subject contrast can be varied by the users, which allow them to choose the levels
of subject contrast tested. Such phantoms will meet the demands of the radiology world. These demands can be met if phantoms are designed to include different sized widths, containing various contrast material (iodine) concentration using a larger phantom design to mimic the abdomen or chest cases. Such phantoms that have the ability to represent the different tissue types in the body will increase the ability to test contrast variation for the CT image, particularly while evaluating a tiny dissimilarity in object contrast. With such facilities, the proposed contrast-details phantom can evaluate the radiographic image quality both in planar imaging and today’s sophisticated computed tomography. It contains the geometrical character of the body parts, such as abdomen or chest, to provide in detail data through the “z” axis compared to the well-known contrast detail radiography (CDRAD) phantom commonly used in radiography, which acquires images in two dimensions only. This research provides information on a newly designed phantom: CT-Contrast detail phantom (CDP), which can detect low contrast detail (LCD) for quality assurance purposes. The mean for validation was using the image quality factor tool via visualization by ten assessors. CT-CDP employs a modification of CDRAD for further application in most CT imaging conditions.

2. Materials and Methods

2.1. CT-Image Contrast Detail Phantom

The CT phantom (CDP) is manufactured of PMMA with drilled holes whose diameters increase gradually and diagonally from the centre where the diameter (d) = 1 mm to the edge where “d” = 12.5 mm following the pattern 1, 2.5, 5, 6, 7.5, 9, 10, 11 and 12.5, respectively, as shown in Figure 1. The variation in diameters of the holes enabled assessing contrast-detection capabilities of the CT scanner at a wide dynamic scale of spatial resolutions. CT-CDP is a cylindrical phantom of 15 cm radius and 20 cm thickness. CT scans were acquired across the depth of the phantom. The holes were concentrated around the centre of the cylinder across a depth of 4 cm, as shown in Figure 2. These holes were filled with iodine contrast (milli-Molar (mM) in water with concentrations that generally range from 2–4.1% to resemble different x-ray attenuation conditions and vary according to the CT testing needed. CT machine’s ability to precisely image the tissue or object with variable contrast and resolution properties was evaluated through the phantom’s CT slice.

![Figure 1](image_url). Shows the holes pattern in the Polymethyl Methacrylate (PMMA) CT phantom. The blue circle shows the centre from which rays of holes extend from the centre to the phantom periphery filled with 3.2% iodine concentration. The red oval shows the variation in holes size from 1 mm to 12.5 mm.
2.2. Image Acquisition

CT scans were taken at The Alfred Hospital—Melbourne—Australia using Discovery CT590 RT (General Electric (GE) Healthcare, USA) scanners. Multi-slice CT images were obtained at 140.0 kVp tube potential and 13.0 mAs tube current-time product by spiral mode following Kalra et al. [24] and using pelvis protocol with a 512 × 512 mm field of view (FOV) and 1.25 mm slice thickness. The reconstruction protocol for image acquisition was based on a filtered back-projection technique with a field size of 50 cm. The phantom with contrast-filled holes located around the centre was situated on the CT couch, and the laser beam was straightened out with the CT phantom’s central point. The holes within the phantom were then filled; as per hole-sets covering from the midpoint to the periphery, with different concentrations of iodine contrast: 2.1%, 2.2%, 2.3%, 2.4%, 2.6%, 2.8%, 2.9%, 3.2%, 3.7% and 4.0%. This is displayed in Figure 2A,B.

2.3. Scoring Process

Following the procedures for scoring in conventional radiography, e.g., used with CDRAD phantom CT-CDP, ten specialized medical professionals were subjectively evalu-
ated: 6 radiographers, 2 medical physicists, and 2 radiologists. Ten participants may be considered a small number for assessment, but they were enough in our case to establish proof of functionality of our designed phantom. Each assessor read the radiographic images obtained from phantom exposure separately, using a grayscale medical-grade monochrome (MGM) due to its greater luminescence than standard colour monitor [25]. The researchers obtained Institutional Review Board (IRB) approval from the human ethics advisory committee, RMIT University, Australia. LCD show, in a dark area, to point out the faded discrete disk radiographs for each regular width. The lowermost-contrast observable hole for specifically established width was used as an indicator. These indicators were averaged for each assessor, adopted for computing of the image quality factor (IQF), as illustrated in Equation (1):

$$\text{IQF} = \sum_{i=1}^{9} C_i \cdot D_{ij}$$  \hspace{1cm} (1)

A calculated image quality factor was used to assess the minimum detection limit value (MDL) for each of the ten participants: “i” and “j” stand for rows and columns, respectively. “D_{ij}” represents the cylinder diameter (mm) (the blue circle in Figure 1), and “C_i” is the contrast material concentration (mM) as illustrated in Figure 1 (see the red oval). The IQF is the calculation of the product of depth (C_i) and the visible diameter (D_{ij}) for all columns. Equation (1) has many similarities with that used for image observation in conventional radiographic images using CDRAD. The only difference lies in the fact that the IQF for conventional radiography relates to the cylinder width and its equivalent depth. In contrast, the depth was substituted by contrast material with a particular concentration concerning the CTCDP phantom.

3. Results and Discussion

Each observer scored the images such as the one shown in Figure 3 to determine the IQF for an individual hole in the phantom for all the observers. The IQF average was then calculated for each image of the barely observable hole recorded by each observer. Table 1 shows the data of one observer for holes with various diameters and different contrast. Each observer scored three images obtained under similar conditions. For each hole-diameter, the limit of concentration of contrast media making the image of the hole barely observable is indicated in Table 1. The limit of the lowest concentration in the hole of a specific diameter is multiplied and used as indicated in Equation (1) to determine the IQF value. The average value of the IQF for each of the ten observers was obtained, as displayed in Table 2. Detecting the smallest diameter hole (1 mm) was difficult for most participants. This evidently defines the particular resolution detection limits for the used CT system. IQF value improved with increasing diameter, as shown in Figure 2.

Table 1. The derived image quality factor from the mean values of the image scores verified by single assessors for iodine concentrations (IC, mM) and diameters (D, mm).

| D    | Image | ICD | D×IC | D | Image | ICD | D×IC | D | Image | ICD | D×IC | D |
|------|-------|-----|------|---|-------|-----|------|---|-------|-----|------|---|
| 1.00 | No detectable hole | 0.0000 | 1.00 | No detectable hole | 0.0000 | 1.00 | No detectable hole | 0.0000 |
| 2.50 | 0.0400 | 0.1000 | 2.50 | 0.0400 | 0.1000 | 2.50 | 0.0400 | 0.1000 |
| 5.00 | 0.0370 | 0.1850 | 5.00 | 0.0320 | 0.1600 | 5.00 | 0.0370 | 0.1850 |
| 6.00 | 0.0280 | 0.1680 | 6.00 | 0.0320 | 0.1920 | 6.00 | 0.0290 | 0.1740 |
| 7.50 | 0.0280 | 0.2100 | 7.50 | 0.0290 | 0.2180 | 7.50 | 0.0290 | 0.2180 |
| 9.00 | 0.0260 | 0.2340 | 9.00 | 0.0280 | 0.2520 | 9.00 | 0.0280 | 0.2520 |
| 10.00 | 0.0260 | 0.2600 | 10.00 | 0.0240 | 0.2400 | 10.00 | 0.0260 | 0.2600 |
| 11.00 | 0.0230 | 0.2530 | 11.00 | 0.0240 | 0.2640 | 11.00 | 0.0230 | 0.2530 |
| 12.50 | 0.0230 | 0.2880 | 12.50 | 0.0230 | 0.2880 | 12.50 | 0.0230 | 0.2880 |
| Image quality factor | 1.6980 | Image quality factor | 1.7130 | Image quality factor | 1.7290 |
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![Figure 3](image-url) It displays a typical slice from the phantom scan used in the process of scoring.

**Table 2.** The mean values of image quality factor (IQF) for the ten assessors and overall mean IQF value.

| Assessors | Mean IQF       |
|-----------|----------------|
| 1         | 0.61 ± 0.01    |
| 2         | 0.65 ± 0.09    |
| 3         | 0.62 ± 0.03    |
| 4         | 0.63 ± 0.05    |
| 5         | 0.69 ± 0.04    |
| 6         | 0.73 ± 0.01    |
| 7         | 0.71 ± 0.06    |
| 8         | 0.71 ± 0.02    |
| 9         | 0.73 ± 0.01    |
| 10        | 0.75 ± 0.00    |
| Overall average | 0.69 ± 0.05 |

This was a clear expression of the subject contrast influence, according to the CT images. The iodine concentration had little influence on the detectability of the holes instead of the diameter size, as indicated in Table 1. As can be seen, the image quality factor for the 9 mm width hole was lower compared the image quality factor for the 10 mm width hole, even though the identical concentration of iodine was used. Table 2 shows the average image quality factor of the ten assessors diverse from 11.610 to 1.750 and illustrates excellent contrast distribution values with slight standard deviation (SD).

These results demonstrate that this modified phantom can pose a valuable means of assessing the low contrast detectability for CT scanners beyond the few points of measurements offered by current phantoms such as Catphan. Additionally, this phantom brings the advantages of using contrast detail phantoms which are typically used in projection radiography into the area of computed tomography.
4. Conclusions
This research presents a new phantom design for CT image quality assessment. A modification of the CTPHAN phantom that proved a success in extending contrast assessment extends to the lower values of detection limits. It improves proportionally with the amount of assessing points. Comparison between the designed phantoms with others was based on the capability to compute the low contrast detectability of objects during CT image modalities. The phantom proved efficient to detect both high subject contrast with large diameters and low subject contrast with small diameters. Therefore, it may be used to assess the abilities of CT scanners to differentiate trivial low contrast elements without compromising high spatial resolution allowing extensive utilization in CT scanning.

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