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Can realistic liver tissue surrogates accurately quantify the impact of reduced-kV imaging on attenuation and contrast of parenchyma and lesions?

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Abstract

Rationale and Objectives—To assess if a liquid tissue surrogate for the liver (LTSL) can emulate contrast-enhanced liver parenchyma and lesions and quantify the impact of reduced-kV imaging as a function of lesion contrast, phase of enhancement, and phantom size.

Materials and Methods—First, CT attenuation of LTSL- and water-iodine solutions was measured as a function of iodine concentration and tube potential. For each solution, the iodine concentration was determined to emulate liver parenchyma at 120 kV. CT attenuation for both solutions was predicted for different tube potentials and compared to published patient data. Second, liver parenchyma in late arterial phase (LA: +92HU at 120kV) and portal venous phase (PV: +112HU at 120kV) was emulated using LTSL-iodine and a two-size phantom. Fourteen setups of hyper- and hypoattenuating lesions (lesion-to-parenchyma contrast (C_{LP}) = -50 to +50HU) were created. Each combination of C_{LP} , phase, and size was imaged at 80, 100, 120, and 140 kV at constant radiation dose. CT attenuation, C_{LP} , and lesion-to-parenchyma contrast-to-noise ratio (CNR_{LP}) were assessed and compared to a theoretical model.

Results—LTSL-iodine more accurately emulated the CT attenuation of liver parenchyma across different tube potentials compared to water-iodine solutions. The theoretical model was confirmed by the empirical measurements using LTSL-iodine solutions: attenuation, C_{LP} , and CNR_{LP} increased when the tube potential decreased ($P < 0.001$). This trend was independent of lesion contrast, phase, and size. The absolute improvement in C_{LP} and CNR_{LP} , however, was inversely related to the magnitude of C_{LP} at 140kV.

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Conflicts of Interest

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Conclusion—LTSL accurately emulated the energy-dependent CT attenuation characteristics of contrast-enhanced liver parenchyma and lesions. The relative improvement in C_{LP} and CNR_{LP} by applying reduced-kV imaging was independent of lesion contrast, phase, and size while the absolute improvement decreased for low-contrast lesions.

Keywords

Computed tomography; Liver; Lesion; Tube Potential; Liquid tissue surrogate

Introduction

Contrast-enhanced computed tomography (CT) has been a central tool in oncologic imaging which facilitates lesion detection, lesion characterization and assessment of tumor treatment response. In CT, lesions are usually detected and characterized based on their CT attenuation in Hounsfield units (HU), their lesion-to-parenchyma contrast (C_{LP}), and/or their attenuation changes during different phases of parenchymal enhancement. Effort has been made to define CT attenuation thresholds that improve lesion characterization and help to differentiate benign from malignant disease. Traditionally, these thresholds are based on CT scans at 120 kV.

In recent years, there has been a trend towards imaging at tube potentials lower than the typical 120 kV in contrast-enhanced CT (reduced-kV imaging). Reduced tube potentials increase the photoelectric effect and decrease Compton scattering and therefore increase the CT attenuation of iodinated structures¹. This results in increased contrast between tissues with different iodine uptake, for example between a lesion and the surrounding parenchyma. Because the CT attenuation of contrast-enhanced structures is energy dependent, reduced-kV imaging leads to changes in the CT attenuation of organ parenchyma as well as in the C_{LP} and contrast-to-noise ratio CNR_{LP} between the parenchyma and lesions. This may lead to mischaracterization and misinterpretation in clinical practice, if the diagnosis is based on the quantitative measurement of these parameters and traditional attenuation thresholds are applied. This is particularly important in liver imaging, where reduced-kV imaging has been shown to improve lesion detectability and conspicuity². Furthermore, the effect of reduced-kV imaging might not be the same for different phases of enhancement of liver parenchyma (e.g. late arterial phase versus portal venous phase) as well as for hyperattenuating versus hypoattenuating lesions.

Therefore, it is desirable to estimate the effect of reduced-kV imaging on the attenuation characteristics of liver parenchyma and lesions as a function of lesion contrast and phase of parenchymal enhancement.

In phantom studies, parenchyma or lesions are often represented using a solution of distilled water or saline with iodinated contrast medium³⁻⁵. However, this method does not reflect the energy dependence of the CT attenuation of liver parenchyma. FitzGerald et al. have recently proposed a method for producing liquid tissue surrogate (LTS) materials which more faithfully represent human tissues⁶. These surrogates mimic the elemental composition and density of several human soft tissues, including the liver parenchyma (LTSL), and could improve the scientific investigation of kV-dependent imaging⁷.

Therefore, the purpose of our study was to validate the use of LTSL for reduced-kV imaging by assessing the quantitative effect of reduced-kV imaging on HU, C_{LP} , and CNR_{LP} as a function of tube potential, lesion contrast, and phase of parenchymal enhancement compared to a theoretical model and published patient data.

Materials and methods

Study Design

Two different phantom experiments were conducted. In the first experiment, we measured the CT attenuation of LTSL-iodine and water-iodine solutions as a function of iodine concentration and tube potential. Based on these measurements, we calculated the iodine concentration of LTSL-iodine and water-iodine solutions required to produce the CT attenuation of patients' liver parenchyma at 120 kV. We then calculated the predicted CT attenuation for each of those solutions at tube potentials of 80, 100, 120, and 140 kV and compared them to published patient data ”.

In the second experiment, based on the results of the first experiment, we emulated the CT attenuation of contrast-enhanced liver parenchyma in each of the late arterial and portal venous phases using LTSL-iodine solutions. We serially scanned a phantom filled with LTSL-iodine representing liver parenchyma in each phase together with additional LTSL-iodine solutions representing relevant lesions of varying C_{LP} : hyperattenuating lesions for the late arterial phase and hypoattenuating lesions for the portal venous phase. The effect of reduced kV imaging on C_{LP} and CNR_{LP} was assessed and compared to a theoretical model.

Experiment 1:

CT attenuation characteristics of water-iodine and LTSL-iodine solutions: A liquid tissue surrogate for liver (LTSL) was created by our Department of Chemistry using the formula described by FitzGerald et al. . In a first step, we investigated the CT attenuation measured at different tube potentials resulting from different concentrations of iodinated contrast medium in either distilled water or the proposed LTSL. Distilled water and LTSL were each mixed with iopamidol (300 mg iodine per mL, Isovue®-300, Bracco Diagnostics, Princeton, NJ) to achieve eight iodine concentrations ranging from non-iodinated water or LTSL to approximately 8 mg iodine per mL solutions of water or LTSL. We separately placed the 16 samples in 50-mL vials and simultaneously scanned them on a 64-detector-row CT scanner (Discovery CT 750 HD, GE Healthcare, Chicago, IL, USA) in single-energy mode at 80, 100, 120 and 140 kV using the large body bow-tie filter and helical mode. The tube current-time product (mAs) was adjusted for each tube potential to achieve a constant volume CT dose index ($CTDI_{vol}$) of 6.2 mGy. This dose value was considered to be sufficient enough to image the vials in air. The scanner was calibrated in air immediately before scanning each setup. Images were reconstructed at a 1.5-mm slice thickness using filtered back projection and a standard kernel. CT attenuation was measured from the reconstructed images by drawing circular regions of interests (ROIs) in the center of each vial. The diameter of each ROI was 4 mm (area = 12.6 mm²). For each solution and tube potential, the mean value of the measurements from ten adjacent slices in the center of the vial was calculated. Using Excel (Microsoft Office 365 ProPlus, Microsoft, Redmond, WA,

USA), we fitted linear equations to the averaged measurements of the CT attenuation versus iodine concentration for the LTSL iodine solutions and for the water-iodine solutions at each tube potential. The resulting equations allowed us to estimate the iodine concentration required to produce a given CT attenuation at a given tube potential using distilled water or LTSL.

Experiment 2:

Emulation of liver parenchyma and liver lesions: A custom-made phantom (Fig. 1) was fabricated using a 3-D printer (Connex 3, Objet 350, Stratasys, Eden Prairie, MI, USA). This phantom consists of a fillable chamber (1.9 liter) and 15 separately fillable cylinders of different sizes (diameters of 5 mm – 20 mm) and locations. The chamber emulates the hepatic parenchyma and is shaped to fit within one or two (7.7-cm and 15.6-cm total thickness, respectively) adipose-equivalent plastic rings to emulate a medium-sized (effective diameter 28.3 cm) and large-sized (effective diameter 36.2 cm) patient. The effective diameter was estimated as a geometric mean of the lateral and anterior-posterior dimensions of the phantom .

Based on the linear equations of experiment 1, we created LTSL-iodine solutions that mimic the CT attenuation of in vivo liver parenchyma. For the late arterial phase, we aimed to achieve the previously-published CT attenuation of 120, 106, 91, and 86 HU at 80, 100, 120, and 140 kV, respectively . For the portal venous phase, published data on CT numbers at 80 kV and 100 kV are not available because 120 kV is currently used as the routine in most clinical practices. Therefore, we aimed for the average CT attenuation of 112 HU at 120 kV that is within the range of published data .

First, the main chamber of the phantom was filled with the LTSL-iodine solution that emulates liver parenchyma in the late arterial phase (setup A). We created additional LTSL-iodine solutions with slightly higher iodine concentration than the parenchymal solution to emulate hyperattenuating lesions with a nominal C_{LP} of +5, +10, +15, +20, +30, +40, and +50 HU. The lesion-representing cylinders of the phantom were filled with these additional solutions. Of the 15 cylinders, 3 were filled with the same solution as the main chamber to emulate “neutral” control lesions (C_{LP} nominally 0), 1 cylinder was filled with distilled water, and, sequentially, the remaining 11 cylinders were all filled with the same one of the 7 solutions described above to emulate hyperattenuating lesions (setups A1-A7).

Second, the main chamber of the phantom was filled with the LTSL-iodine solution formulated to emulate the portal-venous phase (setup B). We created additional LTSL-iodine solutions mimicking hypoattenuating lesions with a nominal C_{LP} of -5, -10, -15, -20, -30, -40, and -50 HU. We filled the phantom’s lesion-representing cylinders using the same approach as for setup A (setups B1 to B7).

CT Scanning

Each of the overall 14 different setups was scanned separately using the medium and the large phantom sizes at 80, 100, 120, and 140 kV, resulting in 112 different CT scans. The scan conditions of Experiment 1 were used except that the tube current-time product (mAs)

was adjusted to achieve a constant CTDI_{vol} of 10 mGy in the medium-sized phantom and 15 mGy in the large-sized phantom. These radiation dose values were chosen according to the recently published dose ranges for abdominal CT of the ACR Dose Index Registry . The gantry rotation time as well as the pitch had to be reduced for certain reduced-kV settings to reach the desired CTDI_{vol}. The variable scanning parameters are summarized in Table 1. The scanner was calibrated in air immediately before scanning each setup. All images were reconstructed at a slice thickness of 5 mm and an increment of 2.5 mm using an adaptive statistical iterative reconstruction (ASiR, GE Healthcare, Chicago, IL, USA) algorithm at 40%blending with filtered back projection (FBP).

Quantitative Assessment of C_{LP} and CNR_{LP}

CT attenuation was measured by drawing ROIs in each of the 11 lesion-emulating cylinders and in the adjacent parenchyma. The measurements were performed using ten consecutive slices at the center of the phantom length. A MATLAB (The Mathworks, Natick, MA, USA) script was developed and used to automate the measurements and standardize the position and size of the ROIs among all images. The diameter of each ROI was 4 mm (area = 12.6 mm²) for the cylinders and 20 mm (area = 314 mm²) for the adjacent parenchyma. We chose to exclude the 5-mm and 6-mm cylinders from analysis in order to minimize potential measurement errors produced by the wall of the cylinders. For each cylinder of each dataset, lesion-to-parenchyma contrast (C_{LP}) was calculated as $\text{abs}(\text{ROI}_L - \text{ROI}_P)$, where ROI_L is the mean CT attenuation in the lesion ROIs averaged over ten slices and ROI_P is the mean CT attenuation of the adjacent parenchyma ROIs averaged over ten slices. Lesion-to-parenchyma contrast-to-noise ratio (CNR_{LP}) of each cylinder was calculated as C_{LP} / N , where N is the standard deviation of the CT attenuation in the parenchyma ROIs, averaged over ten slices.

Theoretical impact of reduced-kV imaging on C_{LP} and CNR_{LP}

CT attenuation is proportional to iodine concentration and directly related to the x-ray energy spectrum, which is dependent on the tube potential . Therefore, the CT attenuation of a lesion (L) at a certain tube potential V_L can be described by the linear equation:

$$L = m * I_L + b_L$$

where m represents the slope of the CT attenuation-to-iodine concentration function in HU/mg iodine/mL for a certain V_L, I_L represents the iodine concentration of the lesion in mg/mL, and b_L represents the zero-iodine-concentration intercept coefficient for the lesion tissue in HU. Similarly, the CT attenuation of the adjacent parenchyma at V_L can be described as:

$$P = m * I_P + b_P$$

where I_P represents the iodine concentration of the parenchyma in mg iodine/mL, and b_P represents the zero-iodine-concentration intercept coefficient for the parenchymal tissue in HU. Here, we assume that the slope m for liver parenchyma and liver lesion is the same (which is most likely reasonable for well-differentiated hepatocellular carcinomas).

For a given case, C_{LP} at a given tube potential V_L is therefore:

$$C_{LP}(V_L) = \text{abs}(L - P) = \text{abs}(m * I_L + b_L - (m * I_P + b_P)) = \text{abs}(m * (I_L - I_P) + (b_L - b_P))$$

In other words, C_{LP} at a certain tube potential depends on i) the slope of the CT attenuation-to-iodine concentration curve for that specific tube potential, ii) the difference in iodine concentration in the lesion and the parenchyma, and iii) the difference between the unenhanced attenuation of the lesion and the parenchyma. For liver tumors for which (iii) is nearly 0, the b terms cancel. This is the case in our experiment, because unenhanced parenchyma and lesions are identical – LTSL.

Based on our fitted linear equations of the LTSL solutions, we created a theoretical model estimating the change in C_{LP} at reduced tube potentials of 80, 100, and 120 kV compared to the baseline of 140 kV. Assuming that image noise is rather constant at different tube potentials at constant radiation dose, the same model can be applied to estimate the change in CNR_{LP} . Further, the measured change (compared to 140 kV) in C_{LP} and CNR_{LP} was calculated for all the measured data. Linear fits were done to determine if the measured data corresponded to the theoretical change in C_{LP} and CNR_{LP} that would be expected for 80, 100, and 120 kV compared to 140 kV.

Statistics

A multi-variate statistical model (linear mixed-effects) was applied to all measured data to determine how measured C_{LP} and CNR_{LP} were affected by tube potential, nominal contrast, acquisition phase, and phantom size. Fixed terms were used for tube potential, nominal contrast, acquisition phase, and phantom size, and a conditional random term (tube potential given nominal contrast, assumed normal distribution) was also included. The statistical model was fitted to the data using MATLAB's statistics and machine learning toolbox (fitglme function, maximum pseudo likelihood fitting method). P-values less than 0.00625 were considered significant (0.00625 was chosen using a Bonferroni correction of 0.05 / 8, due to having 2 models with 4 terms each).

Results

Experiment 1:

CT attenuation characteristics of water-iodine and LTSL-iodine solutions—We found that the CT attenuation of LTSL-iodine solutions (Table 2) and distilled water-iodine solutions (Table 3) as a function of iodine concentration and tube potential were substantially different. Figure 2 shows the linear fits for the LTSL-iodine and water-iodine solutions at all four tube potentials. The baseline CT attenuation of LTSL without the addition of iodine was 62 HU at 120 kV which closely matches the CT attenuation of liver parenchyma in non-contrast CT.

We calculated the required iodine concentration of a LTSL-iodine and water-iodine solution to emulate the CT attenuation of patients' liver parenchyma at 120 kV and assessed how the CT attenuation changes at different tube potentials (at constant iodine concentration). We

found that the LTSL-iodine solution closely matched the CT attenuation of published patient data at different tube potentials whereas the water-iodine solution led to a higher attenuation values at reduced tube potentials. Figure 3 shows CT attenuation as a function of kV for LTSL-iodine and water-iodine solutions (at a fixed iodine concentration) to illustrate this result. As can be seen, higher iodine concentrations must be used for water-iodine solutions compared to LTSL-iodine solutions to emulate the CT attenuation of liver parenchyma at a desired baseline tube potential. This increase in iodine concentration will then cause additional overestimation of the CT attenuation at reduced-kV setting.

Experiment 2:

Theoretical impact of reduced-kV imaging on C_{LP} and CNR_{LP} —Using the fitted linear equations of experiment 1, we estimated the theoretical change in C_{LP} and CNR_{LP} at reduced tube potentials of 80, 100, and 120 kV compared to the C_{LP} at 140 kV (model for C_{LP} shown in Figure 4). These data show that C_{LP} and CNR_{LP} increase at reduced tube potentials and the increase is greater for lesions with a high C_{LP} at 140 kV (e.g. 50 HU) compared to lesions with a low C_{LP} at 140 kV (e.g. 5 HU).

Quantitative assessment of C_{LP} and CNR_{LP} —The results of the quantitative analysis confirmed the realism of our theoretical model, which predicts that C_{LP} and CNR_{LP} improve when reducing the tube potential and that the absolute improvement is inversely related to the lesion contrast. Our first finding from the empirical result was that tube potential had a significant impact on C_{LP} and CNR_{LP} ($P < 0.001$ for both parameters): both parameters increased when reducing the tube potential from 140 kV to 80 kV. When averaged over both phantom size and all lesion sizes, C_{LP} at 140, 120, 100, and 80 kV was 16, 19, 24, and 31 HU, respectively, for the late arterial phase and 15, 18, 22, and 29 HU, respectively, for the portal venous phase. Similarly, CNR_{LP} at 140, 120, 100, and 80 kV was 0.7, 0.9, 1.1, and 1.4, respectively, for the late arterial phase and 0.7, 0.8, 1.0, and 1.3, respectively, for the portal venous phase. Our second finding was that the increase in C_{LP} and CNR_{LP} by reducing the tube potential from 140 kV to 80 kV was higher for high-contrast lesions compared to low-contrast lesions. Figure 5 and Figure 6 show the measured C_{LP} and the measured CNR_{LP} for each tube potential, nominal lesion contrast (defined at 120 kV), phase of enhancement, and phantom size. For both acquisition phases and phantom sizes, the increase in C_{LP} and CNR_{LP} when reducing the tube potential from 140 kV to 80 kV increases with nominal C_{LP} . In Figure 7, we further plotted the increase in CNR_{LP} for 80, 100, and 120 kV compared to 140 kV versus the measured C_{LP} ; we applied a linear fit of the data points for each tube potential. The linear fits demonstrate that the change in CNR_{LP} at reduced tube potentials is proportional to C_{LP} at 140 kV.

Discussion

Our study investigated the utility of a liquid tissue surrogate to emulate the CT attenuation, C_{LP} , and CNR_{LP} of liver parenchyma and lesions as a function of tube potential, lesion contrast, phase of enhancement, and phantom size. Our results demonstrated that LTSL closely matched the CT attenuation of liver parenchyma in patients across different tube potentials. Because the unenhanced attenuation of water (0 HU) is lower than that of liver

(approximately 60 HU), a higher iodine concentration is required when using a water-iodine solution to emulate the CT attenuation of liver parenchyma at 120 kV. Consequently, a water-iodine solution of the same iodine concentration would result in unrealistically increased CT attenuations at lower tube potentials. Therefore, the use of water-iodine solutions could lead to an overestimation and misinterpretation of the potential benefit of reduced-kV imaging in research studies and consequently to an inappropriate translation into clinical practice. On the other hand, LTSL solution emulates more accurate CT attenuation across tube potentials using only a single iodine concentration. This is preferable to investigate the impact and benefits of reduced-kV imaging and to optimize contrast-enhanced CT protocols. LTSs have important implications in research because solutions for various human body tissues can be created and various different phases of enhancement can be emulated by mixing LTS with various iodine concentrations. In addition, this also has potential important clinical implications because, to date, the characterization of certain pathologies is based on CT number thresholds which were defined at 120 kV. For example, an enhancing renal lesion is defined as a lesion that demonstrates an increase in CT attenuation of more than 15 HU from unenhanced to enhanced phase. However, this threshold is only applicable at 120 kV. Taking into account the fact that CT attenuation of iodinated structures is energy-dependent, this threshold would shift if reduced tube potentials are used. Therefore, current guidelines which apply CT attenuation thresholds suffer from the dilemma that these thresholds are not adjusted to the emerging application of reduced-kV imaging. LTS solutions offer the possibility to emulate specific tissue and tissue pathologies like lesions and to investigate and define preliminary energy-dependent CT attenuation thresholds which could then be validated in clinical studies. In addition, LTS solutions are more cost-effective compared to plastic tissue-equivalent phantoms, especially in the evaluation of contrast-enhanced tissues. Furthermore, LTS solutions could be used to perform dynamic experiments like perfusion studies.

In the second part of our study we assessed the feasibility of LTSL in the quantification of C_{LP} and CNR_{LP} of lesions at different lesion contrast and different phases of parenchymal enhancement as a function of tube potential and compared these results to a theoretical predictive model. We investigated the improvement of CNR_{LP} relative to 140 kV as the baseline because 140 kV had the theoretical lowest CNR values. The empirical quantitative measurements using LTSL demonstrated equal trends compared to the predictive model. The results confirmed that reduced-kV imaging improved C_{LP} and CNR_{LP} for both hyperattenuating lesions in the late arterial phase and for hypoattenuating lesions in the portal venous phase. Interestingly, this improvement was independent of the phase of enhancement of the liver parenchyma and also independent of whether a lesion was hyper- or hypoattenuating, but depended only on the difference in iodine concentration between a lesion and the surrounding parenchyma. For example, the improvement in CNR_{LP} by reducing the tube potential from 140 kV to 80 kV was equal for a hyperattenuating lesion of +40 HU at 140 kV in late arterial phase and for a hypoattenuating lesion of -40 HU at 140 kV in portal venous phase because both lesions have the same difference in iodine concentration compared to their surrounding parenchyma.

Former studies have described the improvement in quantitative image quality parameters like C_{LP} and CNR_{LP} by reduced-kV imaging. Compared to these studies, we also

investigated the impact of different lesion characteristics and parenchymal phases of enhancement. Our findings demonstrate that the absolute improvement in C_{LP} and CNR_{LP} when using reduced-kV imaging depends on the initial C_{LP} at the baseline tube potential. For instance, a lesion with a C_{LP} of 5 HU at 140 kV shows less improvement in C_{LP} with reduced-kV imaging compared to a lesion with a C_{LP} of 20 HU. If we interpret this finding in a clinical context, the visibility of lesions that have high C_{LP} at 140 kV improves more when using reduced-kV imaging compared to the visibility of lesions with low C_{LP} at 140 kV. This is especially interesting considering that one of the main purposes of reduced-kV imaging is to improve the visibility and conspicuity of lesions which are difficult to depict at higher tube potentials. In other words, lesions that are difficult to visualize because they have a low C_{LP} at higher tube potentials of 120 kV or more may not demonstrate a significant improvement in conspicuity when using reduced tube potential imaging. Based on our work, further studies are needed to define the threshold of initial C_{LP} for which reduced tube potential imaging is beneficial on lesion's visibility.

Several potential limitations of our study merit consideration. First, we did not have access to patient data on CT attenuation of the liver parenchyma in the portal venous phase at 80 kV and 100 kV because 120 kV is routinely used in most clinical practices. However, we adjusted the CT attenuation of our emulated LTSL solution to match patient data at 120 kV from a previous study population. Second, we used LTSL-iodine solutions as a straightforward alternative to water-iodine solutions. LTS for blood is also available. A future study could use an appropriate mixture of LTS liver and LTS blood for unenhanced liver, and add appropriate concentrations of iodine to represent contrast-enhanced liver parenchyma and lesions. Because LTS liver and LTS blood have similar linear attenuation coefficients for clinically-used CT spectra, we believe this would be a second-order effect in our study. Third, we did not perform a human observer or model observer study to grade the improvement of visibility and conspicuity of the lesions by reducing the tube potential. Previous studies have shown that the detectability of liver lesions in clinical routine is not necessarily correlated to quantitative image quality parameters alone¹⁷. Model observers can be extremely useful in providing a quantitative measure of conspicuity. They are particularly useful when comparing images which have simultaneous differences in noise texture (noise power spectrum) and resolution. In this study, the noise texture and resolution are not drastically different between the lesions or images being compared. Thus, CNR should be a relatively good surrogate for conspicuity in this case. Furthermore, the main purpose of our study was to validate the use of LTSL to emulate energy-dependent CT attenuation characteristics. Fourth, we did not assess the impact of potential beam hardening at lower tube potentials on CT attenuation (and consequently C_{LP} and CNR_{LP}) or lesion visibility. Especially bone structures can cause beam hardening at low tube potentials in patients and therefore alter image quality parameters. A future phantom study could include bone structures to account for this.

Conclusions

In conclusion, our phantom study confirmed that liquid tissue surrogates for the liver accurately emulate the energy-dependent CT attenuation characteristics of liver parenchyma and liver lesions at different tube potentials. Our results indicate that the relative

improvement in C_{LP} and CNR_{LP} at reduced tube potentials is independent of lesion contrast, phase of enhancement, and phantom size while the absolute improvement is inversely related to the lesion contrast. Visibility of low-contrast lesions, which are most difficult to detect in clinical routine, may be increased far less than high-contrast lesions when reduced-kV imaging is used. Future studies should evaluate the impact of reduced-kV imaging on the accuracy of lesion detection dependent on lesions' contrast properties.

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Abbreviations

CT	Computed tomography
HU	Hounsfield unit
C_{LP}	Lesion-to-parenchyma contrast
CNR_{LP}	Lesion-to-parenchyma contrast-to-noise ratio
LTS	Liquid tissue surrogate
LTSL	Liquid tissue surrogate for liver
$CTDI_{vol}$	Volume CT dose index

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Figure 1.

Photo (left) and axial CT image (right) of the custom-made liver phantom with a fillable chamber (X) and 15 fillable cylinders (white arrow). The chamber was filled with LTSL-iodine solutions to emulate liver parenchyma in late arterial phase (shown, right) or portal venous phase. Eleven of the fifteen cylinders were filled with LTSL-iodine solutions of slightly higher or lower iodine concentrations to emulate hyper- or hypoattenuating liver lesions with different lesion-to-background contrast values (shown, right, with hyperattenuating lesions with C_{LP} of +50 HU). Three cylinders were filled with neutral lesions (N) and one with distilled water (W). 5-mm, 6-mm, neutral, and water lesions were excluded from statistical analysis. One or two rings of fat-equivalent plastic were added to emulate medium- and large-sized patients (one ring is shown).

Figure 2.

Plot of CT attenuation of LTS-iodine solutions (brown) and distilled water-iodine solutions (blue) at tube potentials of 80, 100, 120, and 140 kV. Measured data are plotted as markers; the lines are the fitted data, and the coefficients for the linear fits to each curve are shown in the legend. R^2 values were 0.999 for all linear fits.

Figure 3.

Plot showing the CT attenuation of liver parenchyma in patients (black) and of LTS-iodine solutions (brown) and distilled water-iodine solutions (blue) at different tube potentials as calculated from calibration equations. The iodine concentration in these hypothetical solutions was chosen to produce 91 HU at 120 kV, thereby matching the published value for liver in patients .

Figure 4.

Plot showing theoretical change in lesion-to-background contrast (C_{LP}) at 80 (purple), 100 (light blue), and 120 kV (yellow) compared to 140 kV (baseline) versus the nominal C_{LP} . C_{LP} improves by reducing the tube potential. This improvement is proportional to the nominal C_{LP} of the lesion at 140 kV.

Figure 5.

Box-and-whisker plot showing median values and quartiles of measured lesion-to-background contrast C_{LP} for each tube potential, nominal lesion contrast (defined at 120 kV), phase of enhancement, and phantom size. For all lesions, the median C_{LP} increased by lowering the tube potential from 140 kV to 80 kV. However, this median increase in C_{LP} is higher for lesions with an initial high nominal contrast (for example 50 HU) compared to lesions with an initial low nominal contrast (for example 5 HU).

Figure 6.

Box-and-whisker plot showing median values and quartiles of measured absolute CNR_{LP} for each tube potential, nominal lesion contrast, phase of enhancement, and phantom size. For all lesions, the median CNR_{LP} increased by reducing the tube potential from 140 kV to 80 kV. This trend is similar in the medium phantom results.

Figure 7.

Plot showing the measured change in lesion-to-background CNR_{LP} at 80, 100, and 120 kV relative to 140 kV for each measured lesion. The “CNR change” data are calculated for each measured sample as $CNR(\text{measured}) - CNR(\text{mean for same conditions at 140 kV})$; the data in this plot are prior to taking the absolute value, as was done for all other data in this report. The lines depict linear fits to all the data points for each of 80, 100, and 120 kV. Each data point represents the measured change in CNR_{LP} for one image slice and one cylinder of the phantom.

Table 1.

CT scan parameters for each tube potential using the medium (M) and large (L) patient-sized phantom.

	80 kV		100 kV		120 kV		140 kV	
	M	L	M	L	M	L	M	L
X-ray tube current (mA)	690	605	590	570	370	575	260	405
Gantry rotation time (s)	0.8	1.0	0.5	0.8	0.5	0.5	0.5	0.5
Helical pitch	1.375	0.984	1.375	1.375	1.375	1.375	1.375	1.375

Table 2.

CT attenuation of LTSL-iodine solutions at various iodine concentrations and each tube potential.

Iodine concentration (mg iodine /mL solution)	CT attenuation (HU) of LTSL-iodine solutions			
	80 kV	100 kV	120 kV	140 kV
0.00	77	68	62	62
0.82	123	105	92	86
1.11	138	115	101	94
1.64	163	135	118	108
2.26	194	160	138	126
4.40	296	240	204	181
6.36	385	311	262	232
8.53	477	382	322	283

Table 3.

CT attenuation of distilled water-iodine solutions at various iodine concentrations and each X-ray tube potential.

Iodine concentration (mg iodine /mL solution)	CT attenuation (HU) of distilled water-iodine solutions			
	80 kV	100 kV	120 kV	140 kV
0.00	15	11	6	4
0.60	50	35	24	19
1.08	72	53	40	33
1.74	105	78	61	52
2.20	126	94	75	64
3.91	209	160	129	110
6.32	320	248	201	173
8.06	400	310	252	217