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Effect of various environments and computed tomography scanning parameters on renal volume measurements *in vitro*: A phantom study

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Abstract. Kidney volume is an important parameter in clinical practice, and accurate assessment of kidney volume is vital. The aim of the present study was to evaluate the effect of various environments, tube voltages, tube currents and slice thicknesses on the accuracy of a novel segmentation software in determining renal volume on computed tomography (CT) images. The volumes of potatoes and porcine kidneys were measured on CT images and compared with the actual volumes, which were determined by a water displacement method. CT scans were performed under various situations, including different environments (air or oil); tube voltages/tube currents (80 kVp/200 mAs, 120 kVp/200 mAs, 120 kVp/100 mAs); and reconstructed slice thicknesses (0.75 or 1.5 mm). Percentage errors (PEs) relative to the reference standards were calculated. In addition, attenuation and image noise under different CT scanning parameters were compared. Student's t-test was also used to analyze the effect of various conditions on image quality and volume measurements. The results indicated that the volumes measured in oil were closer to the actual volumes ($P<0.05$). Furthermore, attenuation and image noise significantly increased when using a tube voltage of 80 kVp, while the mean PEs between 120 and 80 kVp voltages were not significantly different. The mean PEs were greater when using a tube current of 100 mAs compared with a current of 200 mAs ($P<0.05$). In addition, the volumes measured on 1.5 mm slice thickness were closer to the actual volumes ($P<0.05$). In conclusion, different environments, tube currents and slice thicknesses may affect the volume measurements. In the present study, the most accurate volume measurements were obtained at 120 kVp/200 mAs and a slice thickness of 1.5 mm.

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Introduction

Kidney volume is an important parameter that must be considered for the selection of appropriate management, surgery planning and evaluation of disease status in patients with kidney disease (1-5). Therefore, precise kidney volumetry has been receiving increasing attention clinically. Typically, kidney size or volume measurement is evaluated using various cross-sectional imaging modalities, including ultrasonography (US), computed tomography (CT) and magnetic resonance imaging (MRI) (6-12). US is an easily used and low-cost technique. Using this method, the renal volume is usually calculated by measuring the three axes of the kidney (length, width and thickness) and then applying these values to an ellipsoid formula (6). However, the shape of the kidney varies greatly in the majority of cases, and as a result, errors in volume calculations by US may occur (6,7). MRI provides outstanding sensitivity for the discrimination of soft tissues that facilitates the segmentation of the kidney (8). However, various factors, such as spatial resolution, artifacts and signal non-uniformities, can affect the accuracy in image segmentation by MRI (9,10). With the advancements in multidetector CT (MDCT), improvements of spatial resolution and time resolution, the ability to assess organ volume using three-dimensional (3D) metrics has become feasible (11,12).

Segmentation of the kidney using CT images is an essential step in renal volumetry. Existing image segmentation methods used in organ volumetry are classified into three categories as follows: The thresholding and adaptive region-growing method (13), deformable model (14), and the knowledge-based method (15).

In the current study, a novel segmentation software based on deformable model was constructed in order to measure kidney volume. However, image segmentation may be affected

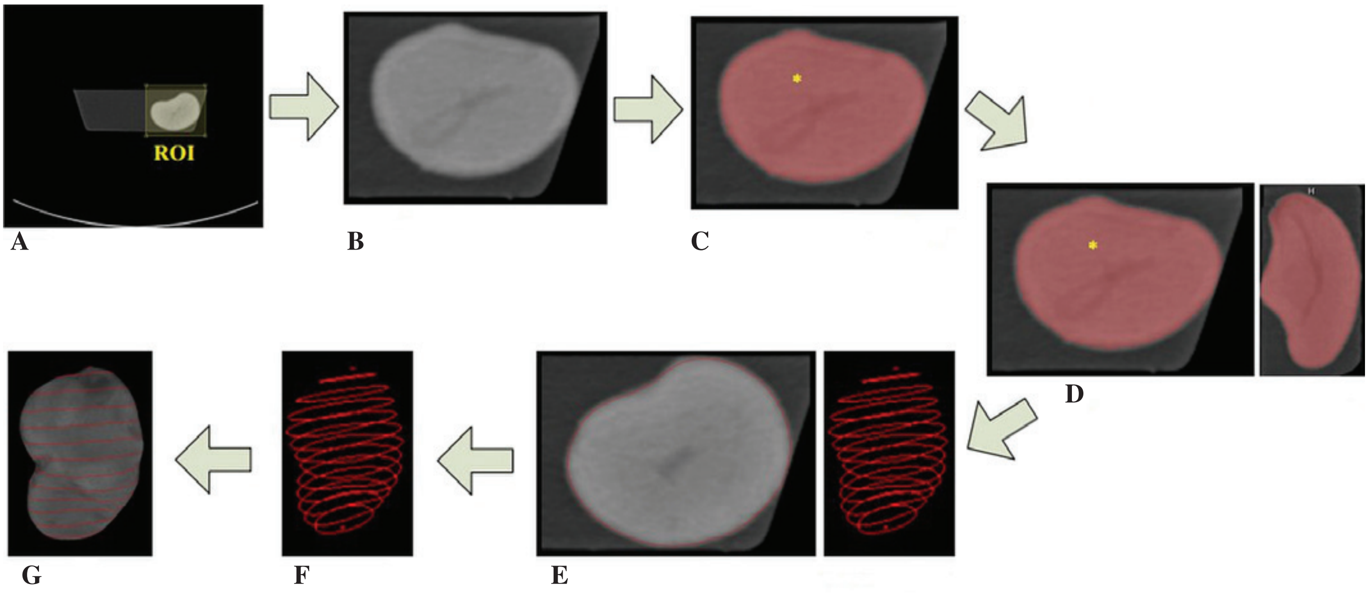

Figure 1. Steps of potato segmentation using CT scanning. (A) Original CT image. (B) Image in the region of interest. (C) Region growing, with the asterisk (*) indicating a seed point. (D) Morphological operators in 2D axial (left) and sagittal (right) slices. (E) Contours of potato obtained using the Snake deformable contour method in 2D slices at the transversal level (left) and a stack of contours of the whole object (right). (F) Contours were manually correct, if needed. (G) 3D reconstruction of the potato. CT, computed tomography; 2D, two-dimensional; 3D, three-dimensional; ROI, rectangular region of interest.

by a number of factors, including slice thicknesses, tube voltages, tube currents and surrounding structures (16-18). The aim of the present study was to investigate the effect of various environments, tube voltages, tube currents and slice thicknesses on the accuracy of renal volume measurements *in vitro*.

Materials and methods

Phantom. A phantom was used as an *in vitro* model of the kidney. Considering the dimensions, shape and density of human kidneys, a total of 12 potatoes and 12 porcine kidneys were selected to simulate human kidneys (similar size and shape to human kidneys). A similar technique has been previously described (19). All surrounding tissues were removed as completely as possible without damaging the integrity of the porcine kidneys.

Water displacement method. Each object was immersed in a container filled with water and the displaced water was measured using a graduated cylinder in order to determine the volume of the object. The volume measured by water displacement method was considered as the actual volume. The measurement was repeated three times, and the mean value was determined for each object.

CT image acquisition. Objects were exposed to the air or immersed in maize germ oil (Xiwang Foodstuffs Co., Ltd., Shangdong, China) in order to simulate the abdomen and surrounding fat of kidney. All CT examinations were performed using a first-generation of dual-source CT scanner with 32 rows of detectors (Somatom Definition AS; Siemens, Munich, Germany). The following scan parameters were used: Tube voltage/tube current parameters included 80 kVp/200 mAs, 120 kVp/200 mAs and 120 kVp/100 mAs; beam collimation, 0.6 mm; and rotation time, 0.33 sec. Images were reconstructed at 0.75 mm and 1.5 mm slice thicknesses, and a medium soft-tissue convolution kernel (B30f) was used (20). Volume measurement by CT and by water displacement method was completed within 2 h.

Volume measurement software. The segmentation method was divided into six steps, as illustrated in Fig. 1, which demonstrates the potato segmentation process. Initially, a rectangular region of interest (ROI) was drawn manually to select the object (potato or porcine kidney) in the 3D CT dataset (CT axial images; Fig. 1A and B). A seed point, as shown by an asterisk in Fig. 1C, was defined in the object in order to initialize a 3D region growing. The seed point served as a trigger point. When the operator entered a seed point in the ROI, the measurement process was initialized. Region growing starts from the seed point inside the ROI and continues until no more pixels satisfy the criteria (similar attenuation) for inclusion in that region. A fixed low threshold (20HU) was applied to extract the object as both potato and porcine kidney CT values are >20HU, regardless of the various situations used during the CT scan in the present study. Next, morphological operators were performed in order to fill the small holes in the extracted regions in the 2D image. This processing was performed in the 2D axial and sagittal slices consecutively (Fig. 1D). An initial binary 3D mask (red regions shown in Fig. 1D) of the object was obtained in this step. Subsequently, a deformable contour method (also known as the 'Snake' model) (21) was performed in each 2D axial slice of the 3D mask image obtained at the previous step. A stack of contours were generated to depict the shape of the object, as shown in Fig. 1E. Once the contours were checked, a 3D region representing the shape of the object was reconstructed by interpolating an implicit surface (22) based on the extracted contours (Fig. 1F and G). The volume of the 3D region (determined using the software tools used in reconstruction) was considered to be the volume of the object.

Table I. Comparison of the mean PEs between measurements performed in air and oil.

Slice thickness	Scan dose[a]	PEs (air), %	PEs (oil), %	P-value
0.75 mm	80 kVp/200 mAs	-3.22±2.23	0.88±1.83	<0.001
	120 kVp/200 mAs	-3.33±2.33	0.80±1.92	<0.001
	120 kVp/100 mAs	-3.15±2.33	0.99±1.85	<0.001
1.5 mm	80 kVp/200 mAs	-3.72±2.31	0.50±1.88	<0.001
	120 kVp/200 mAs	-3.85±2.39	0.45±1.92	<0.001
	120 kVp/100 mAs	-3.69±2.42	0.60±1.91	<0.001

[a]Scan dose indicates the tube voltage/tube current used in the experiments. Data are displayed as the mean ± standard deviation. PE, percentage error.

Table II. Comparison of the mean PEs between tube voltages of 120 and 80 kVp at different slice thickness.

	PE at each tube voltage, %		
Slice thickness	120 kVp	80 kVp	P-value
0.75 mm	0.80±1.92	0.88±1.83	0.240
1.5 mm	0.45±1.92	0.50±1.88	0.421

Based on computed tomography measurements performed in oil. Data are displayed as the mean ± standard deviation. PE, percentage error.

Table III. Comparison of the mean PEs between tube currents of 200 and 100 mAs at different slice thickness.

	PE at each tube current, %		
Slice thickness	200 mAs	100 mAs	P-value
0.75 mm	0.80±1.92	0.99±1.85	0.002
1.5 mm	0.45±1.92	0.60±1.91	<0.001

Based on computed tomography measurements performed in oil. Data are displayed as the mean ± standard deviation. PE, percentage error.

Image quality analysis. Measurements of the objects' CT attenuation and image noise were performed by a single radiologist with 3 years of experience in abdominal CT scanning. The CT attenuation of the potatoes and porcine renal parenchyma were measured in a circular ROI of ~3 cm^2. The CT attenuation of oil was also measured in a circular ROI of ~3 cm^2. Image noise was defined as the standard deviation of CT attenuation of the objects. Contrast noise rate was calculated with the formula: (CT attenuation of kidney - CT attenuation of oil)/image noise. In order to minimize the bias from a single measurement, the mean of three measurements of each ROI was calculated.

Statistical analysis. Percentage error (PE) was used to evaluate the measurement accuracy. The PE and absolute PE were calculated as follows: $PE = (V_{CT} - V_{WD}) / V_{WD}$ x100%, where V_{CT} is the object volume measured on the CT scan and V_{WD} is the object volume measured by the water displacement method. Mean PE values were calculated for the 24 objects under various technique parameters. To assess the difference of the mean PE values and image quality under various situations, Student's t-test was used. The results were considered as statistically significant when the P-value was <0.05. Data were analyzed using SPSS version 13.0 software (SPSS, Inc., Chicago, IL, USA).

Results

PE analysis. Segmentation and volume calculation were successfully performed for all objects. As determined by the water displacement method, the actual volumes of the 12 potatoes ranged between 124 and 420 ml, while the actual volumes of the 12 porcine kidneys ranged between 88 and 163 ml.

Table IV. Comparison of the mean PEs between slice thickness of 0.75 and 1.5 mm at different voltages/currents.

	PE at each thickness, %		
Tube voltage/ tube current	0.75 mm	1.5 mm	P-value
80 kVp/200 mAs	0.88±1.83	0.50±1.88	<0.001
120 kVp/200 mAs	0.80±1.92	0.45±1.92	<0.001
120 kVp/100 mAs	0.99±1.85	0.60±1.91	<0.001

Based on computed tomography measurements performed in oil. Data are displayed as the mean ± standard deviation. PE, percentage error.

The comparison of the mean PEs between the air and oil CT measurements is listed in Table I. Volumes measured in the oil were closer to the actual volumes compared with those measured in the air, regardless of the various slice thicknesses and scan parameters used (P<0.001 for all). The minimal mean PE of objects on volumetric measurements was found to be 0.45±1.92% in the oil, obtained when 120 kVp tube voltage, 200 mAs tube current and 1.5 mm slice thickness were used.

The various technique parameters were then compared for measurements performed in the oil. Further comparison of the mean PEs of object volumetric measurements in the air was not

Table V. Comparison of CT attenuation, image noise and CNR between tube voltages of 120 and 80 kVp.

Slice thickness	120 kVp	80 kVp	P-value
0.75 mm			
CT attenuation (HU)	66.81±17.25	70.59±16.34	<0.001
Image noise	8.83±3.12	10.70±2.69	<0.001
CNR	23.89±6.82	22.61±4.92	0.070
1.5 mm			
CT attenuation (HU)	65.59±16.91	69.67±15.91	<0.001
Image noise	8.18±3.30	9.28±2.94	<0.001
CNR	26.60±9.01	25.81±6.27	0.359

CNR, contrast noise rate; HU, Hounsfield unit. Data are displayed as the mean ± standard deviation.

Table VI. Comparison of CT attenuation, image noise and CNR between tube currents of 200 and 100 mAs.

Slice thickness	200 mAs	100 mAs	P-value
0.75 mm			
CT attenuation (HU)	66.81±17.25	66.90±17.02	0.730
Image noise	8.83±3.12	9.67±2.93	<0.001
CNR	23.90±6.82	21.22±5.09	<0.001
1.5 mm			
CT attenuation (HU)	65.59±16.91	66.00±16.78	0.133
Image noise	8.18±3.30	8.66±3.04	0.007
CNR	26.60±9.01	24.13±6.36	0.009

CNR, contrast noise rate; HU, Hounsfield unit. Data are displayed as the mean ± standard deviation.

performed due to the higher accuracy of measurements obtained in the oil environment, according to the aforementioned results. Table II shows the mean PEs of objects on volumetric measurement obtained at different tube voltages. The results indicated that no statistically significant differences in PEs between the 120 and 80 kVp tube voltages, regardless of the different slice thickness (P>0.05 for both). Therefore, in this study, different tube voltages have very little influence on volume measurements. Similarly, Table III shows the comparison of the mean PEs between different tube currents. At slice thicknesses of 0.75 and 1.5 mm, the PEs were significantly larger when a current of 100 mAs was used, compared with the use of a 200 mAs current (P<0.001). Hence, it was concluded that lower tube current would reduce the accuracy of volume measurements in this study. In addition, the mean PEs of the two slice thicknesses were compared at different tube voltages and currents. Table IV shows that the volumes measured using a slice thickness of 1.5 mm were closer to the actual volumes, in comparison with measurements at 0.75 mm thickness (P<0.05). In this study, a volume measured at 1.5 mm thickness is more accurate than a volume measured at 0.75 mm thickness, regardless of what tube voltages/currents are applied.

Image quality analysis. Results regarding image quality are listed in Tables V and VI, including the CT attenuation, image noise and contrast noise rate. Mean attenuations for all objects were higher at a voltage of 80 kVp compared with 120 kVp, when slice thickness was 0.75 mm (70.59 HU vs. 66.81 HU, respectively; P<0.001) or 1.5 mm (69.67 HU vs. 65.59 HU, respectively; P<0.001). In addition, the mean image noise was increased at a tube voltage of 80 kVp compared with that at 120 kVp, at both slice thicknesses (P<0.001). Regarding the comparison between measurements at a tube current of 200 and 100 mAs, the mean attenuation was almost the same between the two currents (P>0.05), whereas the image noise was slightly increased at lower tube current (P<0.001 and P<0.007 at slice thicknesses of 0.75 and 1.5 mm, respectively). At the same slice thickness, tube voltages had no effect on contrast noise rate (P>0.05), whereas tube currents had a significant influence on contrast noise rate (P<0.05) in this study. Therefore, higher tube currents produce higher contrast noise rates.

Discussion

The segmentation software used in the present study demonstrated good performance on measuring the object volume *in vitro*. The mean PEs of objects on volumetric measurements were <1% when the surrounding environment was oil. Brown *et al* (15) previously reported a mean difference of 1.3% between the *in vivo* CT-based volume of porcine kidneys and the *in vitro* volume of water displacement. In addition, Chul Kim *et al* (23) performed 3-mm thickness CT scanning in 2 porcine kidneys, and reported a 2.9% difference between the volume measurement by the *in vivo* CT scan and the *in vitro* volume measured by water displacement. Cai *et al* (24) performed *ex vivo* CT scanning of 8 pig kidneys, and the mean PE of volume measurement using the dynamic-threshold level set method was 3.38±2.51%, when compared with reference standards determined by fluid displacement. Typically, higher PEs may be observed *in vivo* in comparison with those *in vitro* due to inhomogeneity of the surrounding tissues. Similar to the PEs obtained by the aforementioned previous studies, the mean PEs in the present study were acceptable and negligible. Furthermore, the observers needed ~2 min to segment a potato or a porcine kidney using the segmentation software constructed in the present study. A deformable contour method (Snake) was applied to develop the segmentation method used in the present study, due to the accuracy and the computational efficiency of the Snake method (21). The 2D contours in axial images can be extracted within several seconds on a normal personal computer platform, and the segmentation results can be easily corrected by adjusting the 2D contours manually.

The environments surrounding the object can affect the segmentation process (18). Thus, in the present study, objects were directly exposed to the air or immersed in oil for *in vitro* CT scanning. The results showed that the object volume measured in oil was closer to the actual volume, as observed by water displacement, and the volume measured in the air consistently underestimated the volumes. The boundary of the object segmented by the software was determined by the different attenuation between object and surrounding environment. Since the air attenuation is significantly lower compared with that of oil, the CT value at the object boundary decreases much faster when the objects is exposed in the air (20). Therefore, the cut-off point on the object boundary determined in the air by the software of the current study tended to be reduced when the same threshold value was used.

A tube voltage/current of 120 kVp/200 mAs are the standard parameters for abdominal CT imaging used at the First Affiliated Hospital of Nanjing Medical University (Nanjing, China). For CT scans in thin adult patients and children, a low energy kVp protocol (80 kVp/200 mAs) and a low mAs protocol (120 kVp/100 mAs) is typically selected, according to the 'as-low-as reasonably achievable' rule (25). Dong *et al* (26) demonstrated that a low-energy technique for pediatric patients was possible with the similar image quality. In addition, Nakaura *et al* (27) showed that a low kilovoltage with high tube current improved the image quality and reduced the radiation dose. However, the effect of low kilovoltage and low tube current on segmentation of solid organ has not been widely evaluated. A decrease in the X-ray tube voltage resulted in reduced amount of penetrating radiation, and thus an increase in object attenuation and image noise (28-30). Numerous studies have focused on the lower kilovoltage application in iodinated contrast agent CT scanning, since the lower kilovoltage imaging resulted in evidently higher attenuation of iodine (28,29). However, in the present study, the object density increased by only ~5 HU at 80 kVp, compared with CT images at 120 kVp, while the image noise was slightly increased simultaneously. These two factors had adverse effects on the object segmentation performed using the segmentation software of the current study *in vitro*. Therefore, the accuracy of volume measurements at 80 and 120 kVp was similar in the present study.

A decrease in the X-ray tube current leads to an increase in image noise without affecting the CT attenuation (18,28,29), which was also demonstrated by the present study results. Image noise is certainly a factor that strongly affects the segmentation process, particularly especially at the determination of the initial contour of objects for the deformable model application (18). The current results showed that the volume determined by the software were more accurate at the higher tube current. In addition, the 3D region representing the shape of the object obtained using the Snake model was closer to the actual value at thinner slice thickness due to isometric characteristics (17). However, the best performance of the software in the current study was obtained at a higher slice thickness. The minimal mean PE was observed at a slice thickness of 1.5 mm, irrespective of the tube voltages or currents used. This result may be explained by an increased image noise at the 0.75 mm thickness.

The present study also presents several limitations. The absence of surrounding tissues, which have varying densities, resulted in easier distinction of the object from the background. Porcine kidney examination *in vivo* could have evaluated the semiautomatic segmentation software under a conditions that are more similar to humans. In addition, the density of potatoes and porcine kidneys *ex vivo* was relatively fixed (~70 HU for potatoes and ~50 HU for porcine kidneys). Therefore, the threshold for volume segment did not require modulation in order to avoid omitting kidney volume; by contrast, the attenuation of kidney is variable in patients, particularly when enhanced CT is performed in different phases.

In conclusion, the segmentation software described in the present study is a promising quantitative method for renal volume measurements on CT scans. However, different environments, tube voltages/currents and slice thicknesses can affect the performance of this segmentation software. The mean PE of object volumetric measurements was minimal when performed in oil with a tube voltage of 120 kVp, a tube current of 200 mAs and a slice thickness of 1.5 mm. Extension of this segmentation software to human beings and validation of the current findings should be investigated in future studies.

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