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Automated Quantification of Aortoaortic and Aortoiliac Angulation for Computed Tomographic Angiography of Abdominal Aortic Aneurysms Prior to Endovascular Repair: Preliminary Study

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Introduction

Minimally invasive endovascular aortic aneurysm repair (EVAR) using stent-grafts has become an accepted method of treatment for infrarenal abdominal aortic aneurysms (AAA). However, about 55% of patients are deemed unsuitable for endovascular surgery, largely due to unfavorable aortic morphology (1). Computed tomographic angiography (CTA) is currently the modality of choice for the evaluation of aortic morphology (2). Recently, the degree of angulation of the infrarenal aorta and iliac arteries has emerged as an important factor in assessing eligibility for EVAR (3), predicting operative difficulty (4) and as a risk factor for postoperative complications (3,5). The degree of angulation may be the most important predictive factor for postoperative graft migration and for the occurrence of type I endoleaks that may require secondary procedures to secure stent-graft fixation or necessitate conversion to open surgery. Stent grafts placed in angulated aortas are also at greater risk for progressive kinking that can be independent of length or diameter change in the aneurysm (3).

Several publications (6-9), including a recent paper by Diehm et. al. (10), have reported high variability in manual measurement of angulation and called for a more reliable and accurate method. Our aims were to quantify the variability of manual infrarenal aortoiliac angulation measurements and to reduce inter- and intra-observer variability by developing an automated three-dimensional analysis algorithm.

Materials and Methods

Algorithm

At our institution, the angulation of the aorta and iliac arteries at four anatomic locations are measured manually by our technologists for every patient with an infrarenal aortic aneurysm. Measurements of angulation are provided at (1) the level of the superior boundary of the aneurysm sac (termed the “proximal neck angle”), (2) the middle of the aneurysm sac (termed

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the “aneurysm angle”), and at (3) each common iliac artery origin (termed the “common iliac artery angle”) for a total of four angles. Our algorithm automatically measures the angulation of the infrarenal aorta and common iliac arteries at these locations.

The algorithm first prompts the user to select the following points as inputs to the algorithm: one point within the aortic lumen at the level of the most inferior renal artery origin (IR, Figure 1) (identifying the superior boundary of the proximal neck of the aneurysm), and one point within the right and left distal common iliac artery lumina (RI and LI, Figure 1). These three points need not be exactly centered within the lumen. A previously developed algorithm (11) is then applied to derive a three-dimensional branched median centerline through the infrarenal aorta and common iliac arteries. Our algorithm then automatically determines five points along the centerline of the common iliac arteries, as detailed in Figure 1, and fits three line segments to the median centerline within the aorta (Figure 2). The locations of the two points, “proximal neck” (PN) and “mid aneurysm” (MA), that divide the aortic centerline into three segments, are adjusted by the algorithm as it attempts to minimize the fit error, calculated as the percentage difference between the length of the centerline from the level of the origin of the most inferior renal artery to the aortic bifurcation, compared with the combined length of the line segments. Finally, the four angles subtended by these line segments are calculated as demonstrated in Figure 3.

Validation

This study was conducted under an approved protocol of our Institutional Review Board. To assess the accuracy of our algorithm relative to an absolute standard *in vitro*, we constructed a set of virtual phantoms modeled on a randomly selected patient with an abdominal aortic aneurysm, to simulate the irregular, calcified surface of the aorta. Using mathematical deformation, phantoms were created with known angulations of 0 - 90 degrees at 10 degree intervals, at the proximal neck, mid aneurysm and distal to the aneurysm for a total of 30 phantoms. CT scans of these phantoms were then simulated using CT scanner simulation software. One of the authors provided inputs for the automated angulation-measurement algorithm, which was applied to the phantoms. (Figure 4 shows an example of a phantom with 30° degree compound proximal neck angle.)

Next, we performed a preliminary *in vivo* evaluation of our method using CTAs from 11 patients (3 female, 8 male, mean age 79 years, range 66-93 years) with AAAs and without a history of surgical or endovascular treatment. CT scans were acquired with 1.25-2.5 mm section thickness, 0.8-1.25 mm section spacing, 120 kV, and 320-440 mA using multidetector-row CT scanners. One hundred to one hundred fifty ml of iodinated contrast medium (Omnipaque 350 mg I/ml) was injected at a rate of 4-5 ml/s during the scan. We assessed inter- and intra-observer variability of our automatic method compared to standard manual angle measurement. Five radiologists (three diagnostic radiologists and two interventional radiologists) with extensive experience reading pre-EVAR CTAs, and three dedicated 3-D technologists with a minimum of 3 years experience in processing pre-EVAR CTAs measured the proximal neck angle, aneurysm angle and two common iliac angles on all patients. Each observer repeated these measurements in 3 trials, with a mean of 3 days between trials on the same patient. The observers followed the same written and pictorial guidelines used in our 3D Laboratory for placement of points for angle measurements on clinical images. To assess automated measurement variability, the points specified by each of the observers at the most inferior renal artery point and 20 mm distal to each common iliac artery origin were used as inputs to the automated algorithm, allowing us to obtain one set of automated measurements for each set of manual measurements.

Statistical Analysis

The R^2 value and mean absolute difference between the automated measurements and the known angle measurements in the virtual phantoms were calculated. Schuirmann's Two One-Sided Equivalence Test with a paired design was used to test the null hypothesis that the automated measurements were different from the known measurements in phantoms. The maximum acceptable difference that was clinically unimportant was set at 5%. A p-value of 0.05 was considered to be significant. Interobserver and intraobserver variabilities for both automated and manual measurements were calculated using a standard deviation. Additionally, Levene's tests (12) were used to test the null hypothesis that the interobserver and intraobserver variability of the manual and automated measurements are equal. A p-value of 0.01 was considered to be significant.

Schuirmann's Two One-Sided Equivalence Test (13) with a paired design was used to test the null hypothesis that the angles measured by the automated algorithm were different than those measured using the manual method. The maximum acceptable difference that was clinically unimportant was set at 5%. A p-value of 0.05 was considered to be significant.

A paired t-test was used to test the null hypothesis that the time taken by the automated algorithm was the same as the time taken for manual measurement of angles. A p-value of 0.05 was considered to be significant.

Results

Figure 5 shows the correlations between automatically measured and known angles in the "virtual phantoms", with R^2 values of 0.99 in all cases and slopes very close to one. Over all phantoms and angles, the mean error between the actual and measured angles was 0.7 ± 0.5 degrees (s.d.). The p value for Schuirmann's Two One-Sided Equivalence Test was less than 0.05, allowing us to reject the null hypothesis and conclude that there was no statistically significant difference between angles measured by the automated algorithm and the known angle measurements.

The range of measured angles was 7.1 to 46.6° for the proximal neck, 23.2 to 57.4° for the aneurysm angle and 13.6 to 72.9° for the common iliac angles bilaterally. Table 1 shows inter- and intra-observer variability for manual and automated measurement of proximal neck, aneurysm and common iliac angles. Intraobserver ($p < 0.01$) and interobserver ($p < 0.01$) variability was significantly lower than for manual measurement. The p value for Schuirmann's Two One-Sided Equivalence Test was less than 0.05, allowing us to reject the null hypothesis and conclude that there was no statistically significant difference between manually and automatically measured angles.

Manual measurement of angles required 4.2 ± 1.3 min per patient. Automated measurement required 0.7 ± 0.3 min for manual interaction and 1.9 ± 1.1 min of computer processing time per patient for a total of 2.8 ± 1.4 min per patient ($p = 0.57$). Manual interaction time was significantly lower for the automated method ($p < 0.05$).

Discussion

Manual measurement of aortoiliac angulation may be excessively variable (6,7,10). Clinically significant variability has also been reported in manual measurements of angulation using CT in other structures such as in extremities (8,14) and pulmonary veins (9). The interobserver variability of manual measurements in this paper is slightly less than that reported by Diehm et. al. ($\sim 6^\circ$ vs $\sim 13^\circ$, 20% vs 32%) (10). This could be due to differences in patient population, but also could be because our readers were given explicit written and pictorial instructions on

how measurements were to be performed. Our automated approach is faster, yields measurements that are less variable and as accurate as manual methods, and the absolute amount of variability is very small. This may be partly because the median centerline derived from the user-defined points has been shown to be relatively insensitive to positional variability in the user-defined points (15). If there is complete occlusion of the aorta or common iliac arteries, the interface allows the user to bridge the occluded segment manually, allowing the calculation of a branched median centerline. Our algorithm does not require hardware acceleration and is operating system independent, and so can potentially be implemented as an add-on to commercial CT scanner consoles or postprocessing workstations.

One limitation of the study is that we used a relatively small set of patients, but this set was enough to achieve statistical significance. Our algorithm still requires some user input, but the time required for this is small. We have not tested the algorithm in aortic aneurysms that have concomitant aortic dissections, where there is the potential of interference with the algorithm's ability to delineate the aortic lumen. However, this can also be addressed with manual input of centerline points. A common problem in evaluating algorithms of this type is that there is seldom a reliable reference standard, hindering the determination of accuracy. Thus, many studies report only variability (i.e., precision) and compare it to that derived using some other (e.g., manual) established method, as we have also done. However, we also evaluated accuracy using a novel approach – mathematical manipulation of clinically derived aneurysm shape in which true angulation could be known and varied. While perhaps not completely representative of what might be obtained if true angulations were known over a wide range of CTA data from patients, our approach, which generated realistic aortic morphology over just such a range of angulations, showed that the algorithm had a very good accuracy (≤ 1 degree) in this setting. This, when combined with the low variability between the automated and manual methods reported above, provides reasonable assurance of clinical applicability even in the absence of a solid reference standard for accuracy.

Our automated algorithm could enhance the clinical utility and reliability of CTA for preoperative assessment for EVAR. Our preliminary results support continued development and evaluation in a larger set of patients.

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Figure 1.

Critical points for automatic determination of aortic angles. (a) Antero-posterior and (b) right anterior oblique MIP projections through the aorta and common iliac arteries showing orange colored user-selected points and additional points calculated by the algorithm (red) following application of centerline algorithm. Using these points, a branched centerline is calculated from IR to RI and LI. The set of five points along the centerline that are subsequently calculated by the algorithm, shown in red, are then saved to allow calculation of the right and left common iliac artery angles (Fig. 3). (R: Right; L: Left; AM: Anteromedial; PL: Posterolateral; IR: at the level of the most distal renal artery origin; RI (LI): in the distal right (left) common iliac artery; AB: aortic bifurcation; RIO (LIO): origin of the right (left) common iliac artery; RIE (LIE): 20 mm distal to the origin of the right (left) common iliac artery)

Figure 2.

Algorithmic determination of line segments defining proximal neck and mid-aneurysm angles. (a) Antero-posterior and (b) oblique lateral MIP projections through the aorta and common iliac arteries. The orange circles are points that are selected by the user (Fig. 1). Three line segments (black) are fit to the centerline between IR and AB. The points of intersection between the lines are designated as the proximal neck point (PN) and the mid-aneurysm point (MA). (R: Right; L: Left; AM: Anteromedial; PL: Posterolateral)

Figure 3.

Antero-posterior oblique MIP projections through the aorta and common iliac arteries detailing calculation of four angles using the line segments 1-7. The two line segments that are used to calculate each angle are shown in (a) proximal neck angle, (b) mid-aneurysm angle, (c) right and left common iliac angles. (PL: Posterolateral; AM: Anteromedial)

Figure 4.

Example images illustrating creation of an aortic phantom with a known amount of angulation at the proximal neck using a CTA of the abdomen and pelvis (0.8 mm collimation, 120 kV/440 mA). (a) Coronal and (b) sagittal volume renderings of straightened aortic luminal surface with the straightened median path shown as a black line. (c) Coronal and (d) sagittal volume renderings of aortic luminal surface with a 30 degree compound proximal neck angle created by computationally deforming the aorta anterior and to the left above the proximal neck of the aneurysm. The aortic luminal surface appears irregular due to the impressions of the mural thrombus and calcifications. (P: Posterior; A: Anterior; R: Right; L: Left)

Figure 5.
 Plots of automatically measured versus actual angles in virtual phantoms.

Table 1
Interobserver and intraobserver variability per segment in manual and automatic angulation measurements

Segment Variability	Interobserver (Manual)	Interobserver (Auto)	Intraobserver (Manual)	Intraobserver (Auto)
Prox. Neck	4.6 ± 1.8° (22 ± 11 %)	0.3 ± 0.3° (1.5 ± 2.0 %)	6.9 ± 4.1° (33 ± 21 %)	0.5 ± 0.6° (2.4 ± 3.6 %)
Aneurysm	5.5 ± 2.4° (18 ± 7.5 %)	0.4 ± 0.5° (1.2 ± 1.1 %)	8.2 ± 4.7° (27 ± 17 %)	0.7 ± 1.0° (2.2 ± 3.7 %)
Common Iliac a.	6.2 ± 2.8° (21 ± 8.7 %)	0.4 ± 0.4° (1.4 ± 2.1 %)	8.9 ± 5.4° (31 ± 20 %)	0.6 ± 0.7° (2.1 ± 3.6 %)
All Segments	5.6 ± 2.5° (20 ± 9.1 %)	0.4 ± 0.4° (1.4 ± 1.9 %)	8.2 ± 5.0° (31 ± 20 %)	0.6 ± 0.8° (2.2 ± 3.6 %)