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Daniel Modin, Johan Renner, Roland Gårdhagen, Tino Ebbers, Toste Länne and Matts Karlsson, Evaluation of Aortic Geometries created by MRI Data in Man, 2011, Clinical Physiology and Functional Imaging, (31), 6, 485-491. which has been published in final form at: <u>http://dx.doi.org/10.1111/j.1475-097X.2011.01035.x</u> Copyright: Wiley-Blackwell <u>http://eu.wiley.com/WileyCDA/Brand/id-35.html</u>

Postprint available at: Linköping University Electronic Press http://urn.kb.se/resolve?urn=urn:nbn:se:liu:diva-65905

Evaluation of Aortic Geometries Created by MRI Data in Healthy Volunteers

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December 22, 2011

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Abstract

The development of atherosclerotic plaques has been associated with the patterns of wall shear stress (WSS). However, much is still uncertain with the methods used to calculate WSS. Correct vessel geometries are mandatory to get reliable estimations and the purpose of this study was to evaluate an in vivo method for creating aortic 3D geometry in man based on data from magnetic resonance imaging (MRI) with ultrasound as reference.

Methods: The aortas of ten healthy males, 23.4 ± 1.6 years of age, were examined with a 1.5T MRI system using a 3D gadolinium-enhanced gradient-echo sequence. 3D geometries were created using manual segmentation of images. Lumen diameters (LD) were measured in the abdominal aorta (AA) and the thoracic aorta (TA) with non-invasive B-mode ultrasound as a reference.

Results: The anteroposterior diameter of the AA was 13.6 ± 1.1 mm for the MRI and 13.8 ± 1.3 mm for the ultrasound (NS). Intraobserver variability (CV) for MRI and ultrasound was <0.92% and <0.40% respectively . Interobserver variability MRI and ultrasound was 0.96% and 0.56% respectively. The diameter of the TA was 19.2 \pm 1.4 mm for the MRI, and the intraobserver variability (CV) were <0.78% and interobserver variability (CV) were 0.92%.

Conclusion: Specific arterial geometries can be constructed with a high degree of accuracy using MRI. This indicates that the MRI geometries may be used to create realistic and correct geometries in the calculation of WSS in the aorta of man.

Keywords: Human Aorta, Lumen Diameter, Magnetic Resonance Imaging, Manual Segmentation, Ultrasound

Introduction

Atherosclerotic lesions in arteries are responsible for 50% of all morbidity and mortality in the western world Ross (1993). The atherosclerotic lesions are often distributed at curvatures and branches which has been interpreted as evidence that hemodynamics play an important role in the atherosclerotic process. One of the most important hemodynamic parameters is wall shear stress (WSS), and low as well as oscillatory WSS has been associated with the atherosclerotic process Pedersen et al. (1999, 1997); Malek et al. (1999); Cheng et al. (2006); Taylor et al. (1998); Moore et al. (1994); Karmakar (2001); Pyke and Tschakovsky (2005); Chappell et al. (1998); Bonert et al. (2003); Moore et al. (1999). Detailed information about the velocity profile is required to determine WSS in vivo however. In order to simulate the flow profile it is mandatory to have a correct geometry of the specific artery studied since local flow behavior is strongly dependent on vessel geometry Renner et al. (2009); Leuprecht et al. (2002). This can be obtained using computational fluid dynamics (CFD) in combination with data from MRI scans Wood et al. (2001); Leuprecht et al. (2002); Long et al. (2000); Svensson et al. (2006); Renner et al. (2009). The purpose of this study was to validate aortic geometries obtained with MRI with non-invasive ultrasound as a reference.

Material and Methods

Ten healthy males (23.4 ± 1.6 years) participated in the study. All where non smokers, without history of cardiovascular disease. None of them was on any medication. Body weight 72.7 \pm 4.7 kg. Body mass index $21.8 \pm 1.4 \ kg/m^2$. All subjects abstained from eating or drinking coffee 6 hours before the examinations and were investigated with MRI and ultrasound at the same occasion. They were transported with wheelchair between the MRI and ultrasound examinations in order to have similar hemodynamic condition during each examination. All volunteers gave informed consent, and the study was approved by the regional Ethics Committee for Human Research at Linköping University, Sweden.

Before the examinations all subjects were resting in supine position at least 15 minutes. Blood pressure was obtained in both arms with a sphygmomanometer. First, the anterior-posterior diameter (LD) of the abdominal aorta (AA) at the midpoint between the renal arteries and iliac bifurcation was visualized longitudinally in real-time with aid of ultrasound (Philips ATL HDI 5000, Oceanside, USA) with a 5-2 MHz B mode real-time curved array transducer. The picture was frozen at end diastole according to prevailing standards and stored on a magneto optical disk (3.5" (90 mm) Rewritable 230 MB optical disk, Maxell, New Jersey, USA) and analyzed in a PC (Intel Pentium 4, Dell, Round Rock, USA). The analyzing system (AMS, Chalmers University of Technology, Gothenburg, Sweden) measured mean LD of a one cm long AA segment, from leading edge of near wall to leading edge of far wall using 100 measuring points. All measurements were carried out by two experienced technicians. The data of the ten aortas from the analyzing system was then duplicated and randomized in such a manner that LD was measured from each data set two times by each technician. In one subject the LD could not be measured accurately due to technical problems and was excluded.

MRI was used to collect data from the ten male volunteers using a 1.5 T MRI scanner and a 5 channel receiver coil (Philips Achieva, Philips Medical Systems, Best, the Netherlands). Geometrical information of the complete aorta was obtained within a breath hold using a 3D gadolinium-enhanced gradient-echo sequence (TE 1.6 ms, TR 5.3 ms, and flip angle 40, field of view 400x360x80 mm, acquisition matrix 400x230x80, SENSE factor 1.5). A 30 ml (0.5 mmol/ml) contrast bolus (Omniscan, Amersham Health, Oslo, Norway) was injected at 2.0 ml/s. The group of volunteers were so homogenous that the weight adapted calculations resulted in the same injection interval (30 ml). Randomly segmented central k-space ordering (CENTRA) was used. The three-dimensional volume data was reconstructed to a resolution of 0.78x0.78x1.00 mm.



Figure 1: MRI slice of the aortic arch with sketched examples of manually placed points marking the interface between the luminal wall surface and the blood.

To create the geometrical model of the MRI images and to locate the aortic wall, a computer aided manual segmentation was used. At least 1000 points were placed manually at the aortic luminal surface on every slice containing parts of the aorta with a density of ten points per cm, with aid of an in house-developed software in Matlab (Figure 1). In this manner one of the two investigators defined the geometry of the entire aorta, while the other studied a two cm long segment in the AA at the midpoint in between the renal arteries and the aortic bifurcation, and a three cm long segment in the thoracic aorta (TA) starting at the end of the aortic arch and distally (Figure 2). The total number of points (about 20000) was then used to describe the 3D geometry of the entire aortic luminal surface. This cloud of points was then converted to a tessellation surface in a CAD (Computer Aided Design) software Catia V5 (Dassault Systemes, Vélizy-Villacoublay Cedex - France) which is a surface based on triangles. The MRI data from the ten aortas were then duplicated and randomized in such a manner that 20 aortic geometries were manually created by each investigator, meaning that each data set was measured two times by each technician.



Figure 2: 20000 points describing the 3D geometry of the aorta created by investigator 1. Arrows indicate overlay plots in the AA and the TA created by investigator 2.



Figure 3: A two-cm longitudinal segment at the midpoint of the abdominal aorta, between the renal arteries and the aortic bifurcation shown in the ICEM program. The anteroposterior diameter was measured at 16 different points and a mean LD was calculated

To calculate LD, the geometry was then processed and measured manually with the measuring tool in ICEM 10.0 (ANSYS, Inc., Canonsburg, Pennsylvania, USA) (Figure3). The LD in the (AA) was measured at the same segment studied by ultrasound. In the TA the LD was measured in a segment from the end of the aortic arch and three cm distally. A mean anteroposterior LD of the AA and TA segments was calculated and used in the study.

Statistic Analysis

Mean \pm SD was used describing the data. Variability calculations were performed with the method described by Bland and Altman (Bland and Altman, 1986). The coefficient of variation (CV) was calculated with the formula $CV(\%) = s \cdot \frac{100}{X}$. The formula $s = \frac{SD}{\sqrt{2}}$ was used to calculate the standard deviation of the intraobserver error (s) (Wendelhag et al., 1991). To evaluate the whole subject group the average CV were calculated and used in the analysis. Comparisons were made with the Student t test and P<0.05 was considered as significant, the abbreviation (NS) is used for not significant differences.

Results

The anteroposterior diameter of the AA was 13.6 ± 1.1 mm for the MRI and 13.8 ± 1.3 mm for the ultrasound. In TA were only MRI could be used were the anteroposterior diameter 19.2 ± 1.4 mm. No differences between MRI and ultrasound in AA an TA lumen diameters were found (NS). The intra-individual variability of LD mea-



Figure 4: Intraobserver variability of lumen diameter (LD) measurements with MRI in the abdominal aorta (AA). Difference between measurement 1 and 2 is denoted on vertical axis and mean of measurement 1 and 2 on horizontal axis. Results based on both investigator 1 and 2.

surements in the AA with MRI for investigator 1 and 2 is shown in Figure 4, were CV was 0.86% and 0.91%. Figure 5 shows the MRI inter-individual variability based on the first measurement (the most clinically relevant data) of LD in the AA with MRI, the CV was 0.96%. Figure 6 shows the intra-individual variability of the LD in the AA obtained by ultrasound (US) for technican 1 and 2. CV were 0.39% and 0.27%.



Figure 5: Interobserver variability of lumen diameter (LD) measurements with MRI between investigator 1 and 2 in the abdominal aorta (AA). Difference between investigator 1 and 2 is denoted on vertical axis and mean LD of investigator 1 and 2 on horizontal axis. Results are based on both measurements on the AA.



Figure 6: Intraobserver variability of lumen diameter (LD) measurements with US in the abdominal aorta (AA). Difference between measurement 1 and 2 is denoted on vertical axis and mean of measurement 1 and 2 on horizontal axis. Results are based on both technican 1 and 2 data.



Figure 7: Interobserver variability of lumen diameter (LD) measurements with US between investigator 1 and 2 in the abdominal aorta (AA). Difference between investigator 1 and 2 is denoted on vertical axis and mean LD of investigator 1 and 2 on horizontal axis. Results are based on both measurements on the AA.

Figure 7 shows the ultrasound inter-individual varaibility of the LD in the AA based on the first measusrement of each technican. CV were 0.56%.



Figure 8: Image modality (MRI vs US) variability of lumen diameter (LD) measurements in the abdominal aorta (AA). Differences between MRI and US is denoted on vertical axis and mean LD of MRI and ultrasound on horizontal axis.

In Figure 8 is MRI and US results compared and viewed as a bland-altman plot for both measurements made by investigator 1 and technican 1. Figure 9 shows the intraindividual variability of LD in TA using MRI. LD of the TA was 19.2 ± 1.4 mm, CV 0.72% (investigator 1), and 19.1 ± 1.4 mm (investigator 2) CV 0.77%. There was no difference in lumen diameter (NS). Figure 10 shows the MRI inter-individual variability based on the first measurement (the most clinically relevant data) of LD in the AA with MRI, the CV was 0.92%.



Figure 9: Intraobserver variability of lumen diameter (LD) measurements with MRI in the thoracic aorta (TA). Difference between measurement 1 and 2 is denoted on vertical axis and mean of measurement 1 and 2 on horizontal axis. Results are based on both investigator 1 and 2 data.

Table 1 summarizes the LD±SD, Bias, SD, Bias±SD, LoA (Limits of Agreement and CV of the AA and TA measurements with MRI and ultrasound both for intra- and interobserver variability. The blood pressure for the ten subjects was 126 ± 9.8 mm-Hg and 67 ± 6 mm-Hg, systolic and diastolic pressure respectively.



Figure 10: Interobserver variability of lumen diameter (LD) measurements with MRI in the thoracic aorta (TA). Difference between investigator 1 and 2 is denoted on vertical axis and mean of investigator 1 and 2 on horizontal axis. Results are based on both measurements on the AA.

Table 1: AA, abdominal aorta; TA, Thoracic aorta; CV, coefficient of variation (average); MRI, magnetic resonance imaging; US, ultrasound; LD, lumen diameter; Bias, mean of differences; SD, standard deviation of the differences; LoA, Limits of Agreement (Bias ± 1.96 ·SD); Inv., Investigator; Techn., Technican.

Variability	LD±SD (mm)	Bias (mm)	SD (mm)	LoA (mm)	CV (%)
AA					
MRI					
Intraobserver (Inv. 1)	13.5±1.2	0.01	0.29	[-0.56 0.59]	0.86
Intraobserver (Inv. 2)	13.7±1.1	0.21	0.26	[-0.31 0.72]	0.91
Interobserver	13.6 ± 1.1	-0.20	0.38	[-0.94 0.54]	0.96
AA					
US					
Intraobserver (Techn. 1)	13.8±1.3	0.09	0.12	[-0.14 0.31]	0.39
Intraobserver (Techn. 2)	13.8±1.3	-0.03	0.10	[-0.22 0.15]	0.27
Interobserver	13.8±1.3	0.05	0.20	[-0.35 0.45]	0.56
TA					
MRI					
Intraobserver (Inv. 1)	19.2±1.4	-0.04	0.37	[-0.76 0.68]	0.72
Intraobserver (Inv. 2)	19.1±1.4	0.10	0.34	[-0.57 0.77]	0.77
Interobserver	19.2±1.4	0.04	0.41	[-0.76 0.84]	0.92

Discussion

A new method to create subject specific aortic 3D geometries out of magnetic resonance imaging (MRI) data from in vivo measurements in man was evaluated. The aorta was chosen due to its preponderance for atherosclerotic lesions as well as aneurysm development. The MRI geometries correlated well with ultrasound measurements. Furthermore, the intraindividual and interindividual variability when making MRI geometries was low.

The atherosclerotic disease is described as a chronic inflammatory process (Ross, 1999). Initially, a thickening of the intima with infiltration of blood monocytes and foam cell is found. In presence of additional risk factors like hyperlipidaemia, diabetes, hypertension and smoking the lesions may progress into occlusive disease. Low and oscillatory wall shear stress (WSS), found predominately in curvatures and branches of arteries are associated with increased permeability and up-regulation of cellular adhesion molecules on the endothelial surface, inducing intimal thickening and inflammatory response, all leading to formation of atherosclerotic plaques (Karmakar, 2001; Chappell et al., 1998; Pyke and Tschakovsky, 2005; Malek et al., 1999; Pedersen et al., 1997; Cheng et al., 2006; Moore et al., 1999). However, much of the data that indicate a correlation between low and oscillatory WSS and intimal thickening and plaque formation estimates the WSS in a simplified way (Pyke and Tschakovsky, 2005; Karmakar, 2001). The Hagen-Poiseulle formula ($WSS = \frac{4\mu\overline{u}}{d}$ where μ =viscosity, \overline{u} =mean velocity and d=diameter) does not give the value of the local WSS patterns close to the arterial wall, and the velocity profile in relation to differences in vessel configuration is unknown. The blood flows inside the arteries in a complex pattern which produce different areas with high or low WSS, and realistic WSS patterns with simulations of the arterial geometry needs to be estimated in order to obtain realistic WSS information (Leuprecht et al., 2002; Long et al., 2000). Creating reliable geometries requires a reliable method to visualize the arteries. In clinical work angiography is often used, but 3D reconstructions is impossible because the images are in 2D. Ultrasound has similar difficulties in creating reliable geometrical 3D data. On the other hand, ultrasound gives useful information from a limited arterial segment and has been shown to be an accurate method for measuring the lumen diameter (LD) of the abdominal aorta (AA) (Pedersen et al., 1993; Singh et al., 1998; Länne et al., 1997; Åstrand et al., 2003). Ultrasound results is operator dependent, but with a experienced operator can reliable results be gained at specific sites and limited strerial segments. This is shown by the low coefficient of variation for the LD (2-4%) reported by Åstrand et al. (2003). Computer tomography (CT) gives high resolution 3D images that theoretically could be used to create arterial geometries. One disadvantage with CT is the radiation, another is that the images are collected during a limited period of time without measurement of the blood velocity. MRI, on the other hand, may gather both velocity information and geometrical data.

In this study we used a non-ECG gated 3D contrast enhanced MRI sequence. Today, this is clinically the most common technique to obtain 3D aortic geometries. Cardiac motion can result in blurring of the aortic root and ascending aorta. Gated 3D contrast enhanced MRI is not straightforward, but may reduce this blurring. Two-dimensional

techniques, as balanced turbo field echo cine bright blood and triggered T2 black blood have shown to allow for accurate determination of aortic diameter Groth et al. (2011); Potthast et al. (2010), but do not provide a 3D geometry. Three-Dimensional (3D) navigated Steady-State Free-Precession (SSFP) allows for 3D assessment of the geometry without contrast injection. This technique has shown to produce reproducible aortic measurements with the same diameters as obtained from non-ECG gated 3D contrast enhanced MRI Potthast et al. (2010). Clinical usage of this technique is still limited, however.

MRI technology is rapidly evolving with higher spatial and temporal resolutions providing detailed geometrical data to be used in a clinical setting. However, the information cannot directly be used in computational fluid dynamics (CFD) simulations to calculate the WSS since the data are designed to be shown on a flat screen without geometrical data points to base calculations on. To be able to use the basic geometrical data, the interface between aortic wall and lumen was identified manually, and a 3D geometry model of the aorta created (Figure 1 and 2). The 3D model was then given a surface and the LD measured (Figure 3). The LD of the AA in the MRI geometries agreed well with that of the ultrasound Table 1. The both image modalities do not have any systematic error due to the low values of the Bias in Table 1. Furthermore, the reproducibility of the MRI geometries was good (Figure 4, 5, 9, 10, and Table 1). Due to the longer distance to the MR coil (placed in the heart region) in the MRI measurements in the AA region the image material were of poorer quality at the AA site. This explains the slightly better variability results (Table 1) for the TA section. Based on the measured velocity profile in the proximal aorta together with the geometry, CFD simulations provide information regarding the flow velocity profiles in the entire aorta and may be used to estimate WSS with high spatial and temporal resolution This in turn gives the possibility to better understand the relationship between hemodynamic flow parameters and arterial wall remodeling as well as the pathophysiology of atherosclerosis. The manual work regarding delineation of the luminal-wall interface is time consuming however, and an auto-segmentation technique is now being developed to be used as a standard method to construct MRI geometries of arteries.

In conclusion, A new method to create subject specific aortic geometries out of magnetic resonance imaging (MRI) data from in vivo measurements in man was evaluated. The MRI geometries correlated well with ultrasound measurements. Furthermore, the intraindividual and interindividual variability when making MRI geometries was low. This indicate that the MRI geometries may be used to create realistic and correct geometries in the calculation of WSS in the aorta of man.

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