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<td>Effects of Sac Geometry and Stiffness on Pulse Wave Propagations in FSI Models of the Human Abdominal Aortic Aneurysm</td>
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<td>Abstract:</td>
<td>This study aims at quantifying the effects of geometry and stiffness of aneurysms on the pulse wave velocity (PWV) and propagation in fluid-solid interaction (FSI) simulations of arterial flow. Spatio-temporal maps of both wall displacement and fluid velocity were generated in order to obtain the pulse wave propagations through fluid and solid media, and to examine the interactions between the two waves. The results indicated that the presence of AAA sac and variations in the sac modulus affect the propagations of the pulse waves both qualitatively (e.g. patterns of change of forward and reflected waves) and quantitatively (e.g. PWV decreases inside the sac and increases post-sac as sac stiffness increases). The sac properties can thus be estimated through both the wall displacement and fluid velocity spatio-temporal maps. The sac region is particularly identified on the spatio-temporal maps with a region of disruption in the wave propagation with multiple short-travelling forward/reflected waves, which were caused by the change in boundary conditions within the saccular region. The change in sac stiffness, however, is more pronounced on the wall displacement -rather than fluid velocity- spatio-temporal maps. The findings of the study demonstrate initial findings in numerical simulations of FSI dynamics during arterial pulsations that can be used as reference for experimental and in vivo studies.</td>
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1. Introduction

Changes in wall mechanical properties have been shown to disrupt the normal hemodynamics of the arteries and contribute to various cardiovascular diseases (CVDs)\(^1\). In particular, changes in aortic stiffness have been reported in prior studies as an independent indicator of all-cause and CVD-related mortalities, such as Abdominal Aortic Aneurysm (AAA)\(^2-5\). Assessing arterial stiffness has been collectively and increasingly recommended to be an essential part of clinical diagnosis, therapy and follow up procedures\(^6,7\). AAA is among the leading causes of cardiovascular-related morbidity and mortality in US and worldwide; with an increasingly growing prevalence over the recent years\(^8-10\). The most common criterion currently used in assessing rupture risk and clinical intervention of AAAs is based on sac size, e.g. intervening when a sac diameter reaches \(\sim\)5-5.5 cm or when the growth rate reaches 1 cm per year\(^11-13\). However, studies have shown that diameter, alone, may not be the most decisive factor in decision making, where some small AAAs could rupture while some large ones could survive during the normal life expectancy of the patient\(^14-16\). Therefore, such size-based intervention methods may offer insufficient or unnecessary treatments for different patients; and new criteria are needed for making reliable and effective intervention. On the other hand, it has been shown that the development of AAA constitutes alterations in the contents and fibrillar structures of elastin and collagen, which therefore induces focal changes in the wall mechanical properties\(^9,17\). In particular, degradation in elastin content and structure has been associated with wall focal softening and balloon-like dilation, while collagen failure has been linked to sac failure\(^16,18\). Therefore, changes in the wall structural and mechanical properties are expected to take place at earlier stages of development than geometric changes that could be detected at later stages of the
disease; and using biomarkers related to wall mechanical properties may improve the disease
diagnosis.

A set of techniques for estimation of arterial stiffness is based on dilation-pressure curves\textsuperscript{19}; with
the major drawbacks of being invasive \textit{e.g.} using pressure catheters– or incurring potentially
large inaccuracies, \textit{e.g.} using non-patient-specific peripheral-to-central pressure transfer
functions\textsuperscript{20}. The second category of methods is those based on the velocity of the pulsatile wave
traveling along the aorta. Pulse Wave Velocity (PWV) has been shown to be related to the
underlying wall stiffness by Moens-Korteweg formula\textsuperscript{21–23}. The current clinical gold-standard for
PWV estimation is based on measuring the temporal pulse profiles at carotid and femoral
arteries\textsuperscript{6,24} and obtain an average velocity as over-the-skin-measured distance divided by the time
delay between the pulse profiles\textsuperscript{25,26}. The carotid-femoral-based methods are prone to
inaccuracies being primarily induced by not using the exact arterial geometry, \textit{e.g.} the true
traveled distance. In addition, assuming a single longitudinal flow direction between the carotid
and the femoral arteries leads to further inaccuracies in measuring the true distance\textsuperscript{6,27}.
Moreover, wall stiffness –and thus PWV– have been shown to vary regionally along the vascular
branch\textsuperscript{19}, and assuming an average PWV for the entire carotid-femoral segment might not thus
represent meaningful assessment. To overcome this problem, imaging techniques may be used as
non-invasive alternatives for obtaining regional wall motions\textsuperscript{28–30}. In particular, the ultrasound-
based method of Pulse Wave Imaging (PWI) has recently been developed by our group, aiming
at obtaining the regional PWV non-invasively. The PWI feasibility studies have been performed
on different applications such as normal and aneurysmal murine aortas\textsuperscript{31,32} and human aortas in
vivo\textsuperscript{29,33,34}, human carotid in vivo\textsuperscript{35}, aneurysmal and hypertensive patients\textsuperscript{36,37}, canine aorta ex
vivo\textsuperscript{38}, experimental phantoms\textsuperscript{29,39}, in fully-coupled fluid-structure interaction (FSI) aortic simulations\textsuperscript{39–44}, and in comparison against applanation tonometry\textsuperscript{29}.

As indicated above, there are focal changes in AAA wall stiffness during disease progression, and therefore –based on the Moens-Korteweg equation– the PWV is also expected to vary regionally. The performance of the PWI method could be compromised in applications involving sites of focal changes in wall composition and geometry, such as arterial branching, aneurysms and stenotic lesions, where precise detection of propagating waves becomes challenging due to the disruptive, complex patterns of reflective waves\textsuperscript{32,45}. Obtaining AAA samples at different stages of the disease with precisely controlled geometry and material properties in vivo is impossible. Therefore, simulation studies could deem as an efficient surrogate to simulate the pulsatile flow in idealized human straight and AAA geometries with different sac wall stiffness and to gain insights on the effects of geometry and material properties on the pulse wave propagation and velocity.

Conventional Finite-Element Method (FEM) simulations of vascular biomechanics have primarily relied on Solid-State (SS) modeling of blood vessels experiencing static or pulsatile internal pressure replicating the hemodynamic effects\textsuperscript{46}; and were not suitable for studying the Fluid-Solid Interaction (FSI) effects in the aorta during the AAA development such as the vortex formation. Furthermore, even the analysis of the solid domain in SS simulations may not be as accurate as when the fluid interaction is considered. For instance, the wall stress has been shown to be underestimated by as much as 20\% when SS simulation is used compared to similar results obtained from FSI analysis\textsuperscript{47}. Largely owing to advancements in simulation tools and computational powers over the past few decades, FSI simulations overcome this limitation by modeling both the fluid and solid domains and their interactions. Fully-coupled FSI simulations
using patient-specific geometries and anisotropic finite strain constitutive relations have been carried out for healthy and pathological arteries\textsuperscript{48-51}, aiming at computing biomechanical properties such as fluid velocity and pressure, and wall displacement and stress\textsuperscript{52,53}.

FSI simulations using patient-specific AAA geometries have been previously performed to assess the risk of rupture by measuring the wall stress\textsuperscript{49,50,54-57}. These studies have proposed that wall stress reliably predicts potential rupture, as opposed to the traditionally used marker of AAA size. Fully-dynamic FSI simulations can be used to quantify pulsatile wall radial displacement and its axial motion along the aorta over time, and to calculate the PWV. There have been very few studies reporting temporal data on arterial radial wall displacement during a dynamic FSI simulation but without recording of spatial data, and therefore no PWV measurements were made\textsuperscript{47}. Even though the aortic tissues have been shown to behave under anisotropic, viscoelastic, nonlinear and inhomogeneous regimens\textsuperscript{10,16,50,51,54,58,59}, the majority of arterial FSI simulation studies have been done on simplified material properties such as isotropic, elastic, linear and homogenous. Previous studies using Coupled Eulerian-Lagrangian (CEL) FSI simulations of aortic pulse wave propagation have been reported by our group; particularly on validating against phantom and \textit{in vitro} canine studies\textsuperscript{38,39,60} and on detecting wall focal inclusions\textsuperscript{41,44,61}.

In this study, CEL simulations of human aortas with straight as well as AAA geometries of different sac stiffness were performed to obtain the fluid-induced pulsatile wall deformations. Modeling the aortic pulsation can therefore provide additional insight into the pulse wave analysis in arteries with irregular geometry and wall properties, and can be used as a guideline for animal and human examinations \textit{in vivo}. 

2. Materials and methods

2.1. Modeling parameters

Human aortas with idealized AAA geometry were considered for modeling the arterial pulse wave propagation under cyclic fluid flow, Fig. 1. The idealized geometry is based on the Eurostar standard dimensions using realistic pre-operation measurements on 3,413 patients with infra-renal abdominal aortic aneurysms. In order to reduce the computational burden, a quarter of the entire axis-symmetric 3D model was used, Fig. 2. The Coupled Eulerian-Lagrangian (CEL) explicit solver in Abaqus 6.11-1 (Simulia, RI, USA) was used to perform the fully-coupled Fluid-Solid Interaction (FSI) simulations in order to capture the mutual effects of the fluid and wall pulsatile wave motions on each other. The Eulerian domain was chosen large enough that it could remain inclusive of all Lagrangian elements at the maximum strain levels throughout the pulsatile motions, as required by the CEL solver. The fluid was modeled as water with material properties of density of $\rho = 1000 \, \text{kg/m}^3$, reference sound speed of $c_0 = 1483 \, \text{m/s}$, and viscosity of $\mu = 1 \, \text{mPas}$. The wall material properties were defined to mimic those from similar AAA phantoms being used in our other ongoing studies. Tensile and compressive mechanical testing was performed on specimens within a strain range of $\varepsilon=0$–10 % at rate of $\dot{\varepsilon}=1 \, \%/s$, using an Instron 5848 microtester (Instron®, MA, USA) and ARES-G2 rheometer (TA Instrument, DE, USA) instruments, respectively. Given the rubber-like behavior of the phantom wall, the material behavior was defined as hyperelastic, Fig. 3. Using the Abaqus material assessment, a 3rd order Ogden function was found to provide the best description of the experimental strain-stress data from the combined tensile-compressive testing:
where $\lambda_i$ is the deviatoric principle stretches, and $\alpha_n$ and $\beta_n$ are the material constants. A Digital Image Correlation (DIC) system (*Correlated Solutions Inc.*, SC, US) was used to measure the surface axial and lateral strain fields under a uniaxial tension in order to estimate the Poisson’s ratio. Each axial and lateral strain was calculated as the average of strain measurements at 9 different local points on the specimen’s surface. The accuracy of DIC measurements was verified under a zero strain field, *e.g.*, rigid body motion. The Lagrangian boundary conditions (BCs) were applied by fixing all 6 DOFs on both ends of the tube. On the Eulerian domain, symmetrical BCs were applied as symmetric planes, a free inflow BC on the tube inlet, and non-reflecting and zero-pressure on the tube outlet. A sinusoidal velocity with a magnitude of 3.63 m/s and a frequency of 5.4 Hz was applied as initial condition to the tube inlet. The flow parameters were chosen relevant to similar ongoing experimental studies on aortic phantoms and animal aortas in vitro. Two additional simulations were performed on aortas with the same geometry with the wall stiffness at the sac zone (between sections 3 and 5 in Fig. 1) to be 1.5 times lower (softer sac) and 3 times higher (stiffer sac). Assuming a similar hyperelastic regimen of the material behavior, the same experimental data obtained previously were mathematically scaled down (or up) to model softer (or stiffer) sacs, respectively, before the hyperelastic parametric model fit was performed for material definition in Abaqus.

### 2.2. Computational Framework

A Precision T7600 workstation with Intel® Xeon® CPU (16 cores) @3.10 GHz and 96 GB RAM was used to perform the computations. The mesh topology on the AAA quadrant axis-
symmetric geometry contained a total of 233520 elements, leading to a total computation time of 180 h for a simulation time of 0.5 s (a time increment in the order of 1e-07 s). An output extraction frame rate of 1000 fps was used.

2.3. Output Analysis

The radial component of the wall displacement, \( U_y \), and the axial component of the fluid velocity, \( V_z \), were extracted on particular nodal paths on the lower wall (Lagrangian) and neighboring fluid inside the lumen (Eulerian), respectively (Fig. 2). The data were plotted over time to obtain the spatio-temporal maps, allowing an easy, full visualization of the wave propagations and qualitative understanding of the local pulse wave propagations. Furthermore, the spatio-temporal maps were used to calculate the pulse wave velocity (PWV) of the waves as the slope of the linear fit to the wave peak points \(^{39,41,44}\).

3. Results

3.1. Material Characteristics

The average tensile-compressive stress-strain experimental data for the strain range of \( \varepsilon=-10\text{–}10\% \) are shown in Fig. 3.A. Also shown are the fitted strain-stress data based on different hyperelastic models, e.g. Marlow, Mooney-Rivlin, 3\(^{rd}\) order Ogden and Neo-Hookean, within an extrapolated strain range of \( \varepsilon=-20\text{–}20\% \), to be on the safe side of capturing possible strains applied during the present applications. The fitting results showed the 3\(^{rd}\)-order Ogden model yielding the best description of the experimental phantom stress-strain relationship while also
describing the high-strain stiffening which is typical to fibrillar soft biological tissues such as the aortic wall. The material parameters obtained from fitting to the experimental data, as well as to the scaled-down and scale-up data (i.e. softer and stiffer materials), are listed in Table 1. Figures 3.B and 3.C show the DIC-based axial and lateral strain fields, respectively, for a representative specimen. Linear fitting to the average biaxial strain data resulted in Poisson’s ratio of \( \nu = 0.49 \pm 0.06 \) (n = 6), which is consistent with the nearly-incompressible properties of soft tissues. A wall Poisson’s ratio of \( \nu = 0.49 \) was used in the material properties definition.

3.2. AAA Geometry with the Softer Sac

Figure 4 shows spatio-temporal maps of the wall displacement vertical component and fluid velocity axial component, obtained on the AAA geometry with the 1.5 softer sac. The sac region is marked on the maps based on the known vessel length and dimensions. The spatio-temporal maps of both the wall displacement and fluid velocity indicate the existence of the sac zone. The wall displacement map (Fig. 4.A) shows the sac to undergo higher displacement magnitudes with multiple forward and reflected waves that only travel within the sac zone boundaries. The homogenous propagation of the main forward wave is also seen to be disrupted at the sac region; however, it continues to propagate along the lumen post-sac. Also shown in the same figure is the regional PWV estimates on the forward and reflected waves; showing the pre-sac reflected waves to travel slower (e.g. \( \text{PWV} = 5.8 \pm 1.30 \, \text{m/s} \)) than the post-sac forward waves (e.g. \( \text{PWV} = 9.39 \pm 0.54 \, \text{m/s} \)). Short-traveling waves are also shown within the sac region show the slowest wave propagation (e.g. \( \text{PWV} = 4.55 \pm 1.91 \, \text{m/s} \)).
3.3. AAA Geometry with the Sac of the same Stiffness

Figure 5 illustrates a still frame of the fluid velocity magnitude vectors, $V$, on AAA geometry with the stiffness of the sac to be the same as the stiffness of the rest of the aortic wall. The figure shows the propagation of a main forward-traveling fluid wave which becomes turbulent inside the sac, as characterized by the creation of vortices at the beginning of the sac; an observation that has also been previously reported by others. The time of vortex formation coincides with that of the arrival of the wall displacement forward wave at the front end of the sac which causes the generation of a reflected wave on the wall (not shown). Later on, a severely attenuated chaotic motion was depicted across the wall, corresponding to a steadier and less turbulent flow past the peak of the inlet flow profile, until the next main forward wave comes in from the following cycle. The spatio-temporal maps of the wall displacement vertical component and fluid velocity axial component are shown in Fig. 6. Overall, very similar qualitative patterns were obtained for this model as it was obtained for the soft sac model, Fig. 4. In particular, the sac zone is clearly identifiable on both the wall displacement and fluid velocity spatio-temporal maps. Also shown on the figure are the regional PWV estimates on the forward and reflected waves. The results show the pre-sac reflected waves to travel slower ($e.g.$ $PWV= 7.07 \text{ m/s}$) than the post-sac forward waves ($e.g.$ $PWV= 10.09 \pm 1.43 \text{ m/s}$). Yet, the the short-traveling waves within the sac region travel with the lowest PWV ($e.g.$ $2.84 \text{ m/s}$).

3.4. AAA Geometry with the Stiffer Sac

Figure 7 shows the spatio-temporal maps of the wall displacement vertical component and fluid velocity axial component, obtained on the AAA geometry with the 3 times stiffer sac. Compared
to the two previous AAA models with softer sacs, Figs. 4 & 6, the results here show a lower
displacement magnitude at the sac; however, the sac displacement is still higher than that of the
rest of the wall. The sac zone is not as visible as in the previous softer sac cases through the wall
displacement spatio-temporal map, however, it is still marked pronouncedly via the fluid velocity
spatio-temporal map. The pre-sac reflected and post-sac forward waves travel with PWVs of
4.52 ± 0.55 m/s and 10.87 ± 0.69 m/s, respectively.

In order to obtain better insight on the fluid-solid interactions, Figure 8 illustrates the fluid
velocity and wall displacement profiles at cross sections 2 through 6 along the aorta as shown in
Fig. 1. The results show that the fluid velocity pulse –for the most part– travels ahead of the wall
displacement pulse for about 10-20 ms. As it travels further along the lumen (i.e. increasing
section number), the flow and displacement variations become more synchronized and steady,
which could possibly be explained by further developments of the flow as it travels.

4. Discussion

Fluid-Solid Interaction (FSI) simulations could serve as a powerful and effective tool to study the
arterial pulsatile motions; however, the results from the numerical methods should be interpreted
within the limitations of such methods and under full understanding of the underlying
assumptions. Among the very first limitations that concern any numerical study are those posed
by the computational cost; only limited time of the phenomena under study can be simulated.
The simulation results in the present study were obtained over three flow cycles. In order to
minimize transient effects in the wave propagation, the PWV estimates were performed on
subsequent cycles (e.g. second and third ones).
Majority of the research on FSI modeling of aneurysmal blood vessels have focused on characterizing parameters such as wall shear stress and pressure in order to find the maximum values and the locations they take place\textsuperscript{50,53,55-57}. However, the present study aimed at characterizing the pulse wave propagation (Figs. 4, 6 and 7), and establishing markers that could be used in PWI \textit{in vivo} and carry potentials for early diagnosis of AAA sac based on focal changes. The wall displacement results (Figs. 4.A, 6.A & 7.A) indicate the existence of multiple forward and reflected waves for each inlet flow cycle. Part of such phenomena can be explained by the presence of the sac that generates dispersion in the wave propagation and additional reflections at the neck and within the sac.

Even though the idealized aneurysms were used for the aortic geometries in this study, yet a straight geometry counterpart (same aneurysm geometry with the sac removed) was considered here as a reference for obtaining theoretical insight on the structural and fluid dynamical motions. For the analogous straight geometry (\textit{i.e.} a cylinder filled with fluid and being open on both ends), the theoretical frequencies of natural vibration modes are $f_n = 2.84, 5.68, 8.52, \ldots \text{kHz}$, which fall well beyond the excitation frequency of $f=5.4 \text{ Hz}$ by the inlet flow. Therefore, multiple waves depicted here are unlikely to be part of the natural modes of vibrations. However, it has been shown that not incorporating the damping effects into the structural material properties – which exist otherwise in viscoelastic modeling– could cause oscillations in the pressure profile\textsuperscript{49}, which indicates that it can also cause oscillations in the displacement profile and can partly explain the observations seen here. However, for the same straight geometry aorta and same material properties, an average fluid velocity of $V=8 \text{ m/s}$, obtained from a cross section in the middle of the tube at a peak velocity time, leads to a Reynolds number of $Re=160000$ –well above the laminar/transient threshold of 4000– indicating highly turbulent flow; causing the fluid
motion to be turbulent that would be the primary cause of obtaining reflective waves in the wall. A similar Reynold’s number may not be obtained for the saccular geometry, however, existence of geometric and material irregularities could only add to the chaotic nature of the fluid and solid motions, as partly seen by vortex formation in the AAA models (Fig. 5). Overall, it was found that both wall displacement and fluid velocity spatiotemporal profiles can be used to identify the existence of the sac; however, the wall displacement results were found to be more sensitive to sac stiffness variations. In all AAA-geometry aortas, displacement magnitudes on sac were found to be higher than the rest of the wall –even for the model with a stiffer sac– which indicates that a higher pressure gradient should have been induced inside the sac. Ongoing studies seek to characterize the fluid pressure and the pressure wave propagations as well in order to gain an insight on how pressure wave could be related to velocity and displacement waves. Using an average PWV on the forward waves in the straight model to be 10.30 m/s, the dimensions given in Fig. 1, and the material properties defined, a Young’s modulus of E=806.18 kPa was obtained using the Moens-Korteweg equation. Although the material was described here as a 3rd-order Ogden model based on the fit to the available experimental data, the low strain range, i.e. ε=0-2%, of the same strain-stress curves have measured an average Young’s modulus of E=676.92±36.81 kPa; which indicates a 19.09 % overestimation of the modulus by simulation results. Similar differences between simulation and experimental results have previously been found and explained in details in terms of several assumptions underlying each study\textsuperscript{38,39}. It should be noted that there is also artificial stiffening toward the ends of the tube because of the constrained BCs.

In the present study, the CEL solver was used to simulate the FSI formulation after being validated to provide accurate description of the wave dynamics\textsuperscript{39,41}. In CEL, the fluid dynamics
is being modeled with an Eulerian mesh domain whereas the Lagrangian mesh domain that is used to model the structural mechanics (see Appendix 1). However, another method to simulate fluid dynamics is by solving Navier-Stokes equations using the Computational Fluid Dynamics (CFD) solver of Abaqus and couple the solution to the structural one through the Co-Simulation Engine (CSE). Our preliminary study using CSE for the pulsatile flow problem has shown the method to be more computationally-intensive and more prone to numerical divergence. The study is ongoing in order to improve the CSE simulations and cross-validate the findings to those obtained here using CEL.

Furthermore, the accuracy of the results can be improved by using more physiologically-relevant modeling parameters such as material properties and input/boundary conditions. Other studies have shown that modeling the arterial wall as porous, viscoelastic, nonlinear, inhomogeneous, and anisotropic, and fluid as viscous, incompressible and non-Newtonian, affects the FSI dynamics and results in obtaining different outputs on fluid such as shear stress and pressure. More interestingly, in the present study, it was shown that incorporating the viscosity effects into the structural material properties, versus modeling the structure as purely elastic, help to reduce the oscillations in pulsatile pressure profile, and expectedly on the wall displacement. As far as the input profile is concerned, our preliminary trials of the physiological pressure and velocity profiles have shown the numerical convergence to be more difficult to achieve, as reported by others. Furthermore, the results in this study were also limited by the lack of viscoelastic material properties on the aortic walls and non-physiological boundary conditions along the aorta (soft connective tissues surrounding the aorta). Studies are being presently conducted to take into account the aforementioned factors in an attempt to obtain more physiologically-relevant aortic pulsations.
Conclusion

In this study, dynamic fluid-solid interaction (FSI), finite element simulations have been shown promising in modeling the pulsatile flow through arterial models and obtaining the wall displacement and fluid velocity pulse wave propagations. FSI simulations were performed on aneurysmal geometry aortas with softer, same stiffness and stiffer sacs, respectively, aiming at examining the effects of geometric and material changes on the pulse wave propagations and velocities. Characterizing such effects can best be performed through simulation studies given that experimental and *in vivo* specimens with precisely controlled geometry and stiffness might prove costly or impossible. The results showed that the presence of the aneurysm sac can be detected through both wall displacement and fluid velocity spatio-temporal maps. On average, the within-sac forward waves, pre-sac reflected waves, and post-sac forward waves, exhibited increasing PWV; all of which also increased with the sac stiffness. The present study is the first report on using numerical models to investigate FSI dynamics of arterial pulsations in aneurysmal aortas, and provided insights that can be used for guidance in experimental and in vivo pulse wave analysis studies.

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Appendix 1. Lagrangian and Eulerian Coordinate Systems

In the present study, motions of the particles in solid (i.e. arterial wall) and fluid (i.e. flow) domains are formulated using Lagrangian and Eulerian coordinate system descriptions, respectively.

Defining the initial position of a moving particle in the material at the reference time, $t_0$, as $\mathbf{X}$, the new position of the same particle at the current time, $t$, can be obtained as:

$$\mathbf{x} = \mathbf{X}(\mathbf{X}, t)$$  \hspace{1cm} (2)

where $\mathbf{X}$ denotes the coordinate transfer function, Fig. 9. Considering the motion of this particle in the Lagrangian coordinate system (e.g. such as for the motion of the wall material), the displacement, $\mathbf{U}$, velocity, $\mathbf{V}$, and acceleration, $\mathbf{A}$, of the particle are defined as follows:

$$\mathbf{U}(\mathbf{X}, t) = \mathbf{x} - \mathbf{X} = \mathbf{X}(\mathbf{X}, t) - \mathbf{X}; \mathbf{V}(\mathbf{X}, t) = \frac{\partial \mathbf{U}(\mathbf{X}, t)}{\partial t} = \frac{\partial \mathbf{X}(\mathbf{X}, t)}{\partial t}; \mathbf{A}(\mathbf{X}, t) = \frac{\partial^2 \mathbf{U}(\mathbf{X}, t)}{\partial t^2} = \frac{\partial^2 \mathbf{X}(\mathbf{X}, t)}{\partial t^2}$$  \hspace{1cm} (3)

The same motion of the same particle can also be described in the Eulerian coordinate system (e.g. such as for the motion of the fluid material), in which case the displacement, $\mathbf{u}$, velocity, $\mathbf{v}$, and acceleration, $\mathbf{a}$, of the particle can be defined as follows:

$$\mathbf{u}(\mathbf{x}, t) = \mathbf{x} - \mathbf{X} = \mathbf{x} - \mathbf{X}^{-1}(\mathbf{x}, t); \mathbf{v}(\mathbf{x}, t) = \mathbf{V}(\mathbf{X}^{-1}(\mathbf{x}, t), t) = \frac{\partial \mathbf{X}(\mathbf{X}^{-1}(\mathbf{x}, t), t)}{\partial t}.$$
Based on the principle of coordinate-invariance, the displacement, velocity and acceleration fields obtained from either coordinate system are equal. In finite element formulations, Lagrangian mesh topology can be used to describe the solid domains, in which, elements are affixed to and move with the particles during the material deformation. Fluid domains, however, can be generated with Eulerian mesh topology where elements are affixed in the space while material is allowed to cross in/out of the element boundaries.

\[
\ddot{\mathbf{a}}(\mathbf{x}, t) = \ddot{A}(\mathbf{X}^{-1}(\mathbf{x}, t), t) = \frac{\partial^2 \mathbf{X}(\mathbf{X}^{-1}(\mathbf{x}, t), t)}{\partial t^2}
\]  

(4)
References


26. Millasseau SC, Stewart AD, Patel SJ, Redwood SR, Chowienczyk PJ. Evaluation of
carotid-femoral pulse wave velocity: influence of timing algorithm and heart rate.

   *Hypertension.* 2005;45:222–226.

arterial pressure wave reflections and pulse wave velocity. *Artery Research.*


29. Vappou J, Luo J, Konofagou EE. Pulse wave imaging for noninvasive and quantitative

   doi:10.1038/ajh.2009.272 

assessment from multislice two-directional in-plane velocity-encoded magnetic resonance

imaging and characterization of clinically-significant vascular mechanical properties in

32. Luo J, Fujikura K, Tyrie LS, Tilson MD, Konofagou EE. Pulse Wave Imaging of Normal
and Aneurysmal Abdominal Aortas In Vivo. *IEEE TRANSACTIONS ON MEDICAL


List of Figure Captions:

Figure 1. Schematics of the Eurostar idealized human AAA aortic geometry - Cross sections 1&2 indicate the boundary condition (BC) zone, and cross sections 3&5 indicate the sac zone.

Figure 2. (A) Quadrant 3D meshed geometries, as well as, (B) 2D representation of adjacent paths on the lower wall (Lagrangian nodes) and inside the lumen (Eulerian nodes) to be used for extraction of the wall displacement and fluid velocity outputs, respectively.

Figure 3. Mechanical testing of the phantom wall: (A) Experimental phantom stress-strain relationship along with the fits of different hyperelastic models as indicated, (B) DIC-obtained axial strain field, (C) DIC-obtained lateral strain field.

Figure 4. Spatio-temporal maps from the aneurysm geometry aorta with the sac walls to be 1.5 times softer than the rest of the walls of the aorta: (A) Vertical component of the wall displacement, (B) Axial component of the fluid velocity.
Figure 5. Vector representation of fluid velocity magnitude at the entrance of the sac on t=205 ms. The formation of the vortex in the fluid corresponds to the formation of the reflection wave in the wall.

Figure 6. Spatio-temporal maps from the aneurysm geometry aorta with the sac walls to possess the same stiffness as the rest of the walls of the aorta: (A) Vertical component of the wall displacement, (B) Axial component of the fluid velocity.

Figure 7. Spatio-temporal maps from the aneurysm geometry aorta with the sac walls to be 3 times stiffer than the rest of the walls of the aorta: (A) Vertical component of the wall displacement, (B) Axial component of the fluid velocity.

Figure 8. Wall displacement (blue) and fluid velocity (green) for cross sections 2-6, in each row from left to right, respectively, for (A-E) AAA aorta with 1.5 time softer sac, (F-J) AAA with same stiffness sac, and (K-O) AAA with 3 times stiffer sac.

Figure 9. Illustration of particle kinematics between the reference time, $t_0$, and current time, $t$. 
Figure 1. Schematics of the Eurostar idealized human AAA aortic geometry - Cross sections 1&2 indicate the boundary condition (BC) zone, and cross sections 3&5 indicate the sac zone.
Figure 2. (A) Quadrant 3D meshed geometries, as well as, (B) 2D representation of adjacent paths on the lower wall (Lagrangian nodes) and inside the lumen (Eulerian nodes) to be used for extraction of the wall displacement and fluid velocity outputs, respectively.
Figure 3. Mechanical testing of the phantom wall: (A) Experimental phantom stress-strain relationship along with the fits of different hyperelastic models as indicated, (B) DIC-obtained axial strain field, (C) DIC-obtained lateral strain field.
Figure 4. Spatio-temporal maps from the aneurysm geometry aorta with the sac walls to be 1.5 times softer than the rest of the walls of the aorta: (A) Vertical component of the wall displacement, (B) Axial component of the fluid velocity.
Figure 5. Vector representation of fluid velocity magnitude at the entrance of the sac on $t=205$ ms. The formation of the vortex in the fluid corresponds to the formation of the reflection wave in the wall.
Figure 6. Spatio-temporal maps from the aneurysm geometry aorta with the sac walls to possess the same stiffness as the rest of the walls of the aorta: (A) Vertical component of the wall displacement, (B) Axial component of the fluid velocity.
Figure 7. Spatio-temporal maps from the aneurysm geometry aorta with the sac walls to be 3 times stiffer than the rest of the walls of the aorta: (A) Vertical component of the wall displacement, (B) Axial component of the fluid velocity.
Figure 8. Wall displacement (blue) and fluid velocity (green) for cross sections 2-6, in each row from left to right, respectively, for (A-E) AAA aorta with 1.5 time softer sac, (F-J) AAA with same stiffness sac, and (K-O) AAA with 3 times stiffer sac.
Figure 9- Illustration of particle kinematics between the reference time, \( t_0 \), and current time, \( t \).
Table 1. The material constants of the 3rd order Ogden model for the soft, normal and stiff walls, as obtained by Abaqus material behavior data fitting module.

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