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In-vitro and in-silico modelling of hemodynamics in a deformable aorta.

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ABSTRACT
The work shows the setting up of an in-vitro and in-silico model of the aorta. Tests are carried out to analyse the in-silico model capability to simulate the aorta dynamics. Results show the need of improving in-silico model to also simulate the diastolic stage.

1 INTRODUCTION

The studies on cardiovascular diseases are evermore based on in-vitro as well as in-silico analysis aimed at reproducing physio/pathological conditions in specific anatomical districts [1, 2]. The use of anatomical replica has benefited of a significant improvement over the last decade, mainly due to enhanced techniques, such as the high resolution imaging analysis and the addictive manufacturing thanks to the rapid prototyping by 3D printing [3]. On the other hand, high performance computers allow to a large number of sophisticated software to simulate numerically the effect of deformable body on fluid fields by coupling the solid and the fluid domain [4].

The present work concerns the development of a model that joins the experimental and the numerical approach to simulate the dynamics of the aorta from its root to the distal thoracic segment. The study is aimed at setting up the in-vitro and in-silico models of the vessel and assessing the numerical model with experimental results obtained in a suitably designed hydro-mechanical pulse duplicator that houses the replica of the aorta and is able to reproduce physiological flow and pressure within a simplified systemic circulation.

2 MATERIAL AND METHODS

The anatomy of the replica is parametrized on the basis of 7 characteristic lengths routinely measured by clinicians by means of MRI acquisitions. The silicone phantom of the vessel has been made by pouring bicomponent silicone (shore-hardness 40) into 3D printing mould with a core in Acrylonitrile butadiene styrene (ABS) inside.

Three probes located in three characteristic sections of the replica, i.e. upstream of the aortic valve, at the beginning of the arch, and at the end of the thoracic aorta, measure the fluid pressure with a sampling acquisition of 100 Hz, whereas, upstream of the aortic valve, a magnetic flowmeter (Transonic System Inc.) accurately measures the inflow.

A Sony HXR-NX5E camera, mounted over the phantom records at 50 fps the aorta motion, which for this purpose has been marked along the main axis and in several sections (see Figure 1a). A suitable script implemented in Labview measures the deformation of the marked lengths during the cardiac cycle by imaging analysis. A test has been carried out imposing cardiac output of 4.0 l/min, heart rate of 67 bpm, and mean aortic pressure of 100 mmHg, in order to analyse the dynamics of the arch. The test fluid is buffer saline solution.

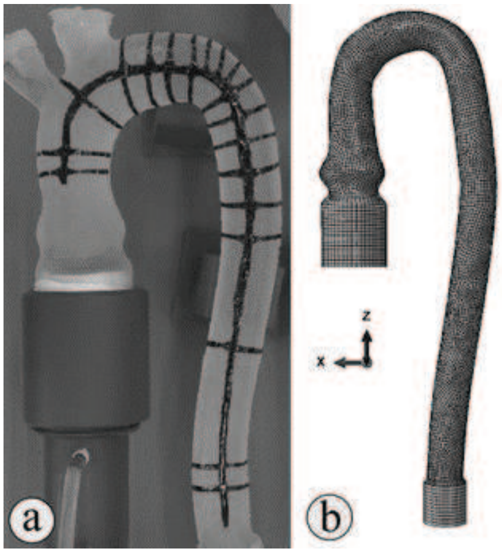

Figure 1. Replica of thoracic aorta. a) silicone phantom housed into the pulse duplicator. b) numerical aorta.

Figure 1b shows the numerical representation of the aorta. The mesh setup as well the simulation of the experiment are carried out by means of Abaqus (Abaqus Simulia – Dassault Systemes). To simulate the silicone of the phantom, a hyperelastic model, suitably calibrated, is assigned to the structural domain. The aorta experiences cycling deformation by means of a coupled fluid model reproducing the stream through the artery. In order to reproduce the same hydrodynamic condition of the experiment, the boundary conditions of the system are flow and pressure measured at the inlet and the outlet of aorta, respectively. The comparison between measured and predicted deformation and pressure along the arch is the mean to establish the capability of the in-silico model to replicate the real experiment.

3 RESULTS AND DISCUSSION

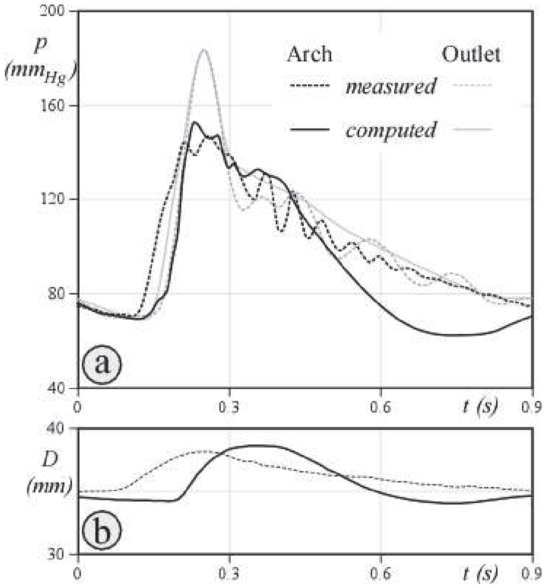

Figure 2. Comparison between in-vitro (black dashed line) and in-silico (black solid line) model at the beginning of the aortic arch. a) predicted pressure and measured pressure, and b) predicted and measured aorta diameter, D. In panel a) grey lines are the outlet pressure.

Figure 2a shows the comparison between computed and measured pressures. During systole, which in the experiment ranges between 0 and 0.32 s, the predicted pressure in the arch fits well the recorded pressure despite of the fact that computed pressure shows a small delay in the early systolic stage, whereas late diastole shows computed pressure lower than the measured one. The reasons of such a discrepancy during the late diastole are under investigation, although the first results suggest that the numerical model is highly sensitive to the artificial damping added in the model itself to stabilize the numerical method. This speculation is partially confirmed by the comparison between the measured and computed deformation of the arch diameter shown in Figure 2b. Also in this case, numerical results captures the range of the measured deformation, but the two curves show a delay due to the additional numerical viscosity.

As expected, the velocity distribution along the arch exhibits some unrealistic features, i.e. unwanted backflow during the diastole. This issue is going to be addressed enhancing the stability of the numerical model, thus reducing the artificial damping. In this way, the delay in the mechanical response of the aorta should reduce and physiological flow waves and velocity distribution would restore in diastolic stage. The development of the present model, that blends together in-vitro and in-silico metodology, is the first step toward the setting up of an integrated apparatus able to fully study migration and residual leakage of stents in the aortic arch, i.e. to recognize their potential failure conditions.

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