35th Annual International Conference of the IEEE EMBS
Osaka, Japan, 3 - 7 July, 2013

An hemodynamic work bench for in-vitro measurements in arterial bifurcations: experimental results and comparison with the output of a simplified CFD model

D. Suárez-Bagnasco, R. L. Armentano, G. Balay C. Negreira

***Abstract*—To quantify fluid-structure interactions in arterial walls, from a biomechanical standpoint, a complete characterization of blood flow, shear stress in the interface between blood and endothelium, wall elasticity and wall stresses distribution are needed.**

A new specific multi-parameter measurement system for an in-vitro characterization of the biomechanics and hemodynamic in arterial bifurcations is described.

Some experimental in-vitro results, under near physiologic pulsated flow, are shown for a physical model of arterial bifurcation surgically implemented from a sample of fresh porcine aorta.

In-vitro results are compared with in-silico results obtained from a simple CFD model of the above mentioned bifurcation.

I. Introduction

Arterial behavior in-vivo, as well as the biomechanical properties of the arterial wall, are influenced by the interaction between blood flow and the arterial wall endothelium, arterial wall elasticity and the stress-strain distribution inside the arterial wall and surrounding tissues, amongst other factors (e.g. viscosity).

To quantify these interactions both from biomechanical and hemodynamic standpoints, a complete characterization and modeling of the dynamic of arterial wall, blood flow, shear wall, wall elasticity and wall strain-stresses are needed. Characterization can be done in-vivo and in-vitro. In-vivo by means of invasive and non-invasive (tonometry, pulsed Doppler, pulse wave velocity, elastographic) methods. In invasive methods, suitable instrumented animal specimens are used to perform the measurements on a given arterial segment. In-vitro, the characterization can be done by means of a work bench simulators in which arterial segments are intercalated in closed fluid circuits that emulates the main features of systemic arterial circulation [1] [2] [3] [4]. A sound in-vitro characterization of arterial segments (including pressure and flow patterns) may furnish valuable information to be applied in-vivo [5]. A specific experimental set-up is needed in order to be able to work with arterial bifurcations under near physiological pulsated flow conditions. Measurements obtained from cryopreserved [6] and fresh arterial bifurcation samples subjected to different hemodynamic regimes allows the characterization of the biomechanical properties of arterial segments. These are obtained by parameter estimation and can be employed in a digital simulation model of the interaction between structure and fluid. Velocity profiles obtained in the work-bench can be compared with the ones obtained by computer fluid dynamics (CFD) simulations with moving wall.

The main purposes of the present paper are:

-Describe the in-vitro hemodynamic work bench simulator for multi-parameter characterization of arterial bifurcations under near physiological pulsated flow conditions.

-Show some representative experimental results for a physical model of arterial bifurcation surgically implemented from a sample of fresh porcine aorta.

-Compare some in-vitro with in-silico results obtained from a simple CFD model of the abovementioned bifurcation.

II. Hemodynamic Work Bench Simulator

The closed circuit of the hemodynamic work bench simulator (HWBS) is composed by a pump, the tubing between the pump exit and the entrance to the arterial bifurcation, the arterial bifurcation being studied, tubing between each branch of the bifurcation exit and the entrance to a fluid reservoir, adjustable constrictions on each branch, a fluid reservoir with mean pressure adjustment, and a returning tubing from the reservoir to the pump (Fig. 1).

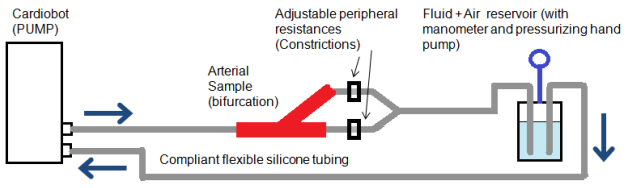

Figure 1. HWBS fluid circuit schematic diagram. Compliant silicone tubing is used for the circuit.

An artificial heart named Cardiobot is used as the pulsated pump, with frequency and ejection volume adjusted at demand. Cardiobot is an electronically controlled programmable pump, specially developed in our laboratory (see authors affiliations) [2] [7] in order to generate different

*Reserach supported by ANII and PEDECIBA.

D. Suárez-Bagnasco is with the Laboratorio de Acústica Ultrasonora (LAU), Instituto de Física, Facultad de Ciencias (FCIEN), Universidad de la República (UdelaR), Montevideo, Uruguay. (phone: +598225258624; fax: +598225250580; e-mail: dsb@ieee.org).

R.L. Armentano is with the Departamento de Fisiología, Facultad de Medicina (FMED), UdelaR, Montevideo, Uruguay. (e-mail: armen@fmed.edu.uy).

G. Balay is with the LAU, Instituto de Física, FCIEN, UdelaR, Montevideo, Uruguay. (e-mail: gbalay@fisica.edu.uy).

C. Negreira is with the LAU, Instituto de Física, FCIEN, UdelaR, Montevideo, Uruguay. (e-mail: carlosn@fisica.edu.uy).

flow and pressure patterns in the HWBS. For each branch, there is an independent restriction located far away from the corresponding branch output. Each restriction introduces localized mechanical impedance mismatch that produces reflected waves. The combination of upstream and downstream pressure pulses produce global pressure patterns in the arterial sample. Measurements are done after a steady regime of pulsated flow is attained in the circuit. The already established dynamic regime and a time reference signal generated by the Cardiobot (trigger at the beginning of each systole), allows relocation and exchange of sensors during measurements (for example, A-Scan and Pulsed Doppler measurements can be done in the same region alternating probe placement). Samples are mounted on an arterial bifurcation fixing system (couplers and adjustable arms) and submerged in a physiological solution pool. This pool is rounded by an external water pool, where temperature control can be done (Fig. 2).

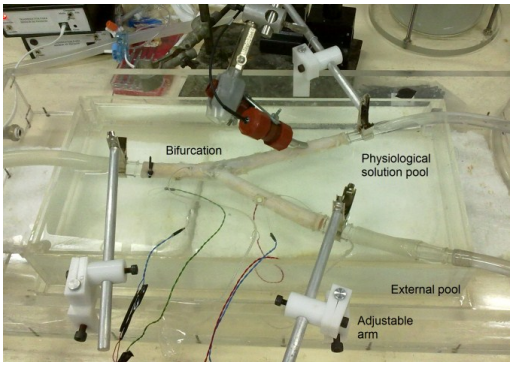

Figure 2. Arterial bifurcation in the physiological solution pool.

The circuit can be filled with physiological solution or blood treated with EDTA sodic. Temperature control can be done, with a set-up point of 37°C. The reservoir is a glass flask containing the fluid and air. It's mean pressure can be adjusted at constant values by means of a hand pump and a differential pressure sensor. With no flow and no transmural pressure in the circuit, A-Scan ultrasonic measurements are done over the bifurcations with a 15MHz Panametrics V313 transducer probe fixed on a mechanical graduated positioning system (Fig. 3). A custom algorithm in used to obtain internal and external diameters and wall thickness [1].

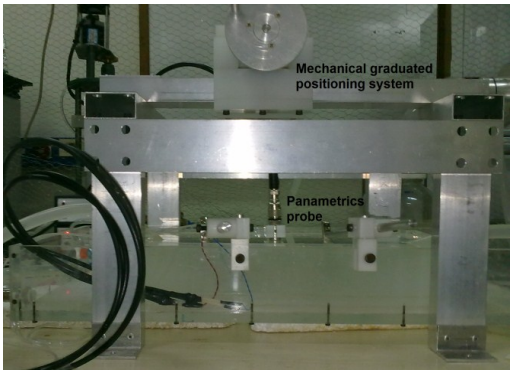

Figure 3. Mechanical graduated positioning system (in x, y and z axes) and Panametrics V313 probe during morphometric measurements..

After that, several hemodynamic conditions are simulated and measured, adjusting the flow pattern of the Cardiobot, reservoir pressure and the variable restrictions. External diameter measurements are taken at suitable points in each segment of the arterial bifurcation using a gold-standard System 6 Mainframe equipped with sonomicrometer ultrasonic modules [8] and ultra resolution techniques [5]. Pressure waveforms are obtained using Königsberg sensors (with high frequency response) located near diameter measurement points. Sensors are sutured so that sensing surface points to the lumen in order to obtain the hydrostatic component of the pressure. Velocity profiles at different locations are obtained using a multi-gate Doppler system (DOP, Signal Processing SA). Micro-spheres powder (scatterers) are used when the hemodynamic work bench simulator (HWBS) circuit is filled with a physiological solution or water-glycerine mixture fluid. Cross sectional flow is measured using suitable diameter perivascular transit time flowmeters (Triton flowmeters). A Statham pressure sensor (located in the tubing near the common segment) is used for monitoring and initial adjustments procedures. Signals are acquired at 3kHz using 12-bit data acquisition modules (NI USB-6009 and LabJack U3-HV) operated from a specially developed Matlab application. Two Tektronix TDS2024B, 100MHz, 4 channel digital oscilloscopes are used for initial adjustments and for monitoring purposes. Pressure and diameter samples are processed in a computer running a specific developed algorithm in order to obtain biomechanical properties of the wall by means of model parameter adjustment [9].

For the reported experiments, a physical model of arterial bifurcation was used, surgically implemented from a sample of fresh porcine aorta (Fig. 4).

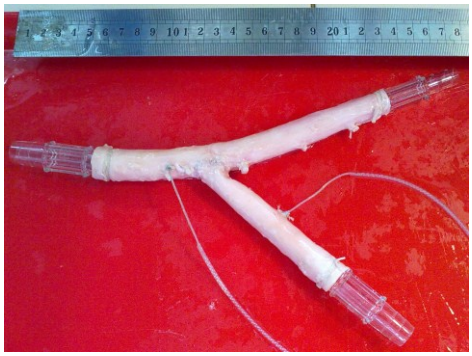

Figure 4. Surgically constructed physical model of arterial bifurcation. The common segment and the two branches are in the same plane, with a bifurcation angle of 60 degrees between branches. Tubing couplers and two Konigsberg sensors are shown.

III. EXAMPLES OF DATA OBTAINED IN-VITRO

Static and dynamic measurements of diameters, pressures, flows and velocity profiles were done in several locations in the common segment and branches of the bifurcations. Fig. 5 shows the simultaneous acquisition of diameter, pressure and cross sectional flow for a location in the common segment.

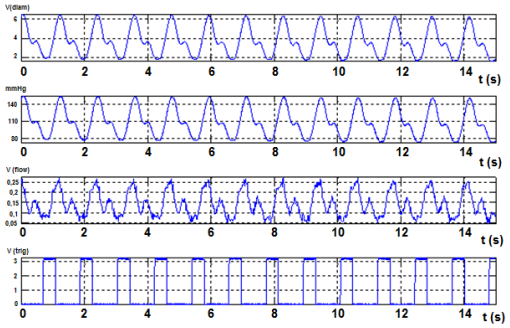

Figure 5. Diameter, pressure and flow for common segment. Time reference signal generated by Cardiobot at each systole can be seen at the bottom.

For this experiment, a physiological solution was used in the circuit, with diluted suspension of microspheres (doppler scatterers). Fig. 6 shows velocity profiles (without filtering) obtained by pulsed Doppler, in the same place. Vessel pulsation can be seen through the variation of the profiles width over the x axis (distance in mm).

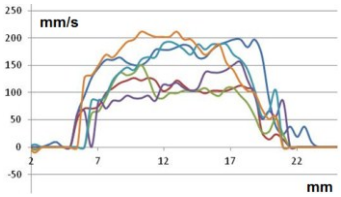

Figure 6. Raw velocity profiles obtained for the common segment at 40mm from the entrance. Vessel pulsation can be seen through the variation of the profiles width over the x axis (distance in mm). The segment presented a radius of 7,5 mm at zero transmural pressure.

IV. Examples of Data Obtained In-Silico

A simplified CFD model was constructed as a first approximation in order to compare in-vitro with in-silico results, in the framework of an improvement of a hemodynamic work bench simulator (HWBS) design (in order to work with arterial bifurcations). The spatial domain used for finite-element digital simulation [10] was determined from geometrical measurements done to the surgically constructed physical model (Fig. 7).

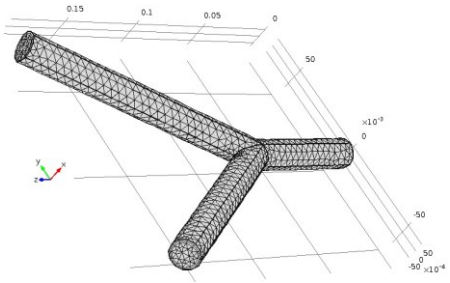

Figure 7. Perspective view of the grid used for CFD simulations.

A mean radius of 7,5mm was used for the common segment and each branch and a rigid wall condition was imposed (non elastic wall). Lengths were taken from the physical model as well as the bifurcation angle (implemented to be 60 degrees).

Time dependent Navier-Stokes equations were solved with a suitable viscosity coefficient, using a commercial software. Fixed and impervious wall conditions were imposed. In the open ends (commons segment entrance and branch exits) pressure boundary conditions were applied, taken from in-vitro experimental measurements [11]. Figure 8 shows the pulsated pressure measured in-vitro at the entrance of the common segment.

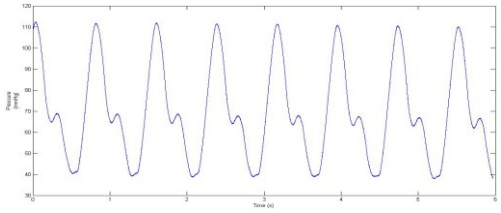

Figure 8. Input pressure waveform for the common segment entrance, with a pulsation frequency of 60 beats per minute.

Fig. 9 shows the velocity profiles obtained at different times in the common segment at 40mm from the entrance.

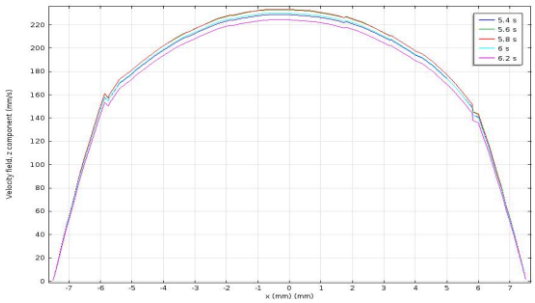

Figure 9. In-silico velocity profiles for the common segment. Velocities are expresed in milimeters per second (mm/s).

Fig. 10 shows the velocity profiles obtained at the same instant at different cross sections. Flow inversion near the outer walls of the bifurcation branches can be seen.

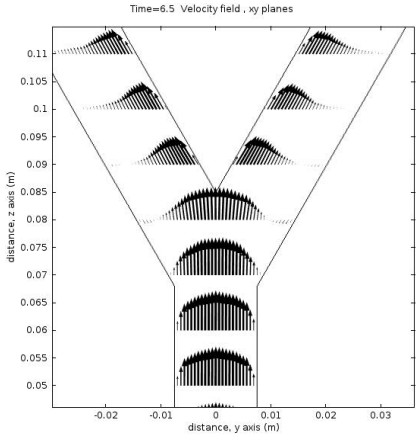

Figure 10. Velocity profiles near the bifurcation for a given time.

Fig. 11 (left) shows the relative magnitudes of the velocities for the whole bifurcation in the plane x = 0 at the same time instant of Fig.10. Flow inversion can't be seen because this figure represents absolute values. Fig. 11 (right) shows shear

rate magnitude for the same time instant. The inner walls of bifurcation branches show maximum values.

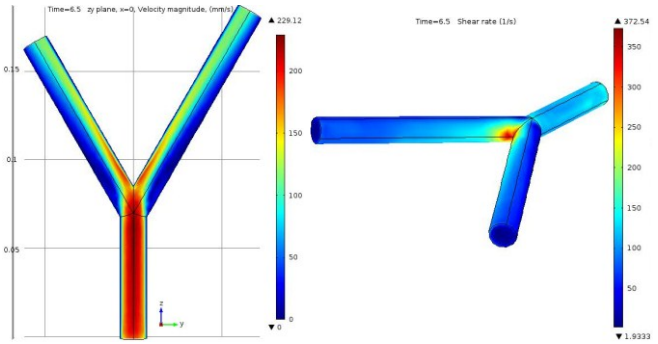

Figure 11. Left: Relative magnitude of velocity (mm/s) for a given time, in the plane x=0. Right: Shear rate magnitudes on the walls of the vessel.

V. DISCUSSION

Near-physiological flow and pressure patterns can be obtained for different kind of arterial bifurcations by means of adjusting compliant tube lengths and hydraulic resistances as well as the Cardiobot pulsation patterns. This kind of flow and pressure patterns are not strictly necessary if the intention is a biomechanical characterization of the arterial wall. However, realistic flow and pressure patterns are necessary to study fluid-structure interactions in scenarios as near as possible to in-vivo ones.

Although it is not completely equivalent to a real bifurcation, physical models like the one presented here, surgically constructed from a fresh artery, enables us to change at will the bifurcation angle, the segment lengths and the tapering.

The dimensions facilitated the execution of the in-vitro measurements, which can be later compared with digital simulation results. Inner radius of common segment, input coupler and the straight tubing are similar enough to minimize entrance lengths effects.

In-vitro experimental measurements afforded realistic waveforms of pressure that were used as boundary conditions at the entrance and exits of the bifurcation CFD model.

Velocity profiles obtained in the common segment during in-vitro measurements shows a time variation in their widths due to the radius variation of the arterial wall, as consequence of pulsed pressure and elastic compliance. Centerline velocities obtained in-silico (210 to 270 mm/s) for the common segment is greater than the one measured in-vitro (90 to 211 mm/s). This difference must be assessed taking into account experimental and numerical computation errors, pulsed doppler measurement limitations in resolution, wall radius variation, energy dissipation in the arterial tissues, and the rigid wall condition of the simplified CFD model.

The experiments reported in the present paper were done in the framework of an improvement of a previous hemodynamic work bench simulator (HWBS) design of the FCIEN-FMED group. The improvement was necessary in order to work with real arterial bifurcations and to use the experimental data as inputs in realistic digital simulation of fluid-structure (FSI) interactions. This kind of digital simulation requires a knowledge of the biomechanical wall properties. These can be obtained by adjusting the parameters of a suitable viscoelastic model of the arterial wall to pressure-diameter data [8] [12] obtained in-vitro using the HWBS and real arterial samples.

Properly fine-tuned in-silico CFD+FSI digital simulation based on in-vitro experimental data allows the determination of physical magnitudes and hemodynamic indexes (like oscillatory shear index (OSI), temporal wall shear stress gradient (TWSSG), spatial wall shear stress gradient (SWSSG), etc. [8] [13]) that can't be obtained easily in experimental in-vitro measurements.

ACKNOWLEDGMENT

Authors thanks ANII for grant BE_POS_2010_1_2588 and PEDECIBA for research support.

REFERENCES

[1] J. Brum, D. Bia, N. Benech, G. Balay, R.L. Armentano, C. Negreira, "Setup of a cardiovascular simulator: application to the evaluation of the dynamical behavior of atheroma plaques in human arteries", Physics Procedia, 2010, vol. 3, pp.1095-1101.

[2] G. Balay, J, Brum, D. Bia, R.L. Armentano, C. Negreira 2010, Improvement of artery radii determination with single ultra sound channel hardware & in vitro artificial heart system, Conf. Proc. IEEE Eng. Med. Biol. Soc., pp. 2521-4, 2010.

[3] D. Bia, R.L. Armentano, Y. Zocalo, W. Barmak, E. Migliaro, E.I Cabrera Fischer, "In vitro model to study arterial wall dymamics through pressure-diameter relationship analysis", Latin American Applied Research, 2005, vol. 35, pp. 217-225.

[4] C.D. Bertram, F. Pythoud, N. Stergiopulos, J.J. Meister, "Pulse wave attenuation measurement by linear and nonlinear methods in nonlinearly elastic tubes", Medical Engineering and Physics, 1999, vol 21, pp 155–166.

[5] W. Nichols, M. O'Rourke, Ch. Vlachopoulos, McDonald's blood flow in arteries: theoretical, experimental and clinical principles, Hodder-Arnold, London, UK, 2011.

[6] R.L Armentano, D. Bia, E. Cabrera Fischer et al., "An in-vitro study of cryopreserved and fresh human arteries: a comparison with ePTFE prostheses and human arteries studied non-invasively in vivo", Cryobiology, 2006, 52(1), pp17-26.

[7] G. Balay, "Elasticity in arterial tissues, design of an in-vitro artificial heart and a new ultrasonic method of arterial elesticity assessment, M.Sc. thesis (Physics), LAU-FCIEN-UdelaR, July 2012.

[8] K. Chandran, S. Rittgers, A. Yoganathan, "Biofluid mechanics: the human circulation", 2nd ed, CRC Press, Boca Raton, FL, 2012.

[9] L.G. Gamero, R.L. Armentano, J.G. Barra, A. Simon, J. Levenson, "Identification of arterial wall dynamics in conscious dogs", Exp Physiol, 2001, vol 86 pp 519-528.

[10] W. Press, S. Teukolsky, W. Vetterling, B. Flannery, "Numerical recipes: the art of scientific computing", 3rd Ed., Cambridge Univ. Press, N.Y., 2007.

[11] J. Anderson, "Computational fluid dynamics: the basics with applications", McGraw-Hill, N.Y., 1995.

[12] Y.C. Fung, "Biomechanics: mechanical properties of living tissue", Springer Verlag, N.Y., 1981.

[13] T. Hsiai, B. Blackman, H. Jo (Eds), "Hemodynamics and mechanobiology of endotelium", World Scientific, Singapore, 2010