Development of a triphasic pulsatile flow phantom for experimental study of abdominal aortic blood flow

Conference Paper · December 2017

CITATIONS
0

READS
265

2 authors:

Albert Yoon-Song Chong
Curtin University
7 PUBLICATIONS **21** CITATIONS
SEE PROFILE

Zhonghua Sun
Curtin University
438 PUBLICATIONS **6,738** CITATIONS
SEE PROFILE

Proceedings of One Curtin International Postgraduate Conference (OCPC) 2017 Miri, Sarawak, Malaysia, December 10 – 12, 2017

Development of a triphasic pulsatile flow phantom for experimental study of abdominal aortic blood flow

Albert Chong, Zhonghua Sun
Department of Medical Radiation Sciences, Curtin University
Kent Street, Bentley, Perth, Western Australia 6102, Australia
Telephone: +61 8 9266 3232
Telephone: +61 8 9266 2377

Corresponding Author: Albert Chong, yoon.s.chong@postgrad.curtin.edu.au

Introduction

Experimental flow phantoms are indispensable tools for in vitro studies, especially in relation to the study of stenoses, aneurysms, stent design, vessel elasticity and related haemodynamic parameters of human arterial blood flow. These phantoms are important tools for studying and validating new treatment methods such as minimally invasive endovascular stent treatments [1, 2]. Phantoms are also used to help interpret clinical scanning, validate physical measurements used in clinical practice and research, characterize machines, help in the development of new medical imaging techniques and validate numerical methods such as finite element analysis (FEA) and computational fluid dynamics (CFD). [3-5]

The requirements of phantom to reproduce realistic arterial flow waveforms were reviewed extensively by Law et al [6], Holdsworth et al [7] and Hoskins et al [8] The basic requirement is that the flow phantom must be capable of reproducing pulsatile flow mimicking human arterial flow physiology and easy to program. These authors have also provided thorough reviews of previous works. Basically, previous devices can be classified according to the type of pump used in the phantom i.e. gear, piston or roller/peristaltic. Gear pumps have been used by Groot Jebbink et al [1], Boersen et al [2] and Hoskins [8] to generate pulsatile waveforms. However, the disadvantage of gear pump is that suspended particles are easily damaged and it is sensitive to cavitation due to the grinding action of the gears. Piston pumps have been used by Holdswoth et al [7], Rudenick et al [9] and Poot et al [10] to simulate peripheral arterial flow. This class of pumps shares the general disadvantage of difficulty in programming. Modified peristaltic pumps have been used by Law et al [6] and DouVille et al [11] to generate physiological flow waveforms by mechanical manipulation of the backplate or computer control of the roller. This technique allows the generation of a limited number of waveforms. It is also difficult to program new waveforms, or produce reverse flow with this method. More recently Shkolnikov et al [12] and Neto et. al [13] developed roller-free peristaltic pump using actuators to produce pulsatile waveforms. However these pumps were developed for use in micro-fluidic applications and not suitable for blood flow investigation of carotid, abdominal and peripheral arteries. The main advantage of peristaltic pump is due to the fact that there is no contact between the pump driving components and the liquid. In addition, the pumping chamber i.e. the platinum cured silicone tubing is disposable [14], which ensures the sterility and avoids cross-contamination [15, 16]. The peristatic pump is a technology which has been

widely used in cardiopulmonary bypass (CPB) and extracorporeal membrane oxygenation (ECMO) procedures carried out over the past seven decades since its introduction in the 1950s. The man limitation of the standard peristaltic pump is that it generates approximately a sinusoidal waveform which does not confirm to human blood flow physiology. The proposed triphasic cardiac pump is an improved version and it overcomes the limitation of the standard peristaltic pump by a unique mechanical design and combined with digital control technology, it has the capability to generate triphasic waveform and is programmable.

Material and methods

A fluid circulation system as shown in Figure 1 was developed based on an improved cardiac pump driven by a 8.5Nm NEMA34 stepper motor 86HBS85 and a closed loop hybrid servo drive HBS860H (Just Motion Control, Shenzhen, China). Arduino Uno microcontroller was used to control the stepper motor and generate the appropriate flow profile. Masterflex platinum-cured silicone tubing, L/S 36 and I/P82 (Cole-Parmer,Vernon Hills, Illinois, USA)

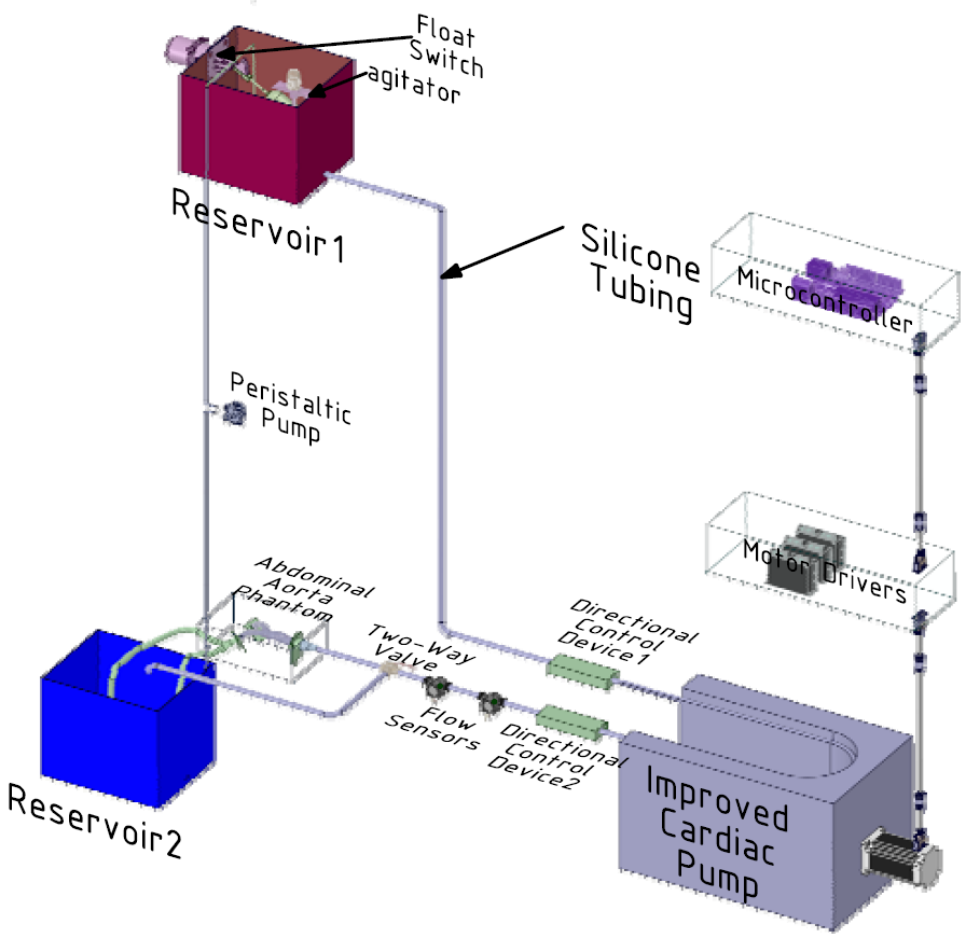

Figure 1 Fluid Circulation System

with tubing internal diameters of 9.7mm and 12.5mm were used to circulate the blood mimic fluid (BMF) made of % by weight, pure water (85.41%), pure glycerol (10.25%), Sigma D4876 Dextran (3.43%) and surfactant (0.92%), density = 1,037 Kgm^{-3} and viscosity = 4.1 (mPa s) [17]. There are two reservoirs in the flow circuit where Reservoir 1 is located at a higher elevation than Reservoir 2. The difference in heights between Reservoir 1 and Reservoir 2 in millimeters (mm) is the gauge pressure in mmH_2O. The height difference was set to 1088mm therefore a gauge pressure of 1088 mmH2O or 80 mmHg was produced at the inlet of cardiac pump and at the start of the cardiac cycle. A manually operated two-way valve located at the outlet of the cardiac pump allows the user to direct the flow either to the phantom or directly to Reservoir 2. When the cardiac pump is in operation, volume of BMF in Reservoir 1 will decrease where as volume of BMF in Reservoir 2 will increase. When the BMF level falls below a pre-determined level in Reservoir 1, the float-switch will activate the peristaltic pump to transfer BMF from Reservoir2 to Reservoir 1 thereby maintain the gauge pressure.

Flow Rate Measurements

The average volumetric flow rates were measured by means of stop watch and the amount of BMF delivered by the cardiac pump directly into the calibrated Reservoir 2. The cardiac pump was programmed to generate 60 pulses per minute. The instantaneous flow rate generated by the cardiac pump is captured by the flow sensor located between the pump outlet and two-way valve. The output from flow sensors were displayed on a display monitor via serial communication link between the Arduino and a laptop computer (Lenovo, Morrisville, North Carolina, USA). To visualize the sensor outputs, Microsoft Excel spreadsheet is used to plot the graph of instantaneous volumetric flow rate (L/min) on the vertical axis and time (s) on the horizontal axis. To facilitate Doppler ultrasound measurements of the centerline velocity, 1200 Grit (15 micron) aluminium oxide powder was suspended in the BMF using an agitator located in Reservoir 1. Ultrasound gel is applied on the outer diameter of the silicone tubing distal to cardiac pump outlet and the probe from a portable ultrasound scanner BV-520T (Shenzhen Bestman Instrument Co. Ltd, Shenzhen, China) is then placed on the tubing with an incident angle of about 30 degree to capture the centerline velocity profile.

Results

Table 1 outlines the measurement results using Masterflex tubings LS36 and IP82 where average flow rates were 1.27L/min and 1.97 L/min respectively.

Table 1. Flowrate measurement results for Masterflex Tubing LS36 and IP82

Masterflex Tubing	Inner Diameter (mm)	Time taken to fill 5L (minutes)	Flow Rate (L/min)
LS36	9.7	3:56	1.27
IP82	12.5	2:32	1.97

Figure 2 shows a typical graph of instantaneous flow rate in L/min plotted against time in second. In this case, the mean flow rate generated from Masterflex Tubing LS36 with 9.7mm inner diameter is 1.2L/min with peak systolic flow rate of 8.25 L/min and a peak diastolic flow rate of 1.68L/min.

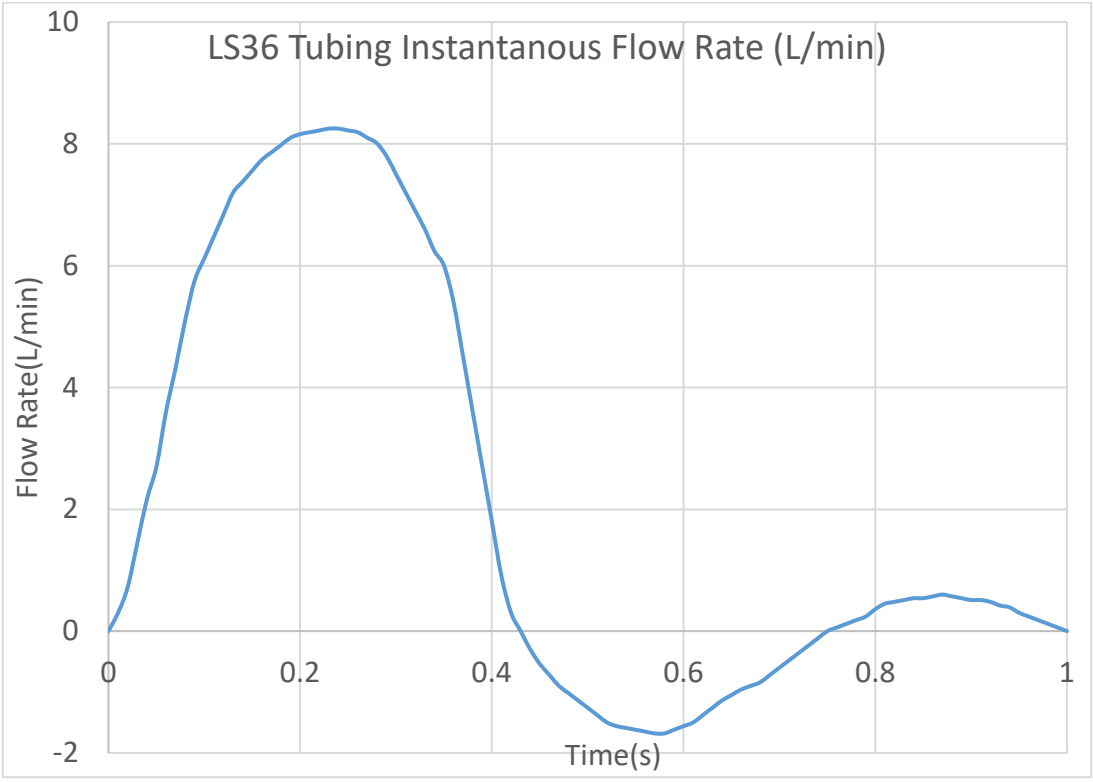

Figure 2. LS36 Tubing Instantaneous Flow Rate

Figure 3 shows the centreline velocity profile from the portable Doppler Ultrasound scanner which indicates triphasic flow profile close to the physiology of human abdominal blood flow.

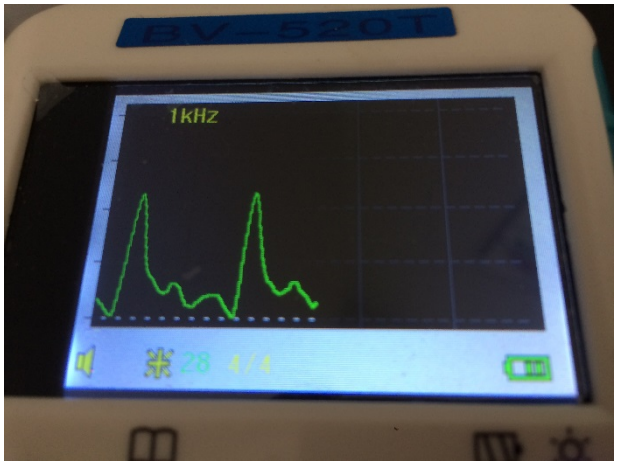

Figure 3 Display of centerline flow velocity profile from the Doppler Ultrasound Scanner

Discussions

Since the mean abdominal aortic flow rate determined by Fraser et al [18] is 13.3mL/s or 0.798L/min therefore the mean flow rate of 1.27L/min generated by the improved cardiac pump using Masterflex LS36 tubing is more than sufficient for phantom studies of abdominal blood flow. In addition, the cardiac pump flow rate and pulsae rate can be programmed to match any patient specific physiology. Further studies of flow pressures is planned by adding transducers at inlet and outlet of the improved cardiac pump. At the time of writing this manuscript, a patent search has been completed which confirmed the nolvelty of the improved cardiac pump design. Intellectual Property (IP) filling is planned for this novel cardiac pump design and computer simulation is also planned to optimise the design.

Conclusions

The newly developed cardiac pump is capable of generating triphasic waveform conforming to human physiology therefore suitable for use in experimental studies of abdominal blood flow.

References:

[1] E. Groot Jebbink, V. Mathai, J.T. Boersen, C. Sun, C.H. Slump, P.C.J.M. Goverde, M. Versluis, M.M.P.J. Reijnen, Hemodynamic comparison of stent configurations used for aortoiliac occlusive disease, Journal of Vascular Surgery, 66 (2017) 251-260.e251.
[2] J.T. Boersen, E. Groot Jebbink, M. Versluis, C.H. Slump, D.N. Ku, J.-P.P.M. de Vries, M.M.P.J. Reijnen, Flow and wall shear stress characterization after endovascular aneurysm repair and endovascular aneurysm sealing in an infrarenal aneurysm model, Journal of Vascular Surgery, (2017).
[3] M.D. Ford, H.N. Nikolov, J.S. Milner, S.P. Lownie, E.M. DeMont, W. Kalata, F. Loth, D.W. Holdsworth, D.A. Steinman, PIV-measured versus CFD-predicted flow dynamics in anatomically

realistic cerebral aneurysm models, Journal of Biomechanical Engineering, 130 (2008) <xocs:firstpage xmlns:xocs=""/>.
[4] E. Soudah, P. Rudenick, M. Bordone, B. Bijnens, D. García-Dorado, A. Evangelista, E. Oñate, Validation of numerical flow simulations against in vitro phantom measurements in different type B aortic dissection scenarios, Computer Methods in Biomechanics and Biomedical Engineering, (2013) 1-11.
[5] P. Hariharan, M. Giarra, V. Reddy, S. Day, K. Manning, S. Deutsch, S. Stewart, M. Myers, M. Berman, G. Burgreen, E. Paterson, R. Malinauskas, Multilaboratory Particle Image Velocimetry Analysis of the FDA Benchmark Nozzle Model to Support Validation of Computational Fluid Dynamics Simulations, J. Biomech. Eng.-Trans. ASME, 133 (2011).
[6] Y.F. Law, R.S.C. Cobbold, K.W. Johnston, P.A.J. Bascom, Computer-controlled pulsatile pump system for physiological flow simulation, Med. Biol. Eng. Comput., 25 (1987) 590-595.
[7] D. Holdsworth, D. Rickey, M. Drangova, D. Miller, A. Fenster, COMPUTER-CONTROLLED POSITIVE DISPLACEMENT PUMP FOR PHYSIOLOGICAL FLOW SIMULATION, Med. Biol. Eng. Comput., 29 (1991) 565-570.
[8] P.R. Hoskins, Simulation and Validation of Arterial Ultrasound Imaging and Blood Flow, Ultrasound in Medicine & Biology, 34 (2008) 693-717.
[9] P.A. Rudenick, B.H. Bijnens, D. García-Dorado, A. Evangelista, An in vitro phantom study on the influence of tear size and configuration on the hemodynamics of the lumina in chronic type B aortic dissections, Journal of Vascular Surgery, 57 (2013) 464-474.e465.
[10] K. Poots, R. Cobbold, K. Johnston, R. Appugliese, M. Kassam, P. Zuech, R. Hummel, A new pulsatile flow visualization method using a photochromic dye with application to Doppler ultrasound, Ann Biomed Eng, 14 (1986) 203-218.
[11] Y. Douville, K.W. Johnston, M. Kassam, P. Zuech, R.S.C. Cobbold, A. Jares, An in vitro model and its application for the study of carotid doppler spectral broadening, Ultrasound in Medicine & Biology, 9 (1983) 347-356.
[12] V. Shkolnikov, J. Ramunas, J.G. Santiago, A self-priming, roller-free, miniature, peristaltic pump operable with a single, reciprocating actuator, Sensors and actuators. A, Physical, 160 (2010) 141-146.
[13] A.G.S.B. Neto, A.M.N. Lima, H. Neff, C.L. Gomes, C. Moreira, Linear peristaltic pump driven by three magnetic actuators: Simulation and experimental results, 2011, pp. 1-6.
[14] N.J. Graf, M.T. Bowser, A soft-polymer piezoelectric bimorph cantilever-actuated peristaltic micropump, Lab on a Chip, 8 (2008) 1664-1670.
[15] D.J. Laser, Areview of micropumps, 2004, pp. R35-R64.
[16] B. Iverson, S. Garimella, Recent advances in microscale pumping technologies: a review and evaluation, Microfluidics and Nanofluidics, 5 (2008) 145-174.
[17] K.V. Ramnarine, D.K. Nassiri, P.R. Hoskins, J. Lubbers, Validation of a New Blood-Mimicking Fluid for Use in Doppler Flow Test Objects, Ultrasound in Medicine & Biology, 24 (1998) 451-459.
[18] K.H. Fraser, S. Meagher, J.R. Blake, W.J. Easson, P.R. Hoskins, Characterization of an Abdominal Aortic Velocity Waveform in Patients with Abdominal Aortic Aneurysm, Ultrasound in Medicine & Biology, 34 (2008) 73-80.