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Pressure pulsation in roller pumps: A validated lumped parameter model

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Abstract

During open-heart surgery roller pumps are often used to keep the circulation of blood through the patient body. They present numerous key features, but they suffer from several limitations: (a) they normally deliver uncontrolled pulsatile inlet and outlet pressure; (b) blood damage appears to be more than that encountered with centrifugal pumps.

A lumped parameter mathematical model of a roller pump (Sarns^{TM} 7000, Terumo CVS, Ann Arbor, MI, USA) was developed to dynamically simulate pressures at the pump inlet and outlet in order to clarify the uncontrolled pulsation mechanism. Inlet and outlet pressures obtained by the mathematical model have been compared with those measured in various operating conditions: different rollers' rotating speed, different tube occlusion rates, and different clamping degree at the pump inlet and outlet.

Model results agree with measured pressure waveforms, whose oscillations are generated by the tube compression/release mechanism during the rollers' engaging and disengaging phases. Average Euclidean Error (AEE) was 20 mmHg and 33 mmHg for inlet and outlet pressure estimates, respectively. The normalized AEE never exceeded 0.16.

The developed model can be exploited for designing roller pumps with improved performances aimed at reducing the undesired pressure pulsation.

Keywords: Roller pump; Lumped parameter model; Pressure dynamics; Uncontrolled pulsatile pressure; Pulsation mechanism

1. Introduction

During open-heart surgery, cardiopulmonary bypass (CPB) is used to relieve heart and lungs of their normal life sustaining functions. Roller pumps are used to undertake the functions of blood circulation in approximately 50% of CPB [1,2].

For decades, roller pumps were the most accepted pumps for CPB [3], due to their numerous key features: low cost; predictable mean flow obtained by controlling the arm speed or choosing different tubes' inner diameter sizes, a quicker response to servo-control; simpler design and thus fewer moving parts which would be subject to malfunction.

Nevertheless, they suffer from several limitations: they normally deliver uncontrolled pulsatile inlet and outlet pressure, which generates high frequency harmonics in pressure signals [4]; a phenomenon known as spallation [5,6] exists, which consists of micro-particles detaching from the tubing inner walls due to cyclic compression by the rollers; blood trauma and microemboli delivered to the patient appears to be more than that encountered with centrifugal pumps [7,8].

In the last years, computational studies and models of CPB hemodynamics have been reported. In particular, they were aimed at simulating patient hemodynamics during artificial perfusion [9,10], developing automated perfusion strategies [11–13], and optimizing the design of CPB circuits [14], but none of them was focused on the pumping element, usually modelled as a pure flow generator. A two-dimensional CFD model of a roller pump is reported in [15]. This study gave useful information about roller pump fluid dynamics; however the model did not focus on the uncontrolled pulsatility dynamics.

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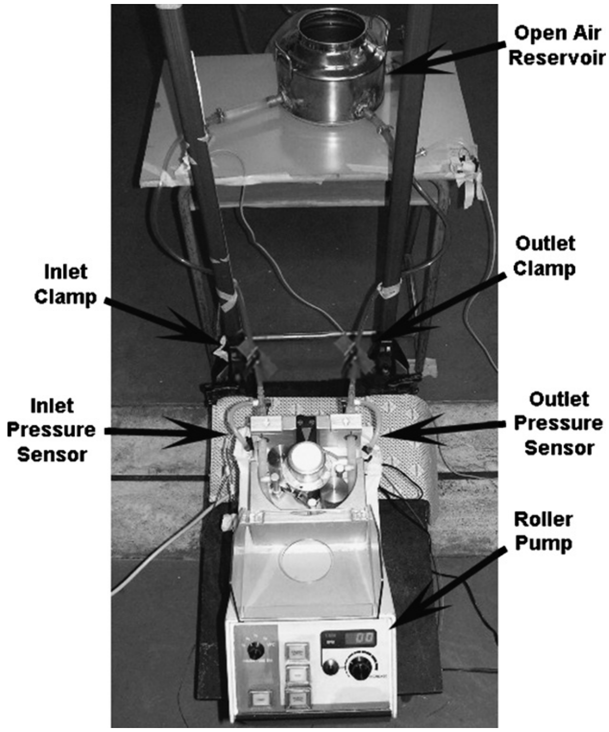

Fig. 1. Picture of the experimental setup. For characteristic dimensions see Table 1.

The aim of the present study is to provide a lumped parameter model which clarifies the uncontrolled pressure pulsatility generation mechanisms in roller pumps, thus allowing considerations on achievable performance improvements of roller-type peristaltic pumps.

2. Material and methods

2.1. Experimental setup

A twin roller pump (Sarns™ 7000, Terumo CVS, Ann Arbor, MI, USA) connected to an open air reservoir by means of two hydraulic lines, made up of 3/8 in. × 1/16 in. PVC pipes, formed the experimental setup (Fig. 1). At the pump inlet and outlet lines, screw clamps allowed the increase of the pump hydraulic load.

Pressures at the pump inlet and outlet sections were measured by using two pressure transducers (MicroSwitch 143PC05D, Honeywell International Inc., Morristown, NJ, USA), connected to the lines by means of 3-way connectors.

The experiments were conducted at room temperature. Water was used as the working fluid, and the water level inside the open-air reservoir determined the filling pressure (equal to 14 mmHg) of the hydraulic circuit.

A software package (LabVIEW™, National Instruments™, Austin, TX, USA) was used for data acquisition. Signals were recorded with a data acquisition board (AT-MIO 16E-2, National Instruments™, Austin, TX, USA) at a sample rate of 1000 Hz.

2.2. Mathematical model and parameter estimation

A complete electric analogue model, of the impedance-type [16], including the roller pump and the hydraulic lines, is shown in Fig. 2. The whole system was divided into four subsystems, each modelled by means of lumped parameters: inlet and outlet hydraulic lines, two pump rollers, and the tube into the pump housing. The rollers' full rotation period was divided into the following sub-periods (Fig. 3): the engaging and disengaging phases (1 and 3, respectively); phase 2 during which the roller feeds into the pump housing; phase 4 during which the roller rotates without pushing the tube. The same phases occur for both rollers, but they are shifted by 180°. The mathematical model was provided with time-varying signals and parameters to determine phase transitions according to the rollers' angular position as shown in Fig. 3. The dynamic model of the roller pump was implemented by using Simulink® (The Mathworks Inc., Natick, MA, USA), and a variable-step solver (ode15s, stiff/NDF) was used with maximum step size fixed to 0.001 s. The integrators in the model were initialized with the value of the hydraulic circuit filling pressure (equal to 14 mmHg) and a flow equal to zero.

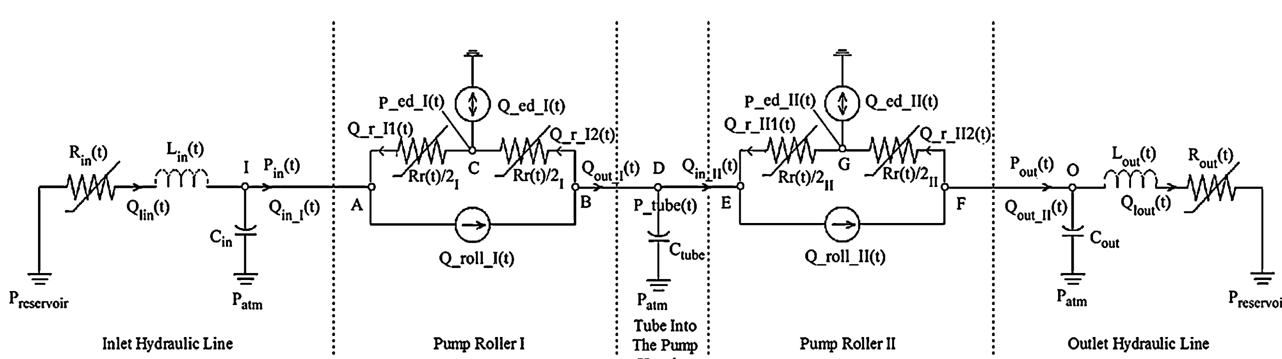

Fig. 2. Complete electric analogue model made up of four subsystems: inlet and outlet hydraulic lines (represented as L networks); pump rollers; tube into the pump housing. For symbols and values see Table 1.