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Short communication

High accuracy differential pressure measurements using fluid-filled catheters – A feasibility study in compliant tubes

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ABSTRACT

High accuracy differential pressure measurements are required in various biomedical and medical applications, such as in fluid-dynamic test systems, or in the cath-lab. Differential pressure measurements using fluid-filled catheters are relatively inexpensive, yet may be subjected to common mode pressure errors (CMP), which can significantly reduce the measurement accuracy. Recently, a novel correction method for high accuracy differential pressure measurements was presented, and was shown to effectively remove CMP distortions from measurements acquired in rigid tubes. The purpose of the present study was to test the feasibility of this correction method inside compliant tubes, which effectively simulate arteries. Two tubes with varying compliance were tested under dynamic flow and pressure conditions to cover the physiological range of radial distensibility in coronary arteries. A third, compliant model, with a 70% stenosis severity was additionally tested. Differential pressure measurements were acquired over a 3 cm tube length using a fluid-filled double-lumen catheter, and were corrected using the proposed CMP correction method. Validation of the corrected differential pressure signals was performed by comparison to differential pressure recordings taken via a direct connection to the compliant tubes, and by comparison to predicted differential pressure readings of matching fluid–structure interaction (FSI) computational simulations. The results show excellent agreement between the experimentally acquired and computationally determined differential pressure signals. This validates the application of the CMP correction method in compliant tubes of the physiological range for up to intermediate size stenosis severity of 70%.

1. Introduction

Differential pressure-based parameters have become indispensable for functional assessment of coronary arteries stenosis (Hamilos et al., 2010), or aortic stenosis (Hays et al., 2006; Jayne et al., 1993). Direct and accurate measurement of arterial blood flow is complicated and quite involved task, and it is therefore convenient, as well as inexpensive, to utilize differential pressure measurement as a surrogate for flow. For this reason, the capability to measure intraluminal differential pressure with a catheter at high accuracy is of great interest.

Recently, our group introduced a new method for high accuracy differential pressure measurement, via an inexpensive, highly accurate fluid-filled double-lumen catheter connected to a differential pressure transducer (Rotman et al., 2014). Gauge pressure measurement using a single fluid-filled catheter can be characterized as a second-order linear system (Lambermont et al., 1998; Romagnoli et al., 2011) The connection of two fluid-filled lumens with a single sensor (differential) is more complicated, as there are two different transfer functions (one for each of the lumens) which are dependent due to the joint sensor. However, direct measurement of differential pressure using a fluid-filled double-lumen catheter is exposed to high distortions that originate from common mode pressure effects (CMP) (Rotman et al., 2014). CMP is the line pressure that is common to both ports of the differential pressure sensor, is not measured directly by the differential pressure sensors, but can superimpose any dynamics on the differential measurement, particularly for low pressure ranges, e.g. 0–2.5 kPa. Our method (Rotman et al., 2014) included an algorithm for

Abbreviations: CMP, common mode pressure; FSI, fluid–structure interaction; CMP_{p2p}, common mode pressure effect peak-to-peak (of P_{cmp}) [mmHg]; Err_{cmp}, common mode pressure estimation error [mmHg]; P_{cmp}, common mode pressure [mmHg]; P_d, true pressure drop [mmHg]; P_d *estimated*, corrected pressure drop [mmHg]; P_d *gauge*, reference pressure drop from two gauge pressure sensors [mmHg]; P_{dm}, measured pressure drop [mmHg]; P_d *fsi*, FSI prediction pressure drop [mmHg]; P_g, gauge pressure [mmHg]

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minimizing CMP distortions from the differential pressure measurement, and therefore maintaining the high accuracy of the differential sensor (see also Appendix 1).

Our previous study (Rotman et al., 2014), dealt with a model of a rigid tube. The objective of the present study was to test and validate the CMP correction method in-vitro in compliant tubes, having physiological bio-mechanical properties.

2. Materials and methods

2.1. Experimental setup

The experimental closed flow loop in-vitro system, which full details are given in Rotman et al. (2014) is depicted in Fig. 1A. The following modifications to the system were performed: compliant tube models, which served as mock arteries, were connected to the flow loop, and placed inside the open water bath. The fluid was distilled water at 23 ± 0.2 °C. Gauge pressure was 3 cm upstream and 3 cm downstream of the compliant tube.

In order to keep the uniformity of the catheter's blocking effect along the entire length of the tube models, a 1.7 mm diameter fishing wire extension was connected to the catheter tip to artificially extend the catheter all the way through the models (Fig. 1B).

Images of the compliant tube were captured at a frame rate of 10 fps using a digital camera (uEye, IDS Imaging). The images were analyzed in real time to calculate the tube inner wall diameter using an edge detection algorithm with sub-pixel accuracy (vision and motion toolkit, LabView, National Instruments Corporation, Austin Texas).

2.2. Experimental plan

The pressure drop correction method was applied on two tube models with varying elasticity (Low-Elasticity (LE), and High-Elasticity (HE)) to cover the physiological range of human coronary arteries (distensibility of ~0–7%, see Eq. (1)) (Shimazu et al., 1986).

$$\Delta D[\%] = \frac{D_{\max} - D_{\min}}{D_{min}} \times 100 \quad (1)$$

where ΔD is the radial distensibility [%], D_{max} is the maximal inner diameter [m], and D_{min} is the minimal inner diameter [m] over a cardiac cycle.

Since arterial stenoses are of great importance in coronary arteries, a third compliant tube (Medium-Elasticity) with a 70% area stenosis was modeled (MEst) (Fig. 2A). The MEst model was set at 4 cm long ($l_{stenosis}$) in this scaled-up coronary model to allow pressure drop measurements inside the stenosis (over 3 cm). The materials used to mold the elastic tubes were RTV-615 (GE Bayer Silicones, Bergen op zoom, The Netherlands) for the LE tube, and MED-6020 (Nusil Silicone Technology, CA, USA) for the HE and MEst tubes. The stress–strain characteristics of these materials are depicted in Fig. 2B, as acquired using stretch testing on 80 mm long and 4.3 mm diameter rods at a rate of 100 mm/min.

The tube models were loaded each in the flow loop. In each case the CMP distortion was first identified and fitted when the catheter side-holes were placed in a no-flow region (Rotman et al., 2014). Then, the catheter side-holes were placed inside the compliant tube to validate the restored pressure drop in a flow region.

In each of the tube models, pressure drop was measured with the fluid-filled double lumen catheter over 3 cm in three locations P_1–P_3 (Fig. 2A).

The flow conditions were set to match a hyperemic flow in a 3 mm coronary artery (Konala et al., 2011) using dimensional analysis (peak $Re = \sim 800$): peak/average flow of 195/62 ml/min at a pulsation rate of 1 Hz in the HE model, allowing the flow to adjust freely when the rest of the models were loaded The resistance

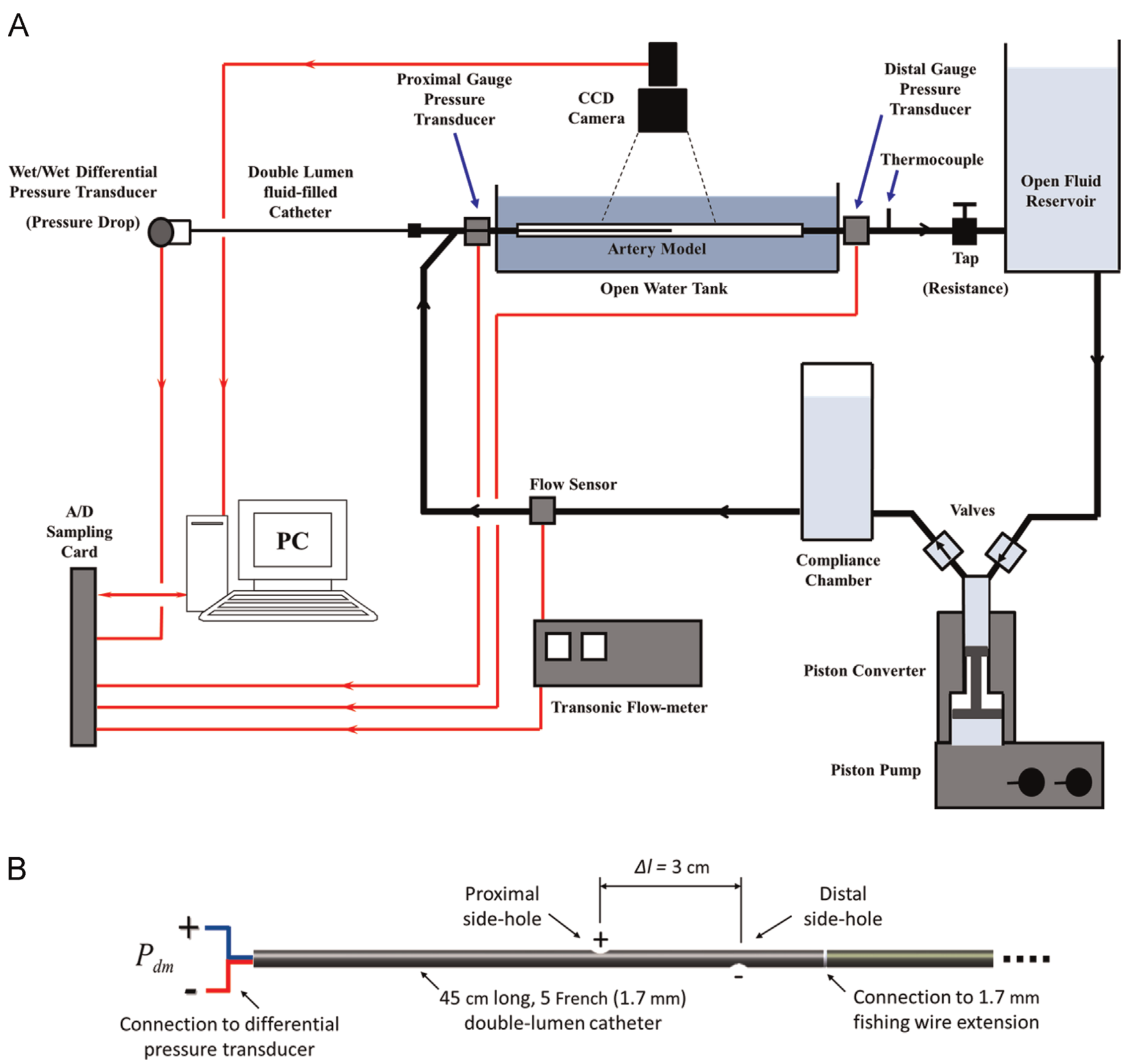

Fig. 1. (A) Illustration of the experimental flow loop. (B) Illustration of the double-lumen catheter setup.