Introducing a Hardware-in-the-Loop Simulation of the Cardiovascular System

Ali Alazmani, David G. Keeling, Peter G. Walker, S. Khawar Abbas, Osama Jaber, Kevin Watterson, and Martin C. Levesley

Abstract—Advances in direct mechanical ventricular actuation devices have been limited by the inability to test the whole device interaction in-vitro. In this study, we introduce a novel technique to produce a realistic, multimodality cardiovascular simulator to mimic the activity of a beating heart. To achieve the mechanical representation of the heart, each ventricle was defined by a real-time modifiable semicircular pattern of post-buckled spring steel strips with adjustable boundary attachments. The mechanical properties of these strips such as stiffness, length, width and boundary conditions approximated the local and global biomechanical properties of the native heart. This physical heart model interfaced with a mathematical model of the cardiovascular system based on hardware-in-the-loop simulation. In-vitro experiments were carried out in an attempt to investigate into the effect that the DMVA system has on PV loop, cardiac output, and overall hemodynamics under different physiological conditions. By employing this in-vitro setting, assist devices can be physically applied to the circulatory models and assessed before animal or clinical trials are conducted. This will significantly aid device behavioural understanding, development time and cost during device’s prototyping.

Index Terms— Hardware-in-the-Loop, Circulatory Model, Direct Mechanical Ventricular Actuation, In-vitro Testing

I. INTRODUCTION

Currently available mechanically driven therapies of end-stage heart disease employ a variety of blood contacting devices to assist the failing heart. However, the potential adoption of these devices in a clinical practice is somewhat limited and frequently ineffective due to cardiopulmonary bypass, immune-system rejection, and complications associated with blood flow over a non-biologic surface, e.g. thromboembolic events, hemolysis and infection [1]. These limitations has caused new interests towards developing direct mechanical ventricular actuation (DMVA) systems which negate blood stream contact by providing a synchronised systolic and diastolic compressive force to the epicardial surface of the ventricle(s) [2]. This compression has been gained using early experiences with patients skeletal muscle [3-5], pneumatic/hydraulic pressure pads [6-8], nano-materials [9, 10] and electromagnetic motors [11].

In the development of DMVA devices, to comprehend the effect of device on myocardial perfusion and overall hemodynamic parameters, it is imperative to utilise in-vitro cardiovascular (CV) circulatory models. According to the specific needs of applications, these models can be based on numerical, physical or hybrid structures.

A. Numerical Models

The numerical simulation of CV systems are being studied with different complexity, modelling assumptions and objectives [12]. Available models are invaluable tools in the early development of cardiac assist devices, however and in contrast to blood contacting assist devices, the physical interaction of DMVAs with their workspace is likely to depend on physical features which are particularly difficult to model in software level (e.g. nonlinear compliance or friction) or yet to be understood. These necessitate use of physical simulation of device’s interaction in the hardware level.

B. Excised Heart Models

Since Langendorff first described the perfusion of an isolated mammalian heart in 1895 [13], his work has been widely used to evaluate the effects of DMVA on physiology and mechanics of cardiac function [14-16]. This technique allows the examination of cardiac functions and overall hemodynamics without the systemic response, neuronal and hormonal complications of an intact model [17, 18]. Thus the use of excised hearts for the development and understanding of DMVA has been invaluable, however, its effects on the closed-loop CV system were not considered and the procedures remained unrepeatable and costly in terms of animal lives, time and resources.

C. Mock Models

A common approach to mimic the physiologically-equivalent CV system has been to produce mechanical mock circulatory systems. Recent attempts have been able to physiologically mimic the atrium, ventricle, and vascular components, through the controlled use of pneumatic or hydraulic drives for flow, along with air tight integrated
chambers to produce vascular resistance and compliance [19]. Although mock circulation loops provide a suitable test platform to evaluate cardiac assist devices [20, 21] and their controllers performance [22, 23], it has difficulty to produce complex nonlinear CV functions and implies high cost and compound mechanisms [24]. By focusing our attention on DMVA, it has not been shown how the mechanical interaction of these systems directly affects the complete mock circulatory system.

D. Hardware-in-the-Loop (HiL) Models

The descriptions of above circulatory models can be optimised by synthesising the features of numerical functions and software with complementary characteristic of physical models. The structure of a physical section of this model is determined by the need of applications. Kozarski et al. introduced an electro-hydraulic interface to create a hybrid CV loop to permit the testing of an intra-aortic balloon pump [25, 26]. Another application is described by Hanson et al., whereby HiL simulation is used as a means of in-vitro testing of a prototyped DMVA device [27], however, their model’s physical section represents an axial slice through the ventricle (two-dimensional model) and it is unable to trial the overall performance of the assist device. This led us to design and manufacture a unique in-vitro testing environment which combines a realistic numerical model of the heart and CV system with a controllable physical heart simulator. By employing this in-vitro setting, DMVA devices can be physically applied to the CV models and assessed before animal or clinical trials are conducted. This will significantly aid device behavioural understanding, development time and cost during device’s prototyping.

II. DEVELOPMENT OF THE HiL SIMULATOR

Our goal was to design a realistic, reliable, reconfigurable HiL model of the complete closed-loop CV system by which DMVA devices could be investigated. In the initial phase of development, a set of design criteria was established to address the limitations of previous models:

- The HiL model must replicate physically and hemodynamically different patient groups, illnesses and even animal complete CV models.
- An appropriate numerical model must be chosen which feature the hemodynamically closed-loop CV system. This produces essential shared variables, between physical and virtual part of the CV model, to assess the effect of applying assistance.
- A physical interface is required to allow the DMVA to be attached as it would in the body. Its mechanical characteristics (e.g. motion, volumes and compliance) and functional parameters (e.g. heart contractility, heart rate) must be adjustable and controllable to enable effective transition between anatomical and physiological conditions dictating from the numerical model.
- The interface and testing section must be isolated in a fully (temperature and humidity) controlled environment for testing of prototyped DMVA device under more realistic conditions.
- The rig must operate as a stand-alone system and run reliably for prolonged time periods.
- The rig also needs to facilitate data logging and remote monitoring equipment to quantitatively assess the performance of the model and the assist device in a realistic workspace condition.

This phase used the criteria above to manufacture an in-vitro testing environment as shown in Fig. 1. The result system is explained in details in the following sections:

A. Software Section (Numerical Model)

Rapid advancement in computational power has allowed the creation of evermore complex simulation models of the CV system, however, they cannot be utilised for representing the entire CV system. Among them, several lumped parameter models have been developed with hemodynamically closed-loop characteristics [28-31]. The blood flow models created by Hanson et al. [32], based on an electrical network analogy, is particularly of our interest because of its relatively low computational cost and straightforward ‘physical’ interpretation which is appropriated for the requirements of the DMVA device’s
applications. The numerical model created consists of six blood storing compartments; both atria and ventricles, along with the aorta and pulmonary artery. Each compartment is modelled separately, allowing localised control over the heart and the development of specific cardiac diseases and conditions. The universal equation used for modelling the pressure, \( P \), of each compartment in terms of flow rate, \( \Phi \), is given by

\[
P = \Phi R + L \frac{\Phi}{dt} + \frac{1}{C} \left( \int \Phi dt - V_o \right) + P_{\text{Additional}}
\]

The electrical analogy terms enable the compartment’s pressure, \( P \), to be defined in terms of its impedance, \( Z \), which encompasses resistance to flow, \( R \), vessel compliance (capacitance), \( C \), and flow inertia (inductance), \( L \). In addition, contributory pressures are localised to some compartments; such as active systolic pressures in the myocardial regions (atria and ventricles), non-linear pressures resulting from compliance curves, and finally specific to the ventricles, assistive pressure from a DMVA device if configured. Although this equation describes vessel flow throughout the CV system, some segments of the circulatory loop, such as the vascular beds are described by only resistive terms, since flow in the capillaries is approximately steady-state and their diminished vessel diameter negates the significance of the inertia.

B. Hardware Section (Physical Heart Model)

In the application of DMVA, physical interaction of assist device with its workspace (mainly ventricles) is essential. Therefore, an interface is required to communicate between the physical assist device and the numerical part of the CV model. When combined, the system can mimic the physical and physiological characteristics of the complete CV system. In our proposed HiL model, this interface takes the form of a beating heart through use of an active compliance actuation system. Considering the concept of a post-buckled beam, three main parts are required to create necessary components for each ventricular motion: (i) a back-drivable linear motor, (ii) compliant members and (iii) boundary attachments. As shown in Fig. 2, each linear motor was utilised to drive a series of thin strips arrange in a semi-circular pattern with initial deformations. This forms a flexible compliance geometry between two adjustable boundary attachments. A fixed frame was also added as a reference body to which the linear motor and the bottom support are fixed. The other support was attached to the linear motor shaft. The variation in the global and local mechanical properties of each ventricle were implemented by tuning the controller’s parameters and changing physical properties of each post-buckled beam (e.g. boundary attachments, beams’ mechanical properties) respectively. This allowed for wide range of adjustments to create more realistic interface. Experiments were conducted to calibrate the physical geometry of this model against a native baseline heart using image intensity and active contour algorithms.

The post-buckled compliant structure was fabricated from cold rolled, oil hardened, tempered spring steel with 0.25mm thickness. This complies with the requirements of BS-1449 for steel trips where severe bending, flexibility and strength is involved. Moreover, the amount of compliance it generates found to be comparable to the native heart model. The compliant strip boundary attachments were manufactured using stereo-lithography technology, to replicate a clamped or hinged boundary condition on each end of the strip and to eliminate radial force on the LinMot® shaft. A layer of Delrin® (DuPont™, US) was added to the fixed frame to further decrease wear and friction, and to increase the load and speed capacity of the system. In contrast to a fluid sac heart model, using these components assist us to easily reconfigure and adjust the physical section to produce evermore complex geometry.

C. Graphical User Interface (GUI)

As illustrated in Fig. 3, the GUI is developed using National Instrument’s LabVIEW™ 2009 graphical programming environment in a Windows™ PC. All physical and physiological factors of the CV system as well as DMVA device’s control parameters are accessible and tuneable for the operator through this interface. It also facilitates data logging for post-processing of the
performance of the model and the assist device. To help satisfy one of our main objectives, the blood flow model of this can be automatically fit to represent real physiological data through the use of a nonlinear least-squares parameter estimation method implemented as a state within the numerical code. This means the heart simulator can accurately reflect the hemodynamics of most patient groups and in-vivo models, helping improve our understanding of the device’s potential effects.

D. Interfacing between Hardware and Software

As shown in Fig. 4, the interfacing between physical heart model and numerical model, and also between the CV simulator and DMVA device are the key factors in this HiL simulation.

The exchange of shared variables between numerical and physical models of CV system can be achieved using pressure-volume and voltage-current respectively for each ventricle. A real-time numerical CV model computes the corresponding local equations for the pre-set cardiac conditions and transmits stimulus information to the physical model. In response, left and right ventricular pressures and volumes are calculated using the inbuilt sensors of the actuators and mathematical prediction of the buckled strip curvature. This dynamic response is transmitted back into numerical model for next circulatory calculations.

Second feedback loop is also used to assess the DMVA assistance to the CV system. Conformable pressure transducers (Pressure Profile System, Inc.) are located at equal intervals around the mechanical heart to provide assistance (compression) data. Investigation into the effect of non-uniform and uniform DMVA demonstrated that 40% and 100%, respectively, of the device compression pressure transfers to the ventricular chambers [14, 15]. This reflects the importance of these pressure transducer for different type of assistance which was not considered in previous HiL study [27]. Keeling et al. has reported how frictional effects inherent to the DMVA device’s operation affect localized surface pressure [33]. Adapting from these studies, type of the DMVA and its position relative to the physical heart is configurable in our proposed HiL simulation. Within the model, this data is converted to an assistive pressure for each ventricle, and the subsequent effect on the blood flow is calculated in real time and causes associated changes in the motion of the physical heart. When combined with assistance pressure reading from pressure transducers, it was allowed the real-time analysis of ventricular pressure-volume loops.

E. Control and Data Acquisition

Because each region of the heart was modelled separately, the heart can be controlled locally, and implementation of specific cardiac conditions and diseases are evermore accessible. The enhancement of this testing environment was achieved by using the LabVIEW™ 2009 and CompactRIO real-time controller. The CompactRIO is used to control the mechanical heart, run the numerical simulation, and send data via transmission control protocol (TCP) to the Windows™ PC host for display and data logging. This real-time controller executes two parallel loops: a high-priority control loop for the blood-flow model and a low-priority communication loop that sends and receives queued TCP data to and from the Windows™ host. The high-priority blood flow model loop runs at 500Hz and converts the two ventricular pressures-volumes to a calibrated currents and positional voltage that is sent to the field-programmable gate array (FPGA) I/O for each of the linear actuators to follow. The FPGA is compiled to deal with all the I/O from the CompactRIO as well as provide proportional-integral (PI) control of a heater, which is used

Fig. 4. Schematic figure of the test rig: (a) temperature controlled test chamber, (b) real-time controllers, and (c) graphical user interface. The variables exchange at interface are pressures-volumes (computed in the numerical model) and voltages-currents (acquired from the physical heart model). Note that the DMVA device and its interface are not shown in this figure.
to keep the physical heart enclosure at a constant 37°C (body temperature). As a result, the HIL simulator could operate as a stand-alone system and run reliably for prolonged time periods. This rugged platform enables our team to conduct in-vitro testing of a novel DMVA assist device, which would not have been possible on a traditional computer.

### III. PRELIMINARY EXPERIMENTS

Preliminary in-vitro experiments were carried out in an attempt to investigate into the effect that the DMVA system similar to [33], has on PV loop, cardiac output, and overall hemodynamics under different physiological conditions. The hemodynamic parameters, collected from an ovine at 87bpm, were used to tune the numerical CV model ran on the proposed HIL simulation. During all experiments, the compressive pressure profile of the DMVA system was measured beneath the device and used to mimic the effect of DMVA on the physical heart model, as shown in Fig. 5. To investigate the effect of contractility on device performance the stiffness of the both the LV and RV were decreased by ~60%. Ventricular PV data for this case with and without assistance was gathered. The effect of afterload on device performance was conducted by decreasing total systemic peripheral resistance, total pulmonary bed resistance, pulmonary artery resistance, and Aortic resistance to 1, 0.1, 0.1 and 0.025 mmHg/ml/s respectively.

For all experiments, ventricular stroke-volume increases with the application of DMVA system, due to the sudden increased ventricular pressure. However, pre-load volume reduces as a consequence of the increased ventricular ejection, thus reducing the augmenting effects of device assistance on stroke volume, causing the PV loop to shift leftwards. Contractility and afterload cause varying the amount of this effect. Fig. 6 depicts that deterioration in cardiac contractility has a positive effect on the assistive performance of the DMVA. The PV plots also show that bi-ventricular assistance has a greater effect on the RV than the LV, causing a greater degree of emptying. By halving vascular resistance this helped both the healthy and weakened heart to pump blood. Device performance also benefitted from the decreased afterload, improving assistive benefit by 0.7 and 2.1 l/min for the healthy and weakened heart when compared to normal vascular resistance. Reduced resistance causes a spike in early systole, which is pronounced during assistance. This is presumably due to the decreased that the heart is pumping against. Decreased vascular resistance reduces mean aortic and PA pressures, but increases pulse pressure. This is due to the increased throughput along the same compliance vessels. Because of

![Fig. 5. Compressive pressure profile measured beneath the DMVA.](image)

### IV. DISCUSSION

The design iterations of prototyped DMVA devices are often precluded by inability of numerical methods to mimic the behaviour of device’s physical interaction or the high cost of animal trials. Consequently, new methods and procedures are welcomed to fully simulate the CV model. This will help reducing the alliance on animal studies and shorten a research and development cycle when designing a new method for cardiac assist devices. In this study, we presented our solution to advance DMVA devices before introducing them into human body or any other clinical environments. The in-vitro reproduction of the complete closed loop CV model has been accomplished through a use of hardware-in-the-loop simulation technique. The dynamics information of patient’s heart in various states of health were computed using a numerical section. Using this information, the physical model utilised the anatomically accurate heart model and will be allowing advanced evaluation, design and testing of the whole DMVA device to be possible (which will be reported in our upcoming papers). DMVA devices can now be implanted, operated and evaluated in a mechanically and anatomically accurate environment. This real-time testing environment not only negates the initial need for animal testing, also it can produce medical conditions that are not easily accessible using animal models. The cardiologist can witness and proof the performance of the assist device to maximise understanding of the device’s interaction with the heart in a more realistic environment.

The classic post buckled beam now appears to offer a great interest to define a new naturally compliant bio-simulator. This modular, reconfigurable structure has been investigated as a universal, robust alternative to a fluid sac
which has fixed body structure and functions. Note that the unique design of the physical heart model makes it possible to alter the baseline dimensions by only changing the boundary attachments, mechanical properties of the compliant strip, linear motors’ angle and also using the silicone skin cover. Benefits include the ability of repeatable test, prolonged trials, considerable control over experimental conditions and the effects of inputs into this cardiac model.

REFERENCES


