# The Role of Simulation for Preoperative Planning in Patients Requiring Mechanical Circulatory Support

Massimo Capoccia[1,2], Sanjeet Avtaar Singh[3], Claudio De Lazzari[4,5]

[1]Department of Cardiac Surgery, University Hospitals of Leicester NHS Trust, UK
[2]Department of Biomedical Engineering, University of Strathclyde, Glasgow, UK
[3]Golden Jubilee National Hospital, Clydebank, Glasgow, UK
[4]National Research Council, Institute of Clinical Physiology UOS of Rome, Italy
[5]National Institute for Cardiovascular Research (I.N.R.C.), Bologna, Italy

*Abstract* – **Mathematical modelling and simulation have the potential to become clinically applicable tools for detailed evaluation of the cardiovascular system and clinical decision-making to guide therapeutic intervention. This approach has a great potential for application in heart failure where the impact of left ventricular assist devices (LVADs) has played a significant role as a bridge to transplant and more recently as a long-term solution for non eligible candidates. We sought to investigate the value of simulation in the context of a heart failure patient with a view to predict or guide further management. CARDIOSIM© software was used to find out the most appropriate course of action to apply in this clinical setting. The outcome of the simulation was compared with the decision previously made in the clinical environment. Although there was an encouraging agreement, other factors play a role in the clinical decision process. Patient-specific modelling may add a more quantitative approach but requires willingness to accept a different way of thinking.**

*Keywords* – **Ventricular assist device, Heart failure, Simulation, CARDIOSIM©, Modelling.**

## I. Introduction

Mathematical modelling and simulation have the potential to become clinically applicable tools for detailed evaluation of the cardiovascular system and clinical decision-making to guide therapeutic intervention. Models based on pressure-volume relationship and zero-dimensional representation of the cardiovascular system may be a suitable choice in view of their simplicity and versatility. Although they provide less detailed predictions of pressure and flow waveforms, these models have shown great flexibility in simulating the haemodynamics of a wide range of cardiovascular conditions and therapeutic interventions maintaining at the same time the ability to be run in real time on desktop, laptop or mobile devices. This approach has a great potential for application in heart failure where the impact of ventricular assist devices (VADs) has played a significant role as a bridge to transplant and more recently as a long-term solution for non eligible candidates. Continuous flow rotary blood pumps are currently the most popular devices in view of their smaller size, increased reliability and higher durability compared to pulsatile flow devices. The trend towards their use is increasing. Simulations for continuous flow VADs are usually performed under steady flow conditions where the inlet boundary is set to a specific steady velocity profile while the outlet boundary is fixed at a steady pressure. The study of the interaction between LVADs and the whole cardiovascular system with a 3-D CFD model is highly demanding although limited parts of the assisted circulation have been developed with this method. A more simplified approach consists of using one-dimensional (1-D) or lumped parameter (0-D) models where space dependence is either confined to the axial coordinate (1-D) or addressed by splitting the cardiovascular system in compartments (0-D). The ventricular behaviour can be modelled according to the time-varying elastance theory, although its application to a mechanically supported left ventricle has come into question more recently. In contrast, a linear model has been proven sufficiently accurate and adequate for realistic simulations of the instantaneous pressure-volume relation. To address the short-comings of the original time-varying elastance theory, a non linear time-varying lumped parameter model of the cardiovascular system can be modified to include the interventricular septum and a rotodynamic, continuous flow LVAD where the end-systolic pressure-volume relationship (ESPVR) is a unimodal function that takes into account the descending limb of the Frank-Starling curve. We sought to investigate the value of simulation in the context of a heart failure patient with a view to predict or guide further management. CARDIOSIM© software was used for this purpose.

## II. Material and methods

### A. Heart model background

CARDIOSIM© is a software package developed by the Cardiovascular Modelling LAB based in the Institute of Clinical Physiology, CNR, Rome, Italy. This is a numerical simulator of the cardiovascular system based on lumped parameter models. The software is interactive and can emulate physiological and pathological conditions for educational purposes and clinical decision-making in a controlled environment. CARDIOSIM© is versatile and based on a modular approach with an updatable library of numerical models of different sections of the cardiovascular system, which can be assembled according to the need of the simulation. The software is particularly suitable to study the


H. Eskola et al. (eds.), *EMBEC & NBC 2017*,
IFMBE Proceedings 65,
DOI: 10.1007/978-981-10-5122-7_140

interactions with pulsatile or continuous flow ventricular assist devices. When the native atrial and ventricular behaviour are modelled, the numerical simulator allows choosing between two different modules based on the time-varying elastance theory. For the purpose of our simulations, we have considered the module where ventricular elastance is described as a function of the characteristic elastance in systole, $E_{lv,s}$, and in diastole, $E_{lv,d}$, and an activation function $\bar{e}_{lv}(t)$ as follows:

$$e_{lv}(t) = E_{lv,d} + \frac{E_{lv,s} - E_{lv,d}}{2} \cdot \bar{e}_{lv}(t)$$

$$e_{rv}(t) = E_{rv,d} + \frac{E_{rv,s} - E_{rv,d}}{2} \cdot \bar{e}_{rv}(t)$$

where

$$\bar{e}_{lv}(t)=\bar{e}_{rv}(t)=\begin{cases} 1 - \cos\left(\frac{t}{T_T}\pi\right) & 0 \le t \le T_T \\ 1 + \cos\left(\frac{t-T_T}{T_{TE}-T_T}\pi\right) & T_T < t \le T_{TE} \\ 0 & T_{TE} < t \le T \end{cases}$$

where $T$ is the duration of the ECG signal (heart period), $T_{TE}$ is the end of ventricular systole and $T_T$ is the T-wave peak time. Ventricular pressure is obtained from the instantaneous ventricular volume and elastance as follows:

$$\begin{aligned} P_{lv}(t) &= P_{lv,0} + e_{lv}(t)[V^{\star}{}_{lv}(t) - V_{lv,0}] \\ P_{rv}(t) &= P_{rv,0} + e_{rv}(t)[V^{\star}{}_{rv}(t) - V_{rv,0}] \end{aligned} \quad (1)$$

wher e $V^{\star}{}_{lv}(t)$ and $V^{\star}{}_{rv}(t)$ is the instantaneous left and right free wall ventricular volume; $V_{lv,0}$ and $V_{rv,0}$ is the resting left and right ventr icular volume.

The same approach is applicable to the atrial chambers, the interventricular (IVS) and interatrial (IAS) septum.

The IVS is the key factor where the properties of one ventricle are a function of the contra-lateral one. When the time-varying elastance theory is applied, we obtain:

$$e_{sp}(t) = E_{sp,d} + \frac{E_{sp,s} - E_{sp,d}}{2} \cdot \bar{e}_{sp}(t)$$

where $E_{sp,d}$ is the IVS diastolic elastance, $E_{sp,s}$ is the IVS systolic elastance and $\bar{e}_{sp}(t)$ is the activation function as follows:

$$\bar{e}_{sp}(t) = \begin{cases} 1 - \cos\left(\frac{t}{T_R}\pi\right) & 0 \le t \le T_R \\ 1 + \cos\left(\frac{t-T_R}{T_{TE}-T_R}\pi\right) & T_R < t \le T_{TE} \\ 0 & T_{TE} < t \le T \end{cases}$$

where $T_{TE}$ is the end of ventricular systole and $T_R$ is the R-wave peak time.

When $P_{lv}(t) > P_{rv}(t)$, the instantaneous interventricular septal volume $V_{sp}(t)$ becomes the volume of shift of the interventricular septum from the neutral position towards the right ventricle as follows:

$$V_{sp}(t) = \frac{P_{lv}(t) - P_{rv}(t)}{e_{sp}(t)} \quad (2)$$

When $P_{rv}(t) > P_{lv}(t)$, a leftwards septal shift occurs with an instantaneous interventricular septal volume as follows:

$$V_{sp}(t) = \frac{P_{rv}(t) - P_{lv}(t)}{e_{sp}(t)} \quad (3)$$

The instantaneous ventricular volume is obtained as follows:

$$\begin{cases} V_{lv}(t) = V^{\star}{}_{lv}(t) + V_{sp}(t) \\ V_{rv}(t) = V^{\star}{}_{rv}(t) - V_{sp}(t) \end{cases} \quad (4)$$

From equations (1), we obtain:

$$\begin{aligned} V^{\star}_{lv}(t) &= \frac{P_{lv}(t) - P_{lv,0}}{e_{lv}(t)} + V_{lv,0} \\ V^{\star}_{rv}(t) &= \frac{P_{rv}(t) - P_{rv,0}}{e_{rv}(t)} + V_{rv,0} \end{aligned} \quad (5)$$

Inserting equations (5), (2) and (3) in (4), we obtain:

$$\begin{cases} V_{lv}(t) = \frac{P_{lv}(t) - P_{lv,0}}{e_{lv}(t)} + V_{lv,0} + \frac{P_{lv}(t) - P_{rv}(t)}{e_{sp}(t)} \\ V_{rv}(t) = \frac{P_{rv}(t) - P_{rv,0}}{e_{rv}(t)} + V_{rv,0} + \frac{P_{rv}(t) - P_{lv}(t)}{e_{sp}(t)} \end{cases} \quad (6)$$

Rearranging equations (6), the instantaneous ventricular pressure becomes:

$$\begin{cases} P_{lv}(t) = \frac{e_{sp}(t) \cdot e_{lv}(t)}{e_{sp}(t) + e_{lv}(t)}\left[V_{lv}(t) - V_{lv,0}\right] + \\ + \frac{e_{lv}(t)}{e_{sp}(t) + e_{lv}(t)} \cdot P_{rv}(t) + \frac{e_{sp}(t)}{e_{sp}(t) + e_{lv}(t)} \cdot P_{lv,0} \\ P_{rv}(t) = \frac{e_{sp}(t) \cdot e_{rv}(t)}{e_{sp}(t) + e_{rv}(t)}\left[V_{rv}(t) - V_{rv,0}\right] + \\ + \frac{e_{rv}(t)}{e_{sp}(t) + e_{rv}(t)} \cdot P_{lv}(t) + \frac{e_{sp}(t)}{e_{sp}(t) + e_{rv}(t)} \cdot P_{rv,0} \end{cases} \quad (7)$$

Equations (7) are essential for the model to simulate ventricular interactions with or without mechanical circulatory support.

### *B. Circulatory model background*

The systemic arterial section consists of the aortic, thoracic and abdominal compartment and is modelled with $RLC$ circuits. The whole venous system is modelled with only one $RC$ circuit ($R_{vs}$ , $C_{vs}$, $R_{vs}$ ). The inertial forces have been neglected. The pulmonary arterial section consists of the main pulmonary artery and the intrapulmonary vascular bed, each modelled with a $RLC$ lumped element. The pulmonary arteriole compartment consists of a single

resistance in view of their relative stiffness and the fact that inertial forces are negligible. The pulmonary capillary section is also modelled with a single resistance. Coupling between the ventricles and the circulation is obtained by using ideal valves.

### C. Berlin Heart INCOR Pump model background

Figure 1 shows the electric analogue of the LVAD model (Berlin Heart INCOR Pump) integrated in the software and used for our simulations.

$Q_{vpo}$ and $Q_{vpi}$ are calculated as follows:

$$(P_{lv} + P_t) - P_{vpi} = Q_{vpi} \cdot R_{vpi} + L_{vpi} \frac{dQ_{vpi}}{dt}$$

$$P_{vpo} - (P_{as} + P_t) = Q_{vpo} \cdot R_{vpo} + L_{vpo} \frac{dQ_{vpo}}{dt}$$

$$Q_{vpi} = Q_{vad} + C_{vpi} \frac{dP_{vpi}}{dt}$$

$$Q_{vpo} = -C_{vpo} \frac{dP_{vpo}}{dt} + Q_{vad}$$

where inlet and outlet cannulae are modelled with $RLC$ elements $R_{vpi}, L_{vpi}, C_{vpi}$ and $R_{vpo}, L_{vpo}, C_{vpo}$.

The LVAD flow is described as follows:

$$Q_{vad} = K_{vad,0} + \omega \cdot K_{vad,1} + K_{vad,2} \cdot (P_{vpo} - P_{vpi}) + \\ + K_{vad,3} \cdot \omega \cdot (P_{vpo} - P_{vpi}) + K_{vad,4} \cdot \omega^2 \cdot (P_{vpo} - P_{vpi}) + \\ + K_{vad,5} \cdot \omega \cdot (P_{vpo} - P_{vpi})^2 + K_{vad,6} \cdot \omega^2 \cdot (P_{vpo} - P_{vpi})^2$$

where

$$\omega(t) = A_0 + A_p \cdot \sin(\tfrac{2\pi t}{T} + \varepsilon_0)$$

$A_0$ is the component of the LVAD speed; $A_p$ is the amplitude of the pulsation component; $\varepsilon_0$ is the phase difference between the LVAD pulsation component and the native cardiac timing.

### D. Patient background

Here we consider a 34-year-old female patient who sustained an extensive anterior wall myocardial infarction which was successfully managed but required ongoing full medical treatment in view of residual severe left ventricular systolic dysfunction with EF 27%. Subsequently, further deterioration required dobutamine infusion and closed monitoring. The presence of co-morbidities, particularly high body mass index (BMI), made her an unsuitable transplant candidate while LVAD insertion remained debatable and unlikely to improve her clinically. Haemodynamic data from repeated right heart catheter showed persistent elevated pulmonary artery pressures and resistance with significantly reduced right ventricular stroke work index, which increased the need for right ventricular support following LVAD insertion with potential for prolonged intensive care need and increased risk and complications. The final decision following a multi-disciplinary team meeting (MDT) was to continue with medical management and palliative treatment.

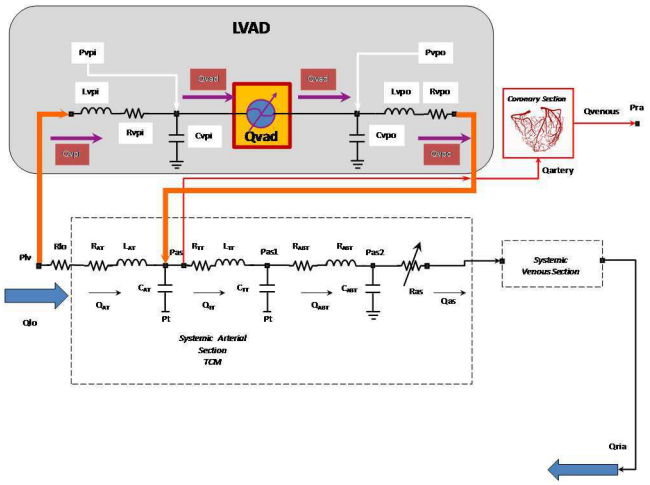


Fig. 1 LVAD and systemic arterial model

Table 1 LVAD model parameters

| Inlet and outlet parameter conduits | | |
|---|---|---|
| **Parameter** | **Value** | **Unit** |
| $C_{vpi}$ [$C_{vpo}$] | 0.1 [0.1] | mmHg·mL$^{-1}$ |
| $R_{vpi}$ [$R_{vpo}$] | 0.01 [0.01] | mmHg·s·mL$^{-1}$ |
| $L_{vpi}$ [$L_{vpo}$] | 1.2e$^{-4}$ [1.2e$^{-4}$] | mmHg·s$^2$·mL$^{-1}$ |
| **LVAD parameters** | | |
| $K_{vad,1}$ | -3.0361e$^{-3}$ | L·min$^{-1}$·rpm$^{-1}$ |
| $K_{vad,2}$ | -1.23045 | L·min$^{-1}$·mmHg$^{-1}$ |
| $K_{vad,3}$ | 5.78974e$^{-4}$ | L·min$^{-1}$·rpm$^{-1}$·mmHg$^{-1}$ |
| $K_{vad,4}$ | -5.8777e$^{-8}$ | L·min$^{-1}$·rpm$^{-2}$·mmHg$^{-1}$ |
| $K_{vad,5}$ | -1.27359e$^{-6}$ | L·min$^{-1}$·rpm$^{-1}$·mmHg$^{-2}$ |
| $K_{vad,6}$ | 2.04834e$^{-10}$ | L·min$^{-1}$·rpm$^{-2}$·mmHg$^{-2}$ |

Table 2 Right heart catheter measurements at different stages

| | RHC (Admission) | RHC (After 4 days) |
|---|---|---|
| RA | 35/17/mean 29 mmHg | 32/22/mean 38 mmHg |
| RV | 61/14/mean 38 mmHg | 71/11/mean 44 mmHg |
| PA | 62/30/mean 42 mmHg | 70/38/mean 50 mmHg |
| PCWP | 36/21/mean 32 mmHg | 35/25/mean 34 mmHg |
| TPG | 10 mmHg | 16 mmHg |
| CO | 2.7 L/min | 2.8 L/min |
| CI | 1.36 L/min/m$^2$ | 1.4 L/min/m$^2$ |
| PVR | 3.7 wu (wood unit) | 5.7 wu (wood unit) |
| RVSWI | 2.41 | 2.3 |

## III. Results

Fig. 2 shows the P-V loop analysis (for the left and right ventricle) without and with LVAD support performed by the software simulator.

Tab 3 shows haemodynamic data from right heart catheter (first three columns). Data simulating the patient conditions are listed in the middle three columns while the last three columns show simulated data obtained after LVAD support.


Fig. 2 Simulation without (right P-V loop in the left upper window) and with LVAD support (left P-V loop in the left upper window)

## IV. DISCUSSION

A simulation-based approach as a potential preoperative strategy in the context of patient-specific modelling may be an additional tool to obtain accurate predictions of the performance of these devices in a clinical setting. Besides, it would be a valuable training opportunity for medical and nursing staff involved with the care of this complex and challenging group of patients.

Interactive software like CARDIOSIM© can emulate physiological and pathological conditions for educational purposes and clinical decision-making in a controlled environment. Examples are the study of the interactions with pulsatile VADs, continuous flow VADs and the intra-aortic balloon pump (IABP).

The simulation outcome for this patient is quite interesting: LVAD assistance does not seem to be beneficial and supports the unanimous clinical decision to decline any further intervention but continue with medical treatment and eventually palliation. Although previous experience, co-morbidities and the risk of potentially fatal complications play a role in clinical decision-making, a simulation setting may well add a more quantitative approach and perhaps reassurance.

## V. CONCLUSION

Patient-specific modelling may become a daily approach for clinical management and training of medical and nursing staff. All of this is available but requires willingness to change attitude towards a different type of approach, which may become the key to successful outcome and the beginning of a complete revolution in patient care. No magic bullets, only different way of thinking.

Table 3
EF ejection fraction; PA pulmonary artery; S/D/M systolic/diastolic/mean; PCWP pulmonary capillary wedge pressure; a/v/m a wave/v wave/mean; CO cardiac output; CI cardiac index; TPG trans-pulmonary gradient; PVR pulmonary vascular resistance; RVSWI right ventricular stroke work index

| | Measured | | | Simulation | | | LVAD (Simulation) | | |
|---|---|---|---|---|---|---|---|---|---|
| EF | 27% | | | 26.5% | | | 30.8% | | |
| PA(S/D/M) (mm Hg) | 62 | 30 | 42 | 42.7 | 38 | 41 | 41.6 | 37 | 39.6 |
| PCWP (a/v/m) (mm Hg) | 36 | 21 | 32 | 32 | 20 | 26.5 | 30.8 | 18 | 24.6 |
| CO (L/min) | 2.7 | | | 2.69 | | | 0.67 | VAD (2.15) | TOT (2.82) |
| CI (L/min/m$^2$) | 1.36 | | | 1.36 | | | 0.34 | | |
| TPG (mm Hg) | 10 | | | 14.5 | | | 15 | | |
| PVR (wood unit) | 3.7 | | | 5.39 | | | 22.38 (15/0.67) | | 5.32 (15./2.82) |
| RVSWI | 2.41 | | | 7.1 | | | 7.2 | | |

## CONFLICT OF INTEREST

The authors declare they have no conflict of interest.

## REFERENCES

1. Doshi D, Burkhoff D (2016) Cardiovascular Simulation of Heart Failure. Pathophysiology and Therapeutics. J Card Fail 22(4):303-311.
2. De Lazzari C, Genuini I, Pisanelli DM, D'Ambrosi A, Fedele F (2014) Interactive simulator for e-Learning environments: a teaching software for health care professionals. BioMedical Engineering OnLine 13:172.
3. Suga H, Sagawa K, Shoukas AA (1973) Load Independence of the Instantaneous Pressure-Volume Ratio of the Canine Left Ventricle and Effects of Epinephrine and Heart Rate on the Ratio. Circ Res 32:314-322.
4. Vandenberghe S, Segers P, Steendijk P, Meyns B, Dion RAE, Antaki JF, Verdonck P (2006) Modelling Ventricular Function during Cardiac Assist: Does Time-Varying Elastance Work? ASAIO J 52:4-8.
5. Korakianitis T, Shi Y (2006) A concentrated parameter model for the human cardiovascular system including heart valve dynamics and atrioventricular interaction. Med Eng & Phys 28:613-628.

Corresponding author:

Massimo Capoccia
University Hospitals of Leicester NHS Trust, Leicester (UK)
Email: capoccia@doctors.org.uk