2021 43rd Annual International Conference of the
IEEE Engineering in Medicine & Biology Society (EMBC)
Oct 31 - Nov 4, 2021. Virtual Conference

2021 43rd Annual International Conference of the IEEE Engineering in Medicine & Biology Society (EMBC) | 978-1-7281-1179-7/21/$31.00 ©2021 IEEE | DOI: 10.1109/EMBC46164.2021.9630113

Identification of an optimal CPR chest compression protocol.

C. Daudre-Vignier[1], M. Laviola[1], A. Das[2], D.G. Bates[2] and J.G. Hardman[1,3]

***Abstract*— In this study, we used a high-fidelity integrated computational model of the respiratory and cardiovascular systems to investigate cardiopulmonary resuscitation (CPR) after cardiac arrest in a virtual healthy subject. For the purpose of this work, a newly developed thoracic model has been integrated to the current model, to study the influence of external chest compressions upon the arrested circulation during CPR. We evaluated the chest compression (CC) parameters, namely, end compression force, compression rate, and duty cycle to optimize the coronary perfusion pressure and the systolic blood pressure, using a genetic algorithm. While the sternal displacement associated with the CC force agreed with the ERC guidelines, the CC rate and duty cycle were respectively higher and lower than the ones recommended by the ERC guidelines. The effect of these CC parameters on cardiac output (CO) were also assessed. The end compression force was the parameter with the largest impact on CO, while the compression rate and duty cycle scarcely influence it.**

***Relevance*— Our results may aid in understanding the underlying pathophysiology of cardiac arrest and help guide research into the refinement of CPR strategies, without sacrificing animals or conducting clinical trials, which are difficult to undertake in crisis scenarios.**

I. Introduction

Cardiac arrest remains a leading cause of death in many countries. Despite the many years of research on CPR and attempts to improve outcomes, survival to hospital discharge remains consistently low. When performing chest compression three main components are considered: depth, rate, and duty cycle. While the European Resuscitation Council (ERC) advance life support guideline [1] advises compression depth of 5-6 cm, a rate of 100-120 compression per minute and a duty cycle of 50% in compression, a majority of studies fail to find an association between these parameters and CPR outcomes [2]. The question therefore arises as to what are the optimal chest compression depth, rate, and duty cycle.

Many human trials have attempted to identify the optimal CPR strategy, however, the ethical constraints, time scale, the presence of confounding variables, the heterogeneity of the population and sample size present major obstacle. Similarly, many animal models fail to summarize the severity of human cardiac arrest due to interspecies physiological discrepancies and lack of methodical rigor [3]. Computer simulation is a new alternative to the animal and clinical trials because it allows replacing animals and not putting at risk patients in crisis scenarios, plus it allows complete reproducibility of the methods, configuration of individualized patients and mechanistic results.

The present paper describes: i) the relevant components of the current recently updated computational model and cardiopulmonary interactions; ii) the new integrated thoracic model; iii) the comparison between cardiovascular model outputs and literature data and iv) the use of a genetic algorithm (GA) to optimize the chest compression parameters.

II. Method

A. Description of the cardiopulmonary model

The Interdisciplinary Collaboration in Systems Medicine (ICSM) simulation suite, used for this study, is a high-fidelity integrated computational model of the respiratory and cardiovascular systems. The core model has been extensively validated in multiple previous investigations [4, 5]. The respiratory model includes a series deadspace volume, 100 independently configurable alveolar compartments and apneic state. The cardiovascular model consists of 19 compartments (Figure 1).

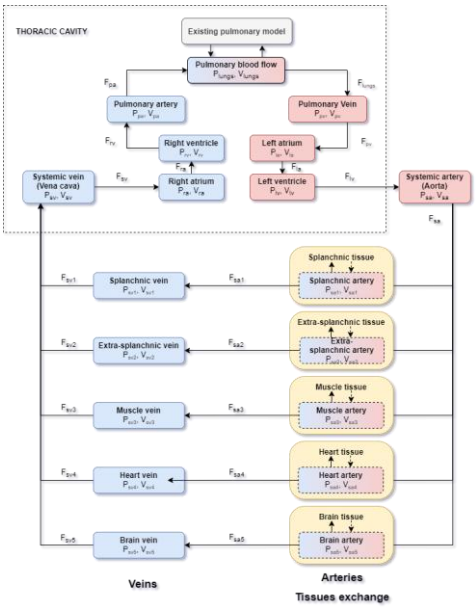

Figure 1. Schematic of updated cardiovascular model.

The equations describing each vascular compartment i have been recently updated by enforcing conservation of mass and balance of forces. The main equations were obtained from Albanese and colleagues' work [6]. Briefly, all the vascular compartments are modelled as a two elements windkessel system with a resistance (R_i), and capacitance (C_i) and in the systemic artery and the pulmonary artery an additional impedance (L_i) was added. The thoracic vein (sv) was also

[1] Academic Unit 3 - Injury, Inflammation and Recovery Science, School of Medicine, University of Nottingham, UK; [2]School of Engineering, University of Warwick, UK; [3]Nottingham University Hospitals NHS Trust, UK. Corresponding author: Clara.Daudre-Vignier@nottingham.ac.uk.

Authorized licensed use limited to: University of Warwick. Downloaded on March 31,2022 at 15:04:05 UTC from IEEE Xplore. Restrictions apply.

updated to enable its collapse. The thoracic vein and the aortic artery resistances were modified to account for the compartments volume. Additionally, a similar partitioning of the tissues was made splanchnic, extra-splanchnic, muscle, brain, and heart (Figure 1). The organ tissues were defined by their oxygen consumption per minute ($V\dot{O}_2$), volume ($Vtiss$), and perfusion ($\dot{Q}$). In addition, the model of ventricular and atrial contractions was recently updated based on Bozkurt work [7].

B. Cardiopulmonary interactions

The model includes the effect of the pulmonary vasoconstriction which results in a 'U' shape change pulmonary vascular resistance (PVR) at around the functional residual capacitance (FRC). The effect of hypoxic vasoconstriction is also modelled by increasing the PVR in response to hypoxia caused by the alveoli collapse.

Additionally, the effect of the pleural pressure has been integrated to the current model [6] and is defined by the following equation:

$$Ppl = P_{alv} - \frac{V_{alv} - V_{u,alv}}{C_{alv}} \tag{1}$$

where V_{alv} is the alveolar volume, P_{alv} is the alveolar pressure, C_{alv} is the alveolar compliance, and $V_{u,alv}$ is the alveolar unstressed volume.

C. Description of the new thoracic model

For the purposes of this study, a new thoracic model has been added to the current computational model and is adapted from Babbs chest wall model [8]. The external chest compression force $F(t)$, applied at the sternum, is expressed as a function of the end compression force (F_{max}), compression rate (CC_{rate}) and the duty cycle ($Duty_{cycle}$) such as:

$$F(t) = \begin{cases} F_{max}\dfrac{1-\cos\left(\pi(t-T1)\right)}{2} & 0 \le t < T1 \\ F_{max}\dfrac{1-\cos\left(\pi\dfrac{(t-T1)}{(T-T1)}\right)}{2} & T1 \le t < T \end{cases} \tag{2}$$

where T is the period of compression and decompression and $T1$ is the compression period ($T1 = T \cdot Duty_{cycle}$).

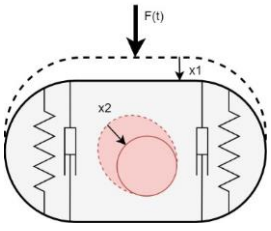

Figure 2. Schematic of the thoracic model. The paired spring and damper represent the elastic behavior of the chest wall under the compression force $F(t)$, including hysteresis. x_1 and x_2 represent the displacements of the sternum and the cardiac chambers under the compression force $F(t)$, respectively. The red and grey area represents the mediastinal tissue.

To relate the external chest compression force $F(t)$ and the sternum displacement (x_1), the thorax is modelled as a simple mechanical spring-damper system (Figure 2). The displacement of the sternum in response to the compression force is given by the following equation:

$$F(t) - kx_1 - \mu\dot{x}_1 = 0 \tag{3}$$

where x_1 is sternal displacement, $\dot{x}_1$ is the velocity of sternal displacement, k is the spring constant, and μ is the damping constant. To relate the sternal displacement (x_1) to the intrathoracic tissue (mediastinum) pressure (P_M), the mediastinum is modelled as an elastic material of Young's modulus of elasticity E.

The rate of change of the cardiac chambers diameters ($\dot{x}_2$) due to the changes in internal blood volume are considered and defined as follows:

$$\dot{x}_2 = \frac{F_{in} - F_{out}}{A_c} \tag{4}$$

where F_{in} and F_{out} are respectively the flows entering and leaving the cardiac chambers, and A_c is the cross-sectional area of cardiac chambers. However, the changes in diameters in the intrathoracic vascular compartments are negligeable ($\dot{x}_2 = 0$). Therefore, the relationship between the mediastinum pressure (P_M), the velocity of sternal displacement ($\dot{x}_1$) and rate of change of the compartment diameters ($\dot{x}_2$) is given by:

$$\frac{P_M}{dt} = \alpha\frac{E(\dot{x}_1 + \dot{x}_2)}{d_0} \tag{5}$$

Where α is a scalar, and d_0 is the combined mediastinal tissues width. The effect of chest compression on the alveolar pressure is also modelled. The resulting pressure on the alveoli (P_{CC}) is related to the sternal displacement (x_1) by the following equation:

$$P_{CC} = \frac{x_1 A_{lungs}}{C_{lungs}} \tag{6}$$

where, A_{lungs} is the cross-section of lung squeezed by sternal compression and C_{lungs} is the combined lung-chest wall compliance. The pressure in each alveolus (pi) is then updated as:

$$pi = \begin{cases} \dfrac{Si(10\cdot vi - 300)^3}{6600} - P_{ext,i} - P_{CC} \ , vi > 0 \\ 0 \ , otherwise \end{cases} \tag{7}$$

$$for\ i = 1, \dots, Ncomp$$

where Si is a scalar that determines the intra-alveolar pressure for a given volume, vi is the volume of the ith alveolar compartment, and $P_{ext,i}$ is the effective net pressure generated by the sum of the effects of factors outside each alveolus that act to distend that alveolus.

The updated alveolar pressure affects the pleural pressure (Ppl) (Equation 1).

In this model, all the cardiovascular compartments enclosed in the thoracic cavity (Figure 1) experience a combination of pleural (Ppl) and mediastinal pressure (P_M), such that:

Authorized licensed use limited to: University of Warwick. Downloaded on March 31,2022 at 15:04:05 UTC from IEEE Xplore. Restrictions apply.

$$\frac{dP_i}{dt} = \frac{F_{in} - F_{out}}{C_i} + \frac{dPpl}{dt} + f_{tp} \cdot \frac{dP_M}{dt} \quad (8)$$

where, $f_{tp} = 0.75$ is the thoracic pump factor which determines the degree to which the ''thoracic pump'' mechanism of CPR is working.

D. Genetic algorithm

In this paper, we used a genetic algorithm method to solve the optimization problem. Briefly, the algorithm: 1) creates a random initial population from the parameter space defined by the boundaries of the input parameters; 2) calculates the individual scores of each member of this population by running the simulation and deriving the cost (8); 3) generates members of the next population through the selecting best individuals of the existing population (elitism), combining the characteristics of two different members (crossover), and making changes to randomly selected individuals (mutation) 4) repeat steps 2)-3) until an optimal solution is found that satisfies a pre-defined criteria, in this case the improvement in cost function between consecutive generations is below 10^{-3} or the number of generations exceed 100.

The aim of the genetic algorithm was to find the sets of chest compression parameters that minimize the cost function defined as:

$$\begin{aligned} &Cost \\ &= w_1 \left| \frac{CPP_{model} - CPP_{desired}}{CPP_{desired}} \right| \\ &+ w_2 \left| \frac{SBP_{model} - SBP_{desired}}{SBP_{desired}} \right| \end{aligned} \quad (8)$$

where, CPP is the coronary perfusion pressure, SBP is the systolic blood pressure, the subscript 'model' defines the simulation output parameters, the subscript 'desired' defines the desired value of these output parameters and $w_1 = w_2 = \frac{1}{2}$ are the weights assigned to each objective. The desired value of CPP and SBP during CPR were defined as those during spontaneous ventilation (Table II).

We selected these outcome parameters (CPP and SBP) because of their association with return of spontaneous circulation. Indeed, CPP is defined as the difference between the aortic diastolic pressure minus the right atria diastolic pressure and gives an indication of the myocardial blood flow during CPR. In an animal study by Maryam et al. [9], targeting a systolic blood pressure of 100 mmHg and CPP>20 mmHg improved survival compared to the guidelines.

Additionally, we added constraints related to the safety and practicality of the chest compressions. The end compression force (F_{max}) was ranged between 0 and 400 N, the chest compression rate (CC_{rate}) between 60 and 150 compressions per minute (cpm), and the duty cycle ($Duty_{cycle}$) between 0.2 and 0.8. Finally, we constrained the genetic algorithm to only give integer values for F_{max} and CC_{rate}, as a results with decimal place would not be relevant.

E. Protocol

A virtual healthy subject was configured as in our previous study [3]. After 10 minutes of spontaneous ventilation (SPV), cardiac arrest (CA) was simulated for 5 minutes by setting the heart rate to 0, effectively forcing the heart to be in constant diastole. Additionally, we activated the apnoea module with obstructed upper airway. During CPR, apnoea was maintained however, the upper airway was no longer obstructed allowing passive ventilation from chest compressions. CPR was simulated for 1 minute following the ERC guidelines ($i.e. F_{max} = 400\ (N)$, $CC_{rate} = 120\ (cpm)$, and $Duty_{cycle} = 0.50$) as well as following the results of the genetic algorithm, i.e. optimized protocol. For comparison of the results, we selected Hyun's work [10] because it is one the most recent work on humans with detailed methodology and cardiovascular data.

The genetic algorithm and model simulations used Matlab version R2019b.v9 (MathWorks Inc., Natick, MA, USA).

III. Results

A. Cardiovascular model outputs

Table I compares the simulated cardiovascular model outputs when CPR is performed following the ERC guidelines *versus* Hyun *et al.*[10] work.

TABLE I. MODEL OUTPUTS VS. LITERATURE HUMAN DATA FROM HYUN *ET AL.* [10] DURING CPR.

Outputs	Unit	*Hyun et al.* data	CPR (ERC)
Peak LV pressure	mmHg	112 ± 37	76.9
Peak aorta pressure	mmHg	105 ± 41	76.9
Peak RA pressure	mmHg	89 ± 27	83.2
End CC LV pressure	mmHg	8 ± 11	-2.6
End CC aorta pressure	mmHg	33 ± 10	38.7
End CC RA pressure	mmHg	8 ± 6	2
CPP	mmHg	25 ± 9	36.7
Stroke volume	ml	25 ± 8	11.6
Ejection fraction	%	34 ± 16	9

LV: left ventricle; RA: right atrium; CC: chest compression.

The model outputs are overall similar to the ones observed in humans [10]. Moreover, when CPR is performed according to the ERC guidelines, CPR only provides 17-27% of the normal cardiac output [10]. Similarly, only 10-30% of the blood flow to the heart and 30%-40% of the normal blood flow to the brain is achieved [11]. Our model provides a maximum cardiac output of 1.39 L/min which is 28% of the normal blood flow. Additionally, 17.7% and 17.7% of the normal blood flow goes to the heart and the brain, respectively.

B. Genetic algorithm

After 17 generations, the genetic algorithm identified the optimal CC parameters to be: $F_{max} = 400\ (N)$, $CC_{rate} = 137\ (cpm)$, and $Duty_{cycle} = 0.28$; the associated maximal sternal displacement was 5 cm. While the sternal displacement agrees with the ERC guidelines, the CC rate and

Authorized licensed use limited to: University of Warwick. Downloaded on March 31,2022 at 15:04:05 UTC from IEEE Xplore. Restrictions apply.

duty cycle are respectively higher and lower than the ones recommended by the ERC guidelines [1].

Table II. shows the model outputs, CPP, SBP and CO at the end of the three different stages: SPV, CA, CPR following the ERC guidelines (ERC) and CPR with the optimised protocol (OPT). CPP, SBP and CO are partially restored after 1 minute for both CPR protocols.

TABLE II. MODEL OUTPUTS DURING SPONTANEOUS VENTILATION (SPV), CARDIAC ARREST (CA) AND CARDIOPULMONARY RESUSCITATION (CPR).

Parameters	Unit	SPV	CA	CPR (ERC)	CPR (OPT)
CPP	mmHg	75	0	36.7	41
SBP	mmHg	114	7.7	76.9	73
CO	L/min	4.98	0	1.39	1.45

CPP: coronary perfusion pressure; SPB: systolic blood pressure; CO: cardiac output; ERC: European Resuscitation Council; OPT: optimised.

Compared to the ERC protocol, the optimised protocol produces values of CPP and CO that are closer to the values produced during normal breathing (Table II).

C. *Effect of external compression force, rate, and duty cycle on cardiac output*

Figure 3 shows the effect of the compression force, compression rate and duty cycle on CO. When a parameter was evaluated, the other two remained constant ($F_{max} = 400\ (N)$, $CC_{rate} = 138\ (cpm)$, and $Duty_{cycle} = 0.28$).

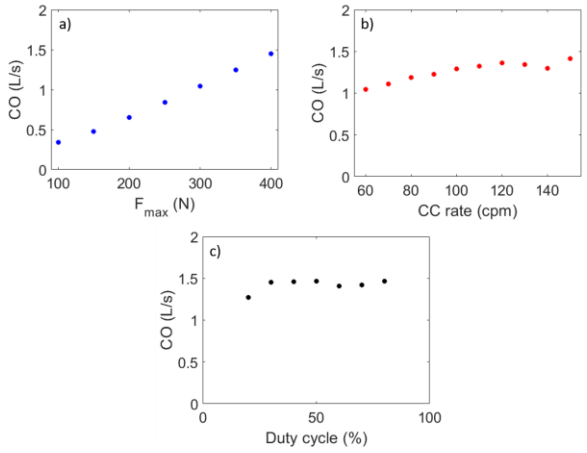

Figure 3. a) cardiac output vs. F_{max} b) cardiac output vs. CC_{rate} c) cardiac output vs. $Duty_{cycle}$.

The model indicates that the CO during CPR is linearly correlated (r^2 =0.99) to the compression force F_{max} and remains fairly constant when the CC_{rate} and $Duty_{cycle}$ change.

IV. CONCLUSION

Integration of a new detailed thoracic model to the current highly integrated cardiopulmonary model has produced a powerful tool to study CPR. The model outputs obtained following the ERC guidelines were similar to the ones observed in the literature data [10, 11], although the stroke volume and the blood flow to the brain achieved were lower.

In this study, the genetic algorithm identified the optimal CC parameters to be: $F_{max} = 400\ (N)$, $CC_{rate} = 137\ (cpm)$, and $Duty_{cycle} = 0.28$ resulting in a sternal displacement of 5 cm. The end compression force (F_{max}) is the parameter with the greatest impact on CO, while the compression rate (CC_{rate}) and duty cycle ($Duty_{cycle}$) scarcely influence CO.

Our simulations did not account for modifications of the tissue's vascular resistances nor the ventilation. Future work will investigate the modelling of the tissue's vascular resistance and ventilation, improving the accuracy of the CPR simulations.

ACKNOWLEDGMENT

This work is supported by National Centre of 3Rs (NC/S001328/1).

REFERENCES

[1] J. Soar, J. P. Nolan, B. W. Böttiger, G. D. Perkins, C. Lott, P. Carli, T. Pellis, C. Sandroni, M. B. Skrifvars, G. B. Smith, K. Sunde, and C. D. Deakin, "European Resuscitation Council Guidelines for Resuscitation 2015: Section 3. Adult advanced life support," *Resuscitation,* vol. 95, pp. 100-47, Oct, 2015.

[2] J. Considine, R. J. Gazmuri, G. D. Perkins, P. J. Kudenchuk, T. M. Olasveengen, C. Vaillancourt, C. Nishiyama, T. Hatanaka, M. E. Mancini, S. P. Chung, R. Escalante-Kanashiro, and P. Morley, "Chest compression components (rate, depth, chest wall recoil and leaning): A scoping review," *Resuscitation*, Sep 16, 2019.

[3] B. K. Fabian-Jessing, M. F. Vallentin, N. Secher, F. B. Hansen, C. Dezfulian, A. Granfeldt, and L. W. Andersen, "Animal models of cardiac arrest: A systematic review of bias and reporting," *Resuscitation,* vol. 125, pp. 16-21, Apr, 2018.

[4] A. Das, M. Haque, M. Chikhani, W. F. Wang, T. Ali, O. Cole, J. G. Hardman, and D. G. Bates, "Development of an Integrated Model of Cardiovascular and Pulmonary Physiology for the Evaluation of Mechanical Ventilation Strategies," *2015 37th Annual International Conference IEEE Engineering in Medicine and Biology Society (EMBC)*, pp. 5319-5322, 2015.

[5] M. Laviola, A. Das, M. Chikhani, D. G. Bates, and J. G. Hardman, "Computer simulation clarifies mechanisms of carbon dioxide clearance during apnoea," *British Journal of Anaesthesia,* vol. 122, no. 3, pp. 395-401, Mar, 2019.

[6] A. Albanese, L. Cheng, M. Ursino, and N. W. Chbat, "An integrated mathematical model of the human cardiopulmonary system: model development," *Am J Physiol Heart Circ Physiol,* vol. 310, no. 7, pp. H899-921, Apr 1, 2016.

[7] S. Bozkurt, "Mathematical modeling of cardiac function to evaluate clinical cases in adults and children," *PloS one,* vol. 14, no. 10, pp. e0224663-e0224663, 2019.

[8] C. F. Babbs, "Design of near-optimal waveforms for chest and abdominal compression and decompression in CPR using computer-simulated evolution," *Resuscitation,* vol. 68, no. 2, pp. 277-93, Feb, 2006.

[9] M. Y. Naim, R. M. Sutton, S. H. Friess, G. Bratinov, U. Bhalala, T. J. Kilbaugh, J. W. Lampe, V. M. Nadkarni, L. B. Becker, and R. A. Berg, "Blood Pressure- and Coronary Perfusion Pressure-Targeted Cardiopulmonary Resuscitation Improves 24-Hour Survival From Ventricular Fibrillation Cardiac Arrest," *Critical care medicine,* vol. 44, no. 11, pp. e1111-e1117, 2016.

[10] H. Kim, S. O. Hwang, C. C. Lee, K. H. Lee, J. Y. Kim, B. S. Yoo, S. H. Lee, J. H. Yoon, K. H. Choe, and A. J. Singer, "Direction of blood flow from the left ventricle during cardiopulmonary resuscitation in humans: its implications for mechanism of blood flow," *Am Heart J,* vol. 156, no. 6, pp. 1222.e1-7, Dec, 2008.

[11] A. R. John, M. Manivannan, and T. V. Ramakrishnan, "Computer-Based CPR Simulation Towards Validation of AHA/ERC Guidelines," *Cardiovasc Eng Technol,* vol. 8, no. 2, pp. 229-235, Jun, 2017.

Authorized licensed use limited to: University of Warwick. Downloaded on March 31,2022 at 15:04:05 UTC from IEEE Xplore. Restrictions apply.