Modelling of respiratory mechanics during cardiopulmonary resuscitation

Carlos Babiera Sancho

LUND
UNIVERSITY

Department of Automatic Control

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Department of Automatic Control
Lund University
Box 118
SE-221 00 LUND
Sweden

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Abstract

Cardiac arrest is a common cause of death. Medical treatment consists of cardiopulmonary resuscitation, comprising chest compressions and ventilation (both of which can be either manually or automatically provided) to recover the normal functioning of the heart.

A good modelling of the process is essential to be able to study different current methods, providing a better understanding of the process itself and, therefore, furthering the developing of new and better cardiopulmonary resuscitation techniques.

This thesis aims to model respiratory mechanics during cardiopulmonary resuscitation. Two different ventilation methods (continuous insufflation of oxygen, CIO, and phase-controlled intermittent intratracheal insufflation of oxygen, PIIO) are simulated and compared in two different simulation models based on the RIC and Mead electrical circuits models.

Although the developed models are simple, they capture some fundamental physiological differences between the ventilation methods, previously observed in porcine studies. To make a reliable quantitative comparison, further data on which to base ventilation and chest compression augmentations of the considered respiratory mechanics models is required.

Acknowledgements

I would like to express my gratitude to Lund University to give me the opportunity of getting involved in a thesis oriented towards biotechnology research, and especially to my supervisors Kristian Soltesz and Harry Pigot for dedicating their valuable time and knowledge.

I also want to mention my family and friends for all the support among years, helping me to achieve this goal that is my master's studies.

Contents

1

Introduction

1.1 Background

The blood provides the cells of the human body with the necessary nutrients and gaseous exchange, and it is pumped by the heart through the arteries in order to go across all the body to reach this purpose. Once all the nutrients have been transported and the gases exchanged, the blood returns across the veins to the heart waiting to obtain gases again from the lungs just before it gets pumped again through the arteries.

As every organ, the heart has its own veins and arteries. Cardiac arrest seizes heart arteries blood flow, also known as coronary perfusion. Cells die without the oxygen provided by respiration and blood circulation, causing irreparable damage to organs. The brain is the most vulnerable one; it only takes 2 or 3 minutes before irreversible damage becomes likely. Thus, in order to avoid necrosis of the brain cells, cardiopulmonary resuscitation (CPR) is applied.

The coronary perfusion is necessary to avoid ischemic damage to the heart, being an essential variable to take into account in CPR techniques and, therefore, in the design of the models. It is difficult to measure coronary perfusion accurately in vivo. However, coronary perfusion is driven by the pressure difference between the diastolic aortic pressure and the right atrium end diastolic pressure, and this pressure difference is readily measurable online in animal models.

More information related to human respiratory system can be found in [Gray and Goss, 1974]], [Weibel et al., 1963], [Zipes et al., 2018].

1.2 Cardiopulmonary resuscitation

CPR is a technique that consists of applying chest compressions, a carbon dioxide ventilation method, and oxygen insufflation. Both parts (chest compressions and air

ventilation) can be provided either manually or automatically.

The chest compressions generate coronary perfusion in every decompression phase by forcing a CPP, avoiding ischemic damage to the heart in order to give it a chance of recovering . Research has shown that a minimum CPP of 15 mmHg is necessary in order to get return of spontaneous circulation (ROSC) upon defibrillation [Paradis et al., 1990].

Ventilation aims to remove carbon dioxide while oxygen insufflation delivers all the oxygen cells need. They both keep organs alive, especially the brain, until the heart can be re-started. From now on, in order to simplify the concepts, both oxygen insufflation and carbon dioxide ventilation will be referred to as ventilation.

Referring to automated mechanic CPR (mCPR), it has been shown there are better outcomes using LUCAS than with manual compressions [Rubertsson and Karlsten, 2005].

LUCAS

LUCAS (first generation pneumatic device, Jolife AB, Lund, Sweden) is a mechanic chest compression-active decompression device. It compresses the thorax and actively brings it back to its natural position by using a suction cup. The LUCAS device can be seen in Figure 1.1.

Figure 1.1 LUCAS device.

The frequency used by LUCAS (gas driven first version) is 100 compressions per minute, which means a compression-decompression cycle of 0.6 seconds.

Boussignac tube

The boussignac tube is an endotracheal tube specially designed for CPR used for supplying oxygen to the lungs of the patient and to ventilate carbon dioxide from the lungs of the patient. It is introduced through the trachea and it consists of two parts: a bigger lumen connects the lungs to the atmosphere, while smaller lumina located in the tube wall supply oxygen into the lungs. A boussignac tube can be seen in Figure 1.2.

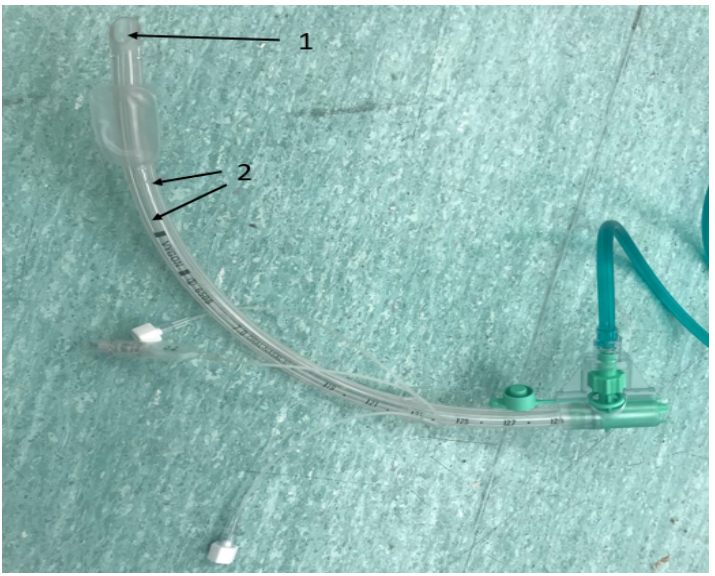

Figure 1.2 A boussignac tube indicating the main lumen (1) and some of the smaller lumina located in the tube wall (2).

1.3 Ventilation methods

In this thesis, two oxygen insufflation methods are considered in simulation: continuous insufflation of oxygen (CIO) and phase-controlled intermittent intratracheal insufflation of oxygen (PIIO). The ventilation methods discussed here are all used in combination with mCPR.

Continuous insufflation of oxygen

The CIO method consists of supplying a continuous oxygen flow to the lungs. By doing this, it has been shown that there are better outcomes, including an increase of CPP and a decrease of the physiological dead space (volume of ventilated air that does not participate in gas exchange during the respiratory process), compared to intermittent positive pressure ventilation (IPPV) [Steen et al., 2004].

Phase-controlled intermittent intratracheal insufflation of oxygen

This method applies an intermittent (squared signal) insufflation pressure. Since all perfusion of the heart takes place during the decompression phase of CPR, the idea is to have the lungs as empty as possible when the decompression begins in order to generate a lower pressure in the thorax during decompression. This entails a decrease of the volume the lungs occupy within the thorax and, therefore, a lower right atrial pressure (RAP) during the decompresion phase. Research has hypothesized that this generates two benefits:

1. It increases the CPP and consequently the coronary perfusion, as the difference between the aortic pressure and the right atrial pressure is bigger.

2. For the same reason, it aids systemic venous return, improving therefore the transportation of oxygen.

There are lowpass dynamics involved in the respiratory process that result in taking to some time to achieve volume variations in the lungs. Thus, assuming ventilation and LUCAS signals as synchronous before shifting, a phase shifting of 120 degrees forward with LUCAS signal (equivalent to 0.2 seconds) is applied to the ventilation in order to start every decompression phase with the lowest lung volume possible. The PIIO method is illustrated in Figure 1.3.

More information related to the PIIO method is provided in [Soltesz et al., 2018], which has demonstrated that PIIO has a significant positive effect on CPP compared to CIO in a porcine study.

1.4 Objective

The purpose of this thesis is to design a basic model for analyzing the respiratory mechanics during cardiopulmonary resuscitation. To this end, two simulation models have been studied: RIC model and Mead model (which will be explained in Chapter 2), both of them tested using PIIO and CIO ventilation methods.

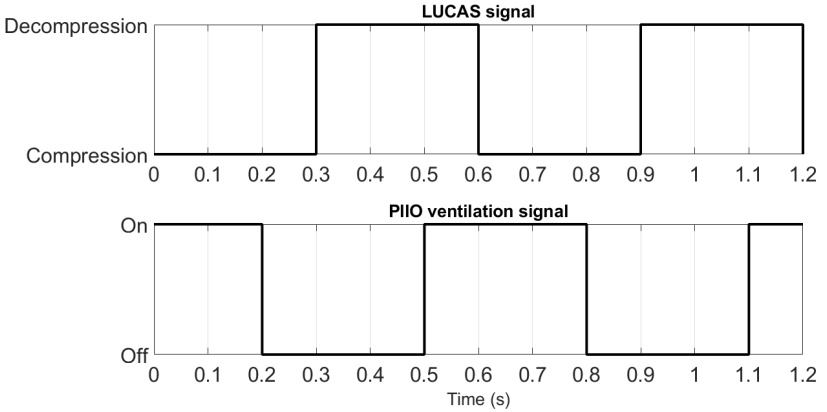

Figure 1.3 Phase shifted synchronization between oxygen insufflation and LUCAS device during PIIO.

The thesis will compare the intratracheal pressure profiles and lung volume variations associated with CIO and PIIO in each model in order to obtain a qualitative characterization. For simplicity, the main goals to be achieved are listed below:

1. Design a basic model for analyzing the respiratory mechanics during cardiopulmonary resuscitation.

2. Compare PIIO and CIO ventilation methods.

1.5 Method

As mentioned in section 1.1, CPP is an important parameter. However, the heart's dynamics are not modelled and therefore the coronary perfusion is not simulated, and neither is the CPP. Since the coronary perfusion takes place every decompression phase depending on the CPP applied, one way to deduce its behaviour is through the intratracheal pressure value during every decompression phase. A lower intratracheal pressure during decompression phase is related to a lower lung volume at the beggining of the decompression, which leads, as mentioned in section 1.3, to a higher CPP.

Besides coronary perfusion, there are also other parameters of interest for evaluating the accuracy of the models.

Taking all into consideration, the study will be focused on the following outputs:

1. Mean intratracheal pressure over a decompression phase cycle. It is an indicator that represents the performance of the coronary perfusion and therefore it is a suitabe variable to look at the simulation results.

2. Minimum intratracheal pressure value over a decompression phase cycle. Since the relation between the coronary perfusion and CPP is not linear, neither is the relation between the coronary perfusion and intratracheal pressure. Due to this, it is considered to take into account the minimum intratracheal pressure of the decompression phase in order to have another indicator that represents the coronary perfusion behaviour.

3. Intratracheal pressure and lung volume response. They provide the basic information needed about the process to understand its behaviour, such as flow direction or how the ventilation method and chest compressions affect the process. Furthemore, the maximum intratracheal pressure difference over a cycle is a value that can be compared with the one obtained in [Soltesz et al., 2018] (which is a study that tested PIIO and CIO ventilation methods in pigs) as well as the amount of volume expelled in each cycle can be contrasted with the human tidal volume (TV, air volume moved in or out during quiet breathing) and TV during mechanical cardiopulmonary resucitation (i.e., considering only LUCAS device, without applying any ventilation).

   Hence, they are both good indicators for evaluating the models. The values obtained in the study, and the tidal volume during quiet breathing[1] and during mCPR ([Steen et al., 2004]) values are shown in Table 1.1.

Table 1.1 Data for evaluating the models (cmH_2O, L).

Ventilation method	Max. pressure difference	TV	TV during mCPR
PIIO	14	0.5	0.08
CIO	7		

[1] Wikipedia, *Lung volumes*, retrieved 2019-01-20

2

Model design

The goal of this thesis is to build a suitable model for analyzing respiratory mechanics during cardiopulmonary resuscitation. Therefore, the model designed will have the outputs mentioned in the previous chapter (lung volume and intratracheal pressure), and 2 inputs corresponding to the oxygen insufflation pressure and the chest compressions applied to the patient during a CPR.

While there are different ways for modelling the human lungs dynamics (pneumatic model, [Shi et al., 2014], non-linear models, [Saatci and Akan, 2007] and [Fix et al., 2018]), an electrical analogy has been applied to the human respiratory system. This analogy has been chosen to linearize the process in an intuitive way, making it easier to change and interpret each part of the model.

The model parameter values have been selected according to [Diong et al., 2007] and [Baswa et al., 2006] (for RIC and Mead model, respectively) which have calculated the minimum impedance error (reactance and resistance) over frequency by means of least squares. In both papers, patient impedance has been obtained through impulse oscillometry tests, followed by least squares fitting (more information about impulse oscillometry technique can be found at [Smith et al., 2005]).

The software Open Modelica has been used instead of other options such as Matlab Simulink. The reason behind this is that it is easier to deal with two-port systems, such as the models used in this thesis, working with differential algebraic equations (DAE, used in Open Modellica) rather than ordinary differential equations (ODE, used in Matlab Simulink).

2.1 Electrical interpretation

The analogy applied to the process has the relations shown in Table 2.1.

There are a few things to take into account:

Table 2.1 Unit conversion of the electrical analogy in International System of Units (SI).

Electrical entity	Electrical units	Hydrological units	Hydrological entity
Current	A	$\frac{m^3}{s}$	Volumetric flow
Voltage	$\frac{Kg \cdot m^2}{A \cdot s^3}$	$\frac{Kg}{m \cdot s^2} = \frac{N}{m^2}$	Pressure
Charge	$A \cdot s$	m^3	Volume
Capacitance	$\frac{A^2 \cdot s^4}{Kg \cdot m^2}$	$\frac{m^5}{N}$	Compliance

- A resistor results therefore in a relation between pressure and volumetric flow rate. It produces a drop of pressure depending on the amount of gas that passes through it.
- Since capacitance constitutes a relation between charge and voltage, a compliance consumes pressure depending on how much is "charged and discharged" of volume, this makes it a suitable component to represent the lungs functionality.
- Similarly, an inductor tends to resist changes in current. Following this in the electrical analogy, the inductor represents opposition to changes in gas flow, acting as a gas flow inertia.

2.2 Resistor-Inductor-Capacitor model

The RIC model has been selected for being the simplest model that captures the basic behaviour of the human respiratory system. Its structure is illustrated in Figure 2.1.

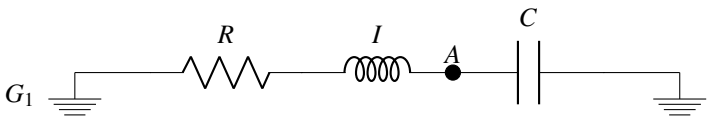

Figure 2.1 Resistor-Inductor-Capacitor model.

The resistor and inductor represent the airway resistance and inertance produced by the trachea in the respiratory process, while the capacitor is added in order to model compliance.

The intratracheal pressure is hence the voltage difference between the dot A and the atmospheric ground G_1, while the lung volume is represented by the charge of the capacitor.

More information about the RIC model can be found in [Ghafarian et al., 2016].

2.3 Mead model

This model was named after Jeremiah Mead, who made an important lung research contribution to society ([Mead et al., 1970], [Mead, 1978]). The reason why Mead model has been chosen is that it is one of the models that has the lowest impedance error according to the impulse oscillometry data, as suggested by several papers, for instance [Woo et al., 2004] and [Nguyen et al., 2007]. Mead model structure is illustrated in Figure 2.2, and its parameter definitions and their representation are shown in Table 2.2.

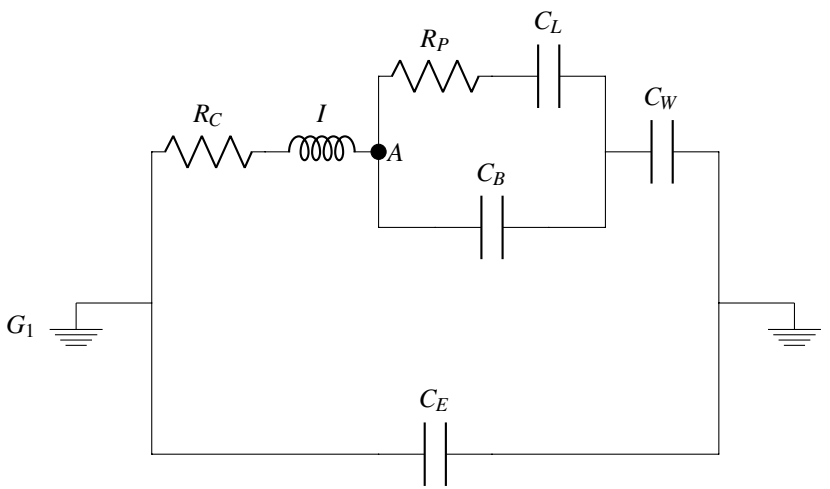

Figure 2.2 Mead model.

The intratracheal pressure is again the voltage difference between the dot A and the atmospheric ground G_1, while the lung volume (deviation from an equilibrium value) is represented by the charge of the capacitor C_L.

More information about the Mead model can be found in [Ghafarian et al., 2016], [Diong et al., 2007].

Table 2.2 Mead model parameter definitions and their representation.

Parameter	Definition	Representation
R_C	Central resistance caused by the trachea	Trachea
I	Tracheal inertance	
R_P	Peripheral resistance caused by the alveoli	Lungs
C_L	Lungs compliance	
C_B	Bronquial tube compliance	
C_W	Chest wall compliance	Chest
C_E	Extrathoracic compliance	Extrathoracic effects

2.4 Implementation

LUCAS device

At first, compressions and decompressions produced by LUCAS device were considered as pressure affecting directly to the lungs. Thus, it had been considered as a squared signal voltage source placed next to the capacitor that represents lungs compliance.

Since LUCAS device returns the chest to its original position, the voltage applied will be positive during the compression phase and null during the decompression phase. The positive value of the signal was calibrated so that the variation of the PIIO intratracheal pressure observed in the porcine study [Soltesz et al., 2018] matched with the one obtained from the simulation; this calibration was applied to both models (RIC and Mead).

This means that the models will be evaluated based on the results of the porcine study while the models used have been built using human data. This may suggest that the human maximum intratracheal pressure difference value of the ventilation methods may be different from the ones used. Nevertheless, the study population was chosen to adhere to the Utstein guidelines [Idris et al., 1996], which is a protocol specifying the size of pigs to use in CPR for which pulmonary mechanics are similar to an adult human.

Ventilation signal

Since the models have been calibrated based on the porcine study, it should imitate the study's ventilation procedure.

The ventilation technique used consists of applying oxygen flow through a boussignac tube which is placed in the trachea. Hence, applying the electrical analogy, it is modelled as a voltage source (in order to represent the oxygen's pressure applied) and a resistor (representing the resistance produced by the boussignac tube). Those will be connected between the atmosphere and at the end of the trachea as seen in Figure 2.4 and 2.5.

The voltage source will produce a squared or constant signal, depending on whether the ventilation technique is PIIO or CIO, respectively.

In the porcine study, it was also stated that for a pressure difference of 2 bar in the boussignac tube, the oxygen flow was 15 L/min. With this information, the resistance of the boussignac tube has been calculated according to Equation 2.1

$$R_{BT} = \frac{2}{15}\frac{\text{bar}}{\text{L/min}} = 8157.76\frac{\text{cmH}_2\text{O}}{\text{L/s}}. \tag{2.1}$$

Improvements

With the purpose of having a more realistic simulation, the following improvements have been implemented:

- Instead of a squared signal, LUCAS input signal is modelled as a trapezoidal signal with a rising/falling time of 0.1 seconds, matching the behaviour of the actual device.
- Modelling compressions and decompressions provided by LUCAS device as pressure affecting directly to the lungs is very inaccurate. A more precise way to model it is by considering each compression as a reduction of the model's volume. Since the hydrologic analogy of the capacitance represents the ratio between volume and pressure, a lower capacitance turns into a lower volume for the same pressure. Thus, the capacitance will be decreased in every decompression phase.

  An illustration of the signal's behaviour is shown in Figure 2.3.

  During every decompression phase, LUCAS returns the chest to its natural position. Thus, the capacitance values in every decompression phase for each model are still the ones obtained through the impulse oscillometry data. The capacitance value in every compression phase is selected following the same

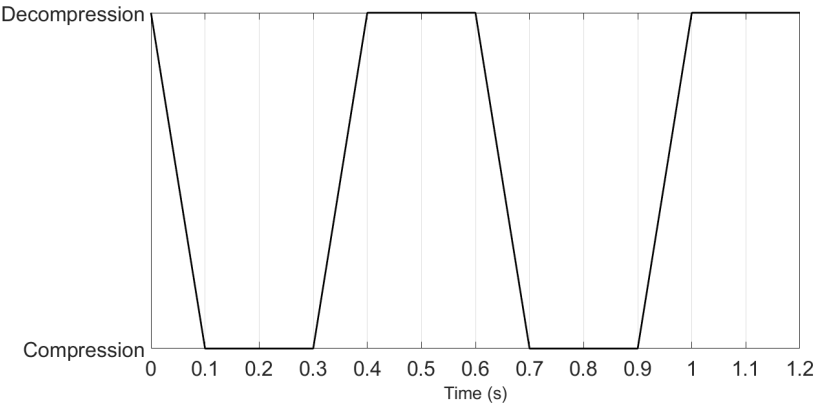

Figure 2.3 Trapezoidal signal for lungs variable capacitance.

logic as before: so that the variation of the PIIO intratracheal pressure shown in the porcine study matches with the one obtained from the simulation.

This variable capacitance is applied to the unique capacitor in RIC model (C shown in Figure 2.1), while in the Mead model is assigned to the chest capacitor (C_W shown in Figure 2.2) since it is where the compressions are applied in reality.

Finally, it is important to appreciate that by considering a variable capacitor in the models, they lose their linearity and therefore it is needed to work in absolute values of pressure.

- As mentioned, absolute pressure is needed in order to make the models work. To do it, a constant voltage source is added before the ground that represents the atmosphere.

Applying the LUCAS and oxygen insufflation signal, and all the improvements mentioned, the resulting RIC and Mead models are shown in Figure 2.4 and Figure 2.5, respectively.

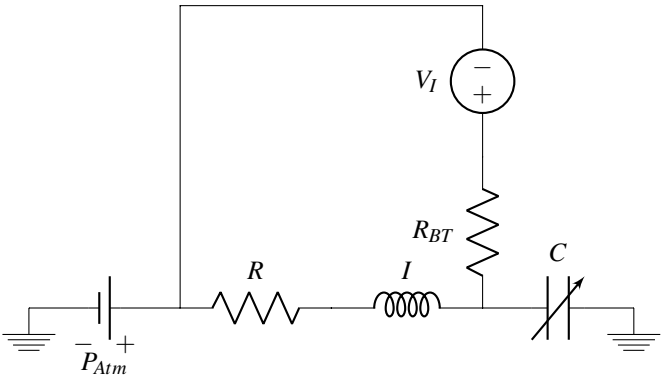

Figure 2.4 Adapted RIC model.

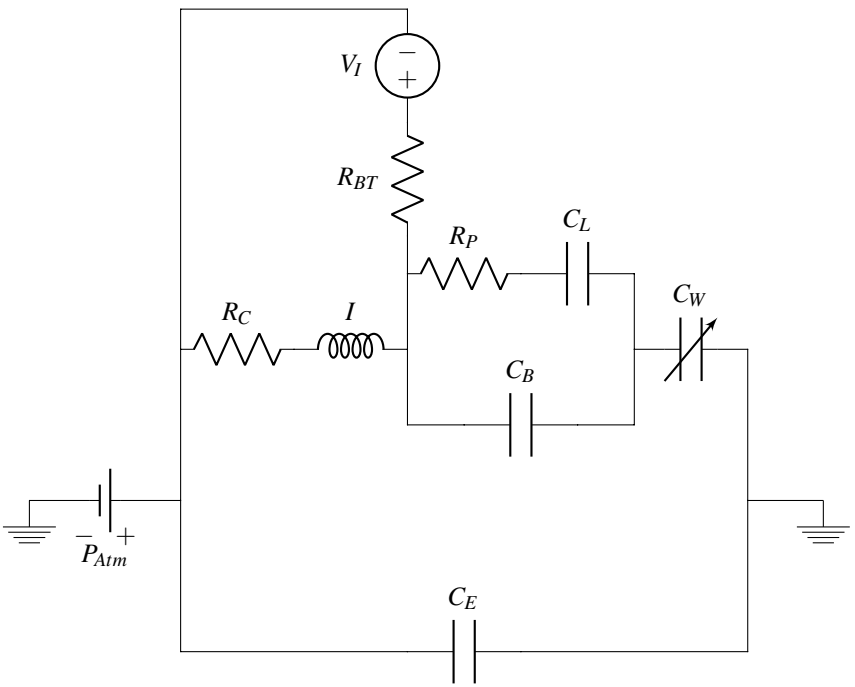

Figure 2.5 Adapted Mead model.

3

Results

The results shown have a time of 1.2 seconds (equivalent to 2 cycles) starting with a decompression phase. Below each graph the LUCAS and PIIO ventilation signals have been included in order to give a clearer perception.

3.1 RIC model

The intratracheal pressure response is shown in Figure 3.1, while the lung volume response can be seen in Figure 3.2.

The mean intratracheal pressure is 25.66% lower for CIO compared to PIIO, while the minimum intratracheal pressure is 10.41% lower in CIO (all values obtained over a single decompression phase).

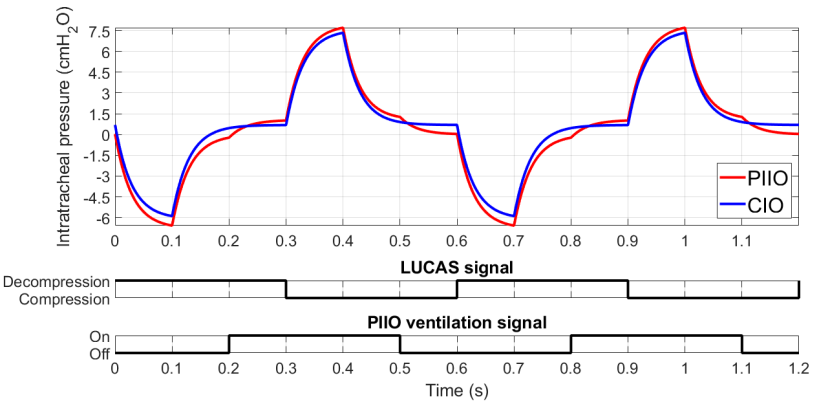

Figure 3.1 CPR PIIO and CIO intratracheal pressure responses simulated in RIC model.

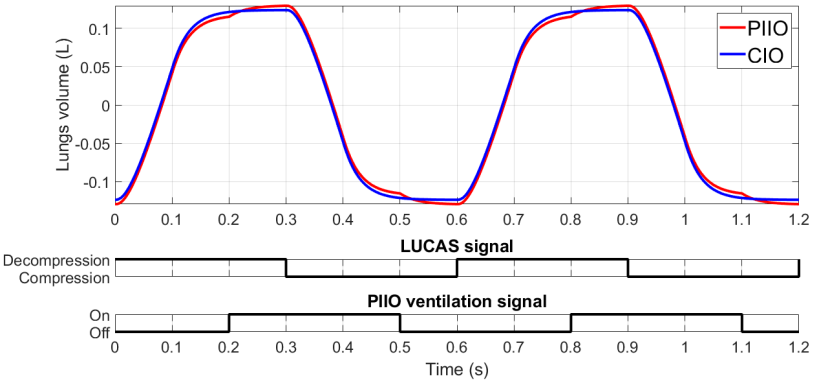

Figure 3.2 CPR PIIO and CIO lungs volume responses simulated in RIC model.

3.2 Mead model

The intratracheal pressure and lung volume responses are shown in Figures 3.3 and 3.4, respectively.

The mean and minimum intratracheal pressure over a single decompression phase are 12.10% and 8.38% lower for CIO compared to PIIO, respectively.

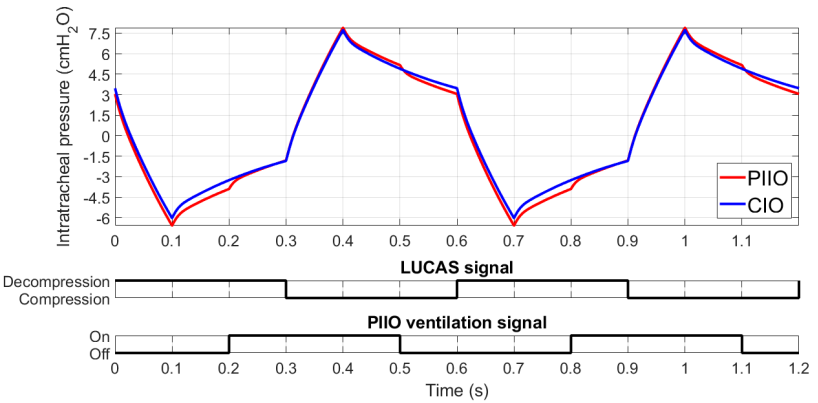

Figure 3.3 CPR PIIO and CIO intratracheal pressure responses simulated in Mead model.

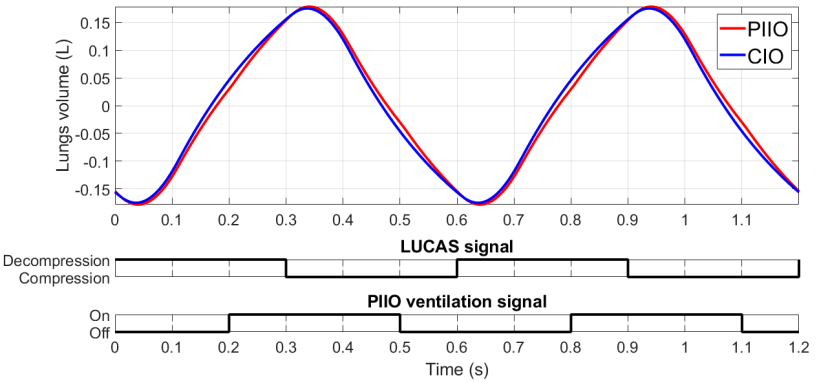

Figure 3.4 CPR PIIO and CIO lungs volume responses simulated in Mead model.

4

Discussion

Concerning the suitability of the models, it is important to remember that a calibration has been applied to the capacitance value during the compression phase in order to match the PIIO maximum intratracheal pressure difference over a cycle in both models (shown in Table 1.1, page 6). As a consequence, the meaningful parameters to evaluate the models will be only the maximum intratracheal pressure for the CIO case and the lung volume variation mentioned in section 1.5.

It can be seen in Figures 3.1 and 3.3 that, even though the CIO case presents a lower maximum intratracheal pressure difference than PIIO, it does not reach the value of 7 cmH_2O observed in the porcine study in either the RIC or Mead model. Nevertheless, it captures the fundamental physiological behaviour of the process. It can be observed in Figure 3.2 how the decompressions and compressions generated by the LUCAS device generate the inhalation and exhalation, respectively, as well as how the PIIO method results in emptier lungs at the beginning of each decompression phase. Similarly, it can also be observed in Figures 3.2 and 3.4 that the lung volume variation is around 0.25 L and 0.34 L, respectively, which are values between the ones presented in Table 1.1, as expected.

The mead model shows slower dynamics than the RIC model. This can be seen in Figure 3.3, where the responses do not reach steady state. A continuous increase of the difference between PIIO and CIO intratracheal pressure is observed from 0.6 s to 0.7 s, which suggests that the fact that the Mead model does not reach a steady state leads to a less significant difference between CIO and PIIO than the one seen in RIC model's response.

With reference to the results of the ventilation method comparison, it has been shown in chapter 3 that in both models PIIO has a lower mean and minimum intratracheal pressure over a decompression phase than CIO, which indicates a higher CPP and therefore better coronary perfusion, as observed in the porcine study.

5

Conclusions

Based on the results discussed, it is concluded that, although the models give values that do not match exactly with the results obtained in the porcine study, they capture qualitative differences between the ventilation methods that make PIIO a better ventilation method in terms of CPP (lower minimum and mean intratracheal pressure values over a decompression phase and higher lung volume variation with PIIO than with CIO, as in the porcine study). This can also be seen in how lung volume and intratracheal pressure are affected by the ventilation method applied or in the lung volume variation values compared to the tidal volumes seen in Table 1.1.

As it has been shown in chapter 3, the Mead model yields less maximum intratracheal pressure difference, partly because the intratracheal pressure cannot reach its steady state. The RIC model is simple and therefore intuitive and easy to understand, which implies the fact that it is also easier to apply hypotheses on it. In the Mead model it is more difficult to represent a chest compression in terms of which components are affected. This important property makes RIC model more likely to work with it in the matter of modelling.

6

Limitations

As discussed in chapter 5, the results provide useful qualitative insight of the process, but further work is needed in order to obtain a reliable quantitative comparisons, especially data on which to base ventilation and compression variations of the considered respiratory mechanics models.

It can be demonstrated that the oxygen flow in the smaller lumina of the boussignac tube produces a substantial suction flow through its main lumen as a result of the Venturi effect, and this has not been taken into account in the models design. In the same way, the models consider the same value for the intrathoracic pressure and lung pressure by not having any component that represents the thorax's dynamics. These are the ones of the main reasons for not having better quantitative comparisons between the ventilation methods.

The chest compression synchronized ventilation (CCSV) method, which applies intermittent positive pressure similar to PIIO, but, in this case, the ventilation signal is phase-shifted 180 degrees with LUCAS signal ([Kill et al., 2015]), was modelled too. However, due to the unmodelled dynamics of the process, the effect of the phase-shifting could not be captured (i.e., there was no difference in the delay of the response produced by the ventilation between PIIO and CCSV).

Finally, CPP is not explicitly modelled. Instead we rely on how it is affected by intratracheal pressure variations over the compression cycle to draw conclusions. Adding a heart dynamics model would be useful in order to directly simulate CPP, being therefore able to build more accurate models by trying to match the CPP results obtained in studies. This would also add more valuable variables to compare during the test, such as the accomplishment of the minimum CPP needed for a successful ROSC upon defibrillation (15 mmHg, [Paradis et al., 1990]).

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