Anesthesiology Research and Practice

Electronic Computer-Based Model of Combined Ventilation Using a New Medical Device

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Contributions of each author to the paper: each author contributed to: literature search, data collection, study design, analysis of data, manuscript preparation, and review of manuscript.

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Abstract

Introduction

The increased demand for mechanical ventilation caused by SARS-CoV-2 could generate a critical situation where patients may lose access to mechanical ventilators. Combined ventilation, in which two patients are connected to a single ventilator has been proposed as a bridge while waiting for new ventilators availability. DuplicAR is a new device that allows individualization of ventilatory parameters in combined ventilation models.

Materials and Methods

With an electronic circuit simulator applet, an electrical model of combined ventilation was created using resistor-capacitor circuits. The DuplicAR system electrical analog was added to the model. Through computational simulation, the model is tested in different scenarios with the aim of achieving adequate ventilation of two subjects under different circumstances: 1) two identical subjects; 2) two subjects with the same size but different lung compliance; and 3) two subjects with different size and compliance. The goal is to achieve the established load per unit of size on each capacitor under different levels of end-expiratory voltage (analog of end-expiratory pressure). Data collected included capacitor load and voltage, and load normalized to the weight of the simulated patient.

Results

In the three simulated stages, it is possible to provide the proper load to each capacitor under different circumstances. If the pair of connected capacitors have different capacitances, adjustments must be made to the source voltage and/or the resistance of the DuplicAR system to provide the appropriate load for each capacitor under initial conditions. In pressure control simulation, increasing the end-expiratory voltage on one capacitor requires increasing the source voltage and the resistance on the other capacitor. On the other hand, in the volume control simulation, it is only required to intervene in the resistance.

Conclusions

Under simulated conditions, the electrical model of the DuplicAR system allows individualization of combined mechanical ventilation.

Keywords: SARS-CoV-2, pandemic, mechanical ventilation, medical device, computational simulation, electrical model.

Introduction

Acute respiratory distress syndrome caused by SARS-CoV-2 increased the demand for mechanical ventilation worldwide [1]. An imbalance between supply and demand for mechanical ventilators could pose a critical scenario in which it would be necessary to decide which patients should be assigned to these devices and which ones should not [2, 3, 4].

Combined ventilation has been postulated as a strategy to face this problem [5,6]. Although this setup may be simple in principle, the dynamics of two patients connected to the same ventilator may require deep understanding of advanced concepts about mechanical ventilation. This fact has led some scientific societies to disencourage this configuration. One of the biggest arguments against this arrangement is the inability to personalize ventilatory parameters for each patient [7].

DuplicAR is a novel device that allows simultaneous and independent ventilation of two subjects with only one ventilator [8]. The goal of this device is to enable mechanical ventilation of two patients with a single ventilator, without cross-contamination, and allowing for independent management of the inspiratory pressure (Pi) and positive-end expiratory pressure (PEEP). In this work we propose an electrical simulation model of combined ventilation and the *in silico* test of the DuplicAR device.

Materials and methods

The DuplicAR device

DuplicAR is a medical device that functions as a complementary adapter to the mechanical ventilator and has the ability to offer adequate and independent pressurization of the system for the two subjects. It consists of two adapters. The inspiratory adapter connects to the inspiratory port of the ventilator and to each subject´s inspiratory line (Figure 1). Each inspiratory line can control the inspiratory pressure through a diameter (i.e. resistance) regulator that allows independent management of the tidal volume (Vt). The expiratory adapter connects to each subject's expiratory line and to the expiratory port of the ventilator. Each expiratory line has a positive end-expiratory pressure (PEEP) controller for independent management of this variable. Cross-contamination is prevented through one-way valves and microbiological filters in each line of the circuit. The effectiveness of the DuplicAR device for combined ventilation is studied in a computer simulation ("*in silico*"), using an electrical model.

Electrical Model Simulation

Our electric combined ventilation model is made up of two systems: the "ventilator system" and the "subject system" (Figure 2). The ventilator system includes a constant voltage or current (depending on simulated ventilation mode) and a ground connection. The first is a power source that represents the inspiratory part of the ventilator, while the second represents the expiratory part. The subject system is represented by a series connection of an electrical resistance and a capacitor (RC circuit).

The connection between both systems is achieved through a relay. The relay is a device that works as a switch triggered by an external circuit. The current from the external circuit passes through a coil generating a magnetic field that moves the switch of the relay. In this way, the switch moves from the power source to ground in a cyclical manner, governed by the current in its external circuit. By manipulating the current in the external circuit, the behavior of the relay can be controlled. Thus, a ventilatory cycle is represented by a current pulse in the external circuit that moves the switch (relay) to the power source and then at the end of the pulse, back to ground. The percentage of the cycle that the external circuit source maintains the voltage across the coil (and therefore the switch on “inspiration”) is determined by the “active cycle” of the relay. For example, set to 33% would represent an I:E ratio of 1:2. The ventilatory rate is represented by the pulse rate. As the “inspiratory port” has a constant voltage source, the system represents the pressure control mode of mechanical ventilation (PCV); When a constant current source is used, the system represents volume control ventilation (VCV). The flows in the inspiratory and expiratory lines are directed by diodes.

To simulate the PEEP, a Zener diode with modifiable breakdown voltage is added prior to grounding the expiratory port. The Zener is a special type of diode that always allows current to flow in one direction, but only allows it in the opposite direction if a certain voltage threshold is exceeded (known as "breakdown voltage"). By setting a zero breakdown voltage, the diode behaves like a simple conductor. Setting a nonzero breakdown value, the diode will “catch” that voltage from “behind” and not allow current to flow to ground. Thus, the Zener diode functions as the equivalent of a PEEP valve.

Figure 3 shows the records obtained during the simulation of PCV (to see the simulation running, video clips can be found on *Supplementary Material* section) .

A theoretical weight is established for each simulated subject to determine the charge with which that RC circuit must be “ventilated” (charged). This normalized charge is arbitrarily chosen at 7 μC/kg.

Given the "fast" nature of electrical phenomena, the basal cycling frequency of the ventilator system is set at 45 Hz and the active cycle at 33.33%, generating cycles of 22.2 ms (1000/45) and “inspiratory times” of 7.39 ms (0.33×22.2). Given the RC circuit characteristics of the simulated subjects, at least 10 expiratory time constants are allowed to take place with this configuration to ensure no charge trapping. This cycling frequency is considered equivalent (in mechanical ventilation) to a respiratory rate of 10 to 12 ventilations per minute in patients with normal resistance and compliance (Table 1) [9].

Combined ventilation model involves placing two “patient systems” (ie. two RC circuits) in parallel with a single power source.

The DuplicAR System is simulated with a variable resistor in the inspiratory line and a Zener diode in the expiratory line of each subject (Figure 2, in blue). For each “patient system”, adjusting the resistance controls the voltage received; and adjusting the Zener breakdown voltage controls the end-expiratory voltage (EEV, equivalent to end-expiratory pressure)

The electrical simulations are performed in Falstad Circuit Simulator 2.2.13js, a Java-based electronic simulator [10]. Equivalences between models are shown in Table 1.

Variable	Real model	Electrical equivalent
Weight	50 kg	50 kg
Target volume or charge	7 ml/kg	7 μC/kg
Compliance or Capacitance	30 ml/cmH2O	30 μF
Resistance	14 cmH2O / l/s	40 Ω
Cycle frequency	10 / minute	45 Hz
Inspiration	2 s	7.39 ms
Expiration	4 s	14.81 ms
I:E Ratio	1:2	1:2
Time constant (τ)	0.42 s	1.2 ms
τ allowed in expiration	9.5	12.3

Table 1. Equivalences between the mechanical and electrical models.

Experimental Setting

Three patient models are simulated, each with particular characteristics of weight and compliance (Table 2).

Subject / RC circuit	Weight	Resistance	Capacitance	Capacitance/Kg
A	50 kg	40 Ω	30 μF	0.6 μF/kg
B	50 kg	40 Ω	20 μF	0.4 μF/Kg
C	28 kg	40 Ω	10 μF	0.35 μF/Kg

Table 2. Types of simulated subjects.

With the three patient models, different stages are recreated:

Stage 1: Two identical RC circuits (A and A) *ventilated* in PCV and VCV modes.

Stage 2: Two RC circuits with the same *size* but different capacitances (A and B), *ventilated* in PCV and VCV modes.

Stage 3: Two RC circuits with different *sizes* and capacitances (A and C), *ventilated* in PCV and VCV modes.

In each of these stages, progress is made through different cases:

Case I: it is considered the “baseline situation”, in which the objective is to achieve the appropriate charge (7 units of charge per size units) for each RC circuit with an end-expiratory voltage (EEV) of 5 V.

Case II: the objective is to achieve the appropriate charge (7 units of charge per size unit) for each RC circuit, but increasing the EEV in one RC circuit to 10 V.

Case III: the objective is to achieve the appropriate charge (7 units of charge per size unit) for each RC circuit, but increasing the EEV in one RC circuit to 15 V.

The three stages are summarized in Table 3.

	Stage 1		Stage 2		Stage 3	
Subject / RC circuit	A	A`	A	B	A	C
Size (kg)	50		50		50	28
Capacitance (μF)	30		30	20	30	10
Resistance (Ω)	40		40		40	
Capacitance/size (μF/kg)	0.6		0.6	0.4	0.6	0.35
I:E	1:2		1:2		1:2	
Cycle frequency (Hz)	45		45		45	
Target load/size (μC/kg)	7		7		7	
Target charge (μC)	350		350		350	196
Target EEV Case I (V)	5		5		5	
Target EEV Case II (V)	5	10	5	10	5	10
Target EEV Case III (V)	5	15	5	15	5	15

Table 3. Summary of the 3 stages, with the characteristics of each subject, main configuration of the ventilator, the target charge and the EEV of each subject.

Stage 1

Two 50 kg subjects are simulated, represented by an RC circuit with a capacitance of 30 μF (0.6 μF/kg) and a resistance of 40 Ω. A target charge of 7 μC/kg is established, 350 μC for each capacitor. At basal conditions (case I), Zener breakdown voltage of the ventilator is set to 5 V, the inspiratory voltage to 17.5 V in PCV and the inspiratory current to 94.5 mA in VCV mode. Given that cycles have a frequency of 45 Hz and the I:E ratio is 1:2, the inspiratory time (Ti) is 7.39 ms. During that time, 700 μC is delivered in VCV mode (7.39 ms $\times$ 94.5 mA), representing the sum of the target charges for each subject.

Stage 2

Two 50 kg subjects are simulated and represented by RC circuits. Subject A capacitance is 30 μF (0.6 μF/kg) and Subject B at 20 (0.4 μF/kg). Both resistances are set to 40 Ω. A target charge of 7 μC/kg is established, being 350 μC for each capacitor. As a baseline condition (case I), Zener diode breakdown voltage of the ventilator is set to 5 V, the inspiratory voltage to 23.3 V in PCV mode and the inspiratory current to 94.5 mA in VCV mode. Given that cycles have a frequency of 45 Hz and the I:E ratio is 1:2, the inspiratory time (Ti) is 7.39 ms. During that time, 700 μC is delivered in VCV mode (7.39 ms $\times$ 94.5 mA), representing the sum of the target charge for each subject.

Stage 3

Two subjects with different sizes and capacitances are simulated. Simulated subject A size is 50 kg and capacitance is 30 μF (0.6 μF/kg). Simulated subject C size is 28 kg and capacitance is 10 μF (0.35 μF/kg). Both resistances are set to 40 Ω. A target charge of 7 μC/kg is established, resulting in 350 μC for capacitor A and 196 μC for capacitor C. As a baseline condition (case I), the Zener diode breakdown voltage of the ventilator is set to 5 V, the inspiratory voltage to 25.4 V in PCV mode and the inspiratory current to 73.71 mA in VCV mode. Given that cycles have a frequency of 45 Hz and the inspiration:expiration ratio is 1:2, the inspiratory time (Ti) is 7.39 ms. During that time, 546 μC is delivered in VCV mode (7.39 ms $\times$ 73.71 mA), representing the sum of the target charges for each subject.

In all stages, by regulating the Zener diode breakdown voltage in the DuplicAR device, the EEV of one RC unit is raised to 10 V and 15 V, representing cases II and III in each stage. Necessary adjustments are made in the inspiratory voltage of the ventilator and/or resistance of the inspiratory line of the DuplicAR device in the other subject to ensure the correct charge for each one.

Results

Stage 1

Data is shown in Table 4.

Stage 1		Case I		Case II		Case III	
Subject / RC circuit		A	A´	A	A´	A	A´
PCV	Vinsp ventilator (V)	17.5		22.5		27.5	
	Vmax (V)	17.0	17.0	17.1	22.0	17.0	27.0
	Vmin (V)	5.4	5.4	5.4	10.3	5.4	15.3
	ΔV (V)	11.6	11.6	11.7	11.7	11.7	11.7
	Charge (μC)	348.5	348.5	350.9	349.8	350.2	349.8
	Charge/Size (μC/kg)	7.0	7.0	7.0	7.0	7.0	7.0
	R DuplicAR (Ω)	0.0	0.0	160.0	0.0	275.0	0.0
	Zener DuplicAR (V)	0.0	0.0	0.0	5.0	0.0	10.0
VCV	Total charge delivered (μC)	700.0		700.0		700.0	
	Vmax (V)	17.1	17.1	17.1	22.1	17.1	27.1
	Vmin (V)	5.5	5.5	5.5	10.4	5.5	15.4
	ΔV (V)	11.7	11.7	11.7	11.7	11.7	11.7
	Charge (μC)	350.0	350.0	349.9	350.1	349.8	350.2
	Charge/Size (μC/kg)	7.0	7.0	7.0	7.0	7.0	7.0
	R DuplicAR (Ω)	0.0	0.0	105.0	0.0	211.0	0.0
	Zener DuplicAR (V)	0.0	0.0	0.0	5.0	0.0	10.0

Table 4. Data from Stage 1 simulation.

It is first evidenced that, without manipulation of the DuplicAR device, the voltage difference in each capacitor is the same (11.64 ± 0.03 V), and therefore its charge is also the same (349.28 ± 0.85 μC). Charge on each capacitor is calculated according to $q = C \times \Delta V$.

In both ventilatory modes, PCV and VCV, it is possible to modify the EEV of one RC unit (ie. the subject) without compromising its driving voltage or modifying the parameters of the other (Figures 4 and 5). The *driving voltage* is the difference between the maximum and minimum voltage of the capacitor during a cycle, equivalent to driving pressure in mechanical ventilation. To increase the EEV of a RC unit, the Zener breakdown voltage corresponding to that unit (on DuplicAR) must be increased.

In each of the three cases, the driving voltage is preserved. This is accomplished in PCV mode by two maneuvers: first, increasing the inspiratory voltage of the power source in the same amount as the Zener breakdown voltage; second, restricting the voltage to the other by regulating the inspiratory resistance in the DuplicAR device (Figure 6).

In VCV mode, as the EEV of one RC unit increases, it is necessary to adjust the inspiratory resistance of the DuplicAR device in the other (Figure 7). This is mandatory since, otherwise, no charge would enter the first RC unit until the breakdown voltage of the Zener diode is reached, causing a disproportionate increase in charge in the second one.

Stage 2

Data is shown in Table 5.

Stage 2		Case I		Case II		Case III	
Subject / RC circuit		A	B	A	B	A	B
PCV	Vinsp ventilator (V)	23.3		28.3		33.3	
	Vmax (V)	17.0	22.9	17.0	27.9	17.1	32.9
	Vmin (V)	5.3	5.3	5.4	10.3	5.4	15.3
	ΔV (V)	11.7	17.6	11.7	17.6	11.7	17.6
	Charge (μC)	350.2	351.0	350.9	352.2	351.5	352.2
	Charge/Size (μC/kg)	7.0	7.0	7.0	7.0	7.0	7.0
	R DuplicAR (Ω)	180.0	0.0	292.0	0.0	400.0	0.0
	Zener DuplicAR (V)	0.0	0.0	0.0	5.0	0.0	10.0
VCV	Total charge delivered (μC)	700.0		700.0		700.0	
	Vmax (V)	17.1	23.0	17.1	27.9	17.2	32.8
	Vmin (V)	5.5	5.4	5.5	10.4	5.5	15.4
	ΔV (V)	11.6	17.5	11.6	17.6	11.7	17.5
	Charge (μC)	349.1	350.9	349.0	351.0	350.5	349.7
	Charge/Size (μC/kg)	7.0	7.0	7.0	7.0	7.0	7.0
	R DuplicAR (Ω)	93.0	0.0	189.0	0.0	287.0	0.0
	Zener DuplicAR (V)	0.0	0.0	0.0	5.0	0.0	10.0

Table 5. Data from Stage 2 simulation.

Both simulated subjects are the same size, resulting in the same charge requirement (350 μC). Since RC circuit B presents a reduced capacitance, it requires a greater driving voltage than RC circuit A ($350\ \mu C/30\ \mu F = 11.7\ V$ for circuit A; $350\ \mu C/20\ \mu F = 17.5\ V$ for B). To achieve these voltage differences in PCV mode, the inspiratory voltage needs to be first configured in the ventilatory system to guarantee the charge to the RC circuit with the lowest capacitance and then to be decreased in the other RC circuit using the inspiratory resistance of the DuplicAR device (Figure 8, case I).

In VCV mode, total configured charge is delivered by the ventilatory system to the group, but its distribution results asymmetric regarding differences in RC circuits characteristics. The precise charge distribution for each subject needs to be regulated by the inspiratory resistance regulators of DuplicAR device in the patient with greater capacitance (Figure 9, case I).

In this way, with a single ventilator configuration, different driving voltages are established. In both ventilatory modes (PCV and VCV) it is possible to modify the EEV of a RC circuit without compromising its driving voltage or modifying the other circuit´s parameters (Figures 10 and 11).

In each of the three cases, the driving voltage is preserved. In PCV mode, this is accomplished by increasing the inspiratory voltage of the ventilator system to maintain the driving voltage of the RC circuit with an increased EEV. Accurate driving voltage in RC circuit A is modulated by the inspiratory resistance of the DuplicAR device, preventing excessive voltages. In VCV mode the total charge is set. Since EEV of RC circuit B increases, the inspiratory resistance of RC circuit A needs to be adjusted in order to achieve an adequate charge distribution.

Stage 3

Data is shown in Table 6.

Stage 3		Case I		Case II		Case III	
Subject / RC circuit		A	C	A	C	A	C
PCV	Vinsp ventilator (V)	25.4		30.4		35.4	
	Vmax (V)	17.1	25.1	17.0	30.1	17.0	35.1
	Vmin (V)	5.3	5.3	5.3	10.2	5.3	15.2
	ΔV (V)	11.7	19.8	11.7	19.9	11.7	19.9
	Charge (μC)	352.1	197.9	350.6	198.8	350.9	198.8
	Charge/Size (μC/kg)	7.0	7.1	7.0	7.1	7.0	7.1
	R DuplicAR (Ω)	226.0	0.0	339.0	0.0	447.0	0.0
	Zener DuplicAR (V)	0.0	0.0	0.0	5.0	0.0	10.0
VCV	Total charge delivered (μC)	546.0		546.0		546.0	
	Vmax (V)	17.1	25.0	17.1	29.8	17.1	34.8
	Vmin (V)	5.5	5.4	5.5	10.3	5.5	15.3
	ΔV (V)	11.7	19.6	11.7	19.6	11.7	19.5
	Charge (μC)	349.7	196.4	350.1	195.6	350.3	195.4
	Charge/Size (μC/kg)	7.0	7.0	7.0	7.0	7.0	7.0
	R DuplicAR (Ω)	116.0	0.0	207.0	0.0	304.0	0.0
	Zener DuplicAR (V)	0.0	0.0	0.0	5.0	0.0	10.0

Table 6. Data from Stage 3 simulation.

It is important to highlight the disparity of voltage differences needed to ensure the target load. Capacitor A requires a driving voltage of 11.7 V to reach 350 μC ($350\ \mu C/30\ \mu F = 11.7\ V$) and capacitor C requires 19.6 V to achieve 196 μC ($196\ \mu C/10\ \mu F = 19.6\ V$), reflecting the difference in their *sizes* and capacitances. In both cases the RC units receive the target charge of 7 μC/kg. To achieve these voltage differences in PCV mode, the inspiratory voltage needs to be first configured in the ventilatory system to ensure the charge to the RC circuit with the lowest capacitance and then to be decreased in the other using the inspiratory resistance of the DuplicAR device (Figure 12, case I).

In VCV mode, total configured charge is delivered by the ventilatory system to the group, but its distribution results asymmetric regarding differences in RC circuits characteristics. The precise charge distribution for each RC circuit needs to be regulated by the inspiratory resistance controllers of the DuplicAR device in the unit with greater capacitance (Figure 13, case I).

In this way, with a single ventilator configuration, different driving voltages are established. In both ventilatory modes (PCV and VCV) it is possible to modify the EEV of a RC circuit without compromising its driving voltage or modifying the other circuit´s parameters (Figures 14 and 15).

In each of the three cases, the driving voltage is preserved. In PCV mode, this is accomplished by increasing the inspiratory voltage of the ventilator system to maintain the driving voltage of the RC circuit with an increased EEV. Accurate driving voltage in capacitor A is modulated by the inspiratory resistance of the DuplicAR device, preventing excessive voltages. In VCV mode, total charge is set. Since EEV of capacitor C increases, the inspiratory resistance of RC circuit A needs to be adjusted in order to achieve an adequate charge distribution.

Discussion

Understanding complex systems can be facilitated by using reductionist models which describe the behavior of variables in simpler ways. With this approach, mechanical ventilation can be represented by an electrical model [11]. By simplifying the respiratory system to an electrical model, we seek to find the relationship between variables of interest (pressure, volume, flow, etc.).

In our simulation system, each variable in mechanical ventilation has an electrical equivalent. The variables pressure, volume, flow, resistance and compliance have their equivalent (voltage, charge, current, electrical resistance and capacitance, respectively). On the other hand, the components of the ventilatory circuit, such as tubes and valves, are equivalent to cables and diodes, respectively.

According to the equation of motion, pressure required to drive gas into the airways and inflate the lungs is caused by the resistive and elastic elements. Lung inflation pressure (P_{aw}) can be expressed:

$$P_{aw} = F \times R_{aw} + Vt/C_{rs}$$

Where F is flow; R_{aw} airway resistance; Vt tidal volume and C_{rs} compliance of the respiratory system. Using the electrical analog of this equation, the respiratory system can be modeled as an RC circuit, with an electrical resistance connected in series with a capacitor. In these types of circuits, the total voltage (ΔV) should be equal to the sum of voltages on the resistor and capacitor. In this way, the equation that describes the behavior of the system is now:

$$\Delta V = I \times R + q/C$$

Where ΔV is the potential difference applied to the system, I is the current that flows, R the electrical resistance, q the charge of the capacitor and C the capacitance.

Several studies have addressed the issue of multiple ventilation [12, 13]. In most of these studies, there is no capacity to safely and effectively control the ventilatory parameters for each patient. The only scenario in which adequate combined ventilation is achieved without the need to intervene is in the presence of two identical subjects (same size and compliance) and with the same PEEP. In COVID-19 patients, ventilatory requirements can be quite disparate, and also can change over time. Where changes in compliance and/or resistance occur, there can be rapid and substantial alteration in the Vt delivered to the other patient.

In these cases, individualization of the ventilatory parameters is mandatory, and devices such as the DuplicAR system are necessary. DuplicAR system proved to be useful in the ventilation of animals in a previous pilot work (in vivo testing) [8]. Similarly, and with results pending publication [14], the device was also evaluated in test lungs (in vitro test) with results equivalent to those of the present simulation (in silico testing).

The present work shows the performance of the device in an electronic computer-based model. With this tool it is possible to generate "cleaner" scenarios and even push the device to the limit safely, quickly and with practically no cost.

The results of simulations, transferred to mechanical ventilation, show that it is possible to ventilate two subjects with different sizes and/or compliances and/or PEEP requirements adequately. According to the model, this could be achieved by adjusting the Pi of the ventilator or the PEEP valve of each subject or the inspiratory resistances of the DuplicAR device. The type and magnitude of the adjustment in each component of the model depends on the ventilatory mode and the characteristics of the subjects.

While these simulations show that the ventilatory goals can be achieved in any of the ventilatory modes, PCV has advantages over VCV. In PCV, the driving pressure can be established and delivered during the inspiratory cycle to both subjects, regardless of their RC characteristics, that is, of their airway resistance and pulmonary compliance. In this mode, there is certainty about the Pi of each subject. As a general limitation of this ventilation mode, it is not possible to know the volume that will be delivered to the system or to each subject. On the other hand, in VCV mode, a total volume to be delivered must be established, that is, the sum of the Vt of both subjects. This volume is not delivered equally to each subject. It is rather distributed among them according to their RC characteristics, and there is no control over the Pi and the Vt.

Although not demonstrated in our simulation, another advantage of PCV mode is that changes in RC characteristics of one subject (i.e. dynamical compliance variability, endotracheal tube obstruction, etc.) do not affect pressurization of the system, and therefore have no consequences on the contralateral subject. On the contrary, in VCV mode, changes in resistance or compliance in one subject directly impact the other subject.

In the model, a similar phenomenon is observed when modifying the EEV in one RC circuit. By increasing the EEV in VCV, the contralateral RC circuit receives a greater charge, since it is necessary that the inspiratory lines reach greater voltage to allow charge to be delivered to the RC circuit with greater EEV. In other words, greater proximal voltage is needed to generate a potential difference that allows charge entry into the RC circuit. This is accomplished at the expense of increasing the charge on the capacitor with no modification of the EEV. In contrast, in PCV, the increase in a RC circuit EEV (without modifying the power source) only affects the unit in whom the modification was made. As in the model the changes in EEV do not affect capacitance, the driving voltage is lower and therefore the charge received. This behavior of the electric model to changes in the EEV is consistent with the results of the mechanical model [14].

Finally, PCV spontaneously compensates for the compliance added to the system by the tubing, which in combined ventilation is expected to contain twice the volume compared to single (conventional) ventilation. This must be manually compensated in VCV mode by adding an extra volume to the total charge delivered by the ventilator. According to the results of the simulations of this study, the most efficient and safest ventilatory mode to use DuplicAR is PCV.

Limitations

The present study has limitations. A first observation is that the capacitance for each capacitor in the electrical model is constant. Thus, we assume that ventilation takes place in the region between the lower and upper inflection points of the compliance curve, where it behaves like a linear function. According to this, capacitance is constant, independent of the voltages applied. In clinical practice this would imply that regardless of the pressure applied, the relationship between a volume and a pressure differential would behave as a constant.This assumption may not be real in mechanical ventilation scenarios. In addition, when the simulation takes place at higher voltages, the gas compression that would occur at these simulated pressures cannot be modeled in the electrical model.

As a second limitation, despite DuplicAR allows for a gradual regulation of the diameter of the inspiratory lines of each patient, it is not possible (with this first version of the device) to make an accurate quantitative measurement of the resistance. The precise titration of the inspiratory resistance of each patient achieved in the electrical model can hardly be transferred to the real model.

In our model, the ability of the ventilator to "pressurize" the system is not a problem, as the voltage or current source has unlimited capacity. In real life, this can be a problem. The ventilator must be able to pressurize a larger system with greater absolute compliance. The electrical model does not consider the greater compliance of a system with more connections. This would produce a "volume steal" in VCV, not contemplated in our simulations.

Finally, the objective of the simulated mechanical and electrical models is to represent, in a reductionist way, the interaction between variables in a combined mechanical ventilation scenario. Such models do not take into account other aspects that should be considered in clinical practice, such as hemodynamic status, quality of gas exchange, and the possibility of inadvertent spontaneous ventilation in patients under combined ventilation.

Conclusions

The electrical computer-based simulations are a safe, pedagogical, and effective tool for understanding combined ventilation. In this model, the DuplicAR system was shown to be effective in precisely controlling the distribution of load between capacitors. Even in scenarios with different capacitance and with disparate end-expiratory voltages.

This, in mechanical ventilation, implies that it is possible to individualize the ventilatory parameters in two patients connected to a single ventilator.

Conflicts of Interest

Lugones I. is the trademark holder and the author of the patent application.

Funding Statement

None.

Acknowledgements

Diego Riva acknowledges the support of the National University of La Plata, CONICET and IP-COVID (Fundación Bunge y Born – Agencia I+D+i) in his projects.

Supplementary Material

Two video clips running the simulations in both PCV and VCV are included as supplementary material. The simulated values are not strictly those used during the experimental protocols. The electrical circuit used and the morphology of the graphs obtained are shown.

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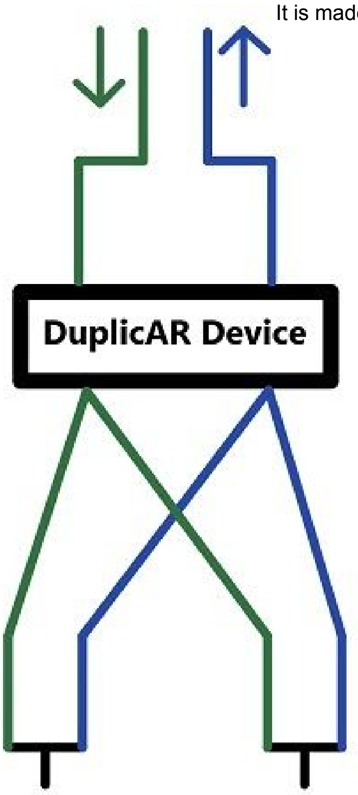

Figure 1. The DuplicAR device: schematic drawing (green: inspiratory lines; blue: expiratory lines; arrows: ventilator ports).

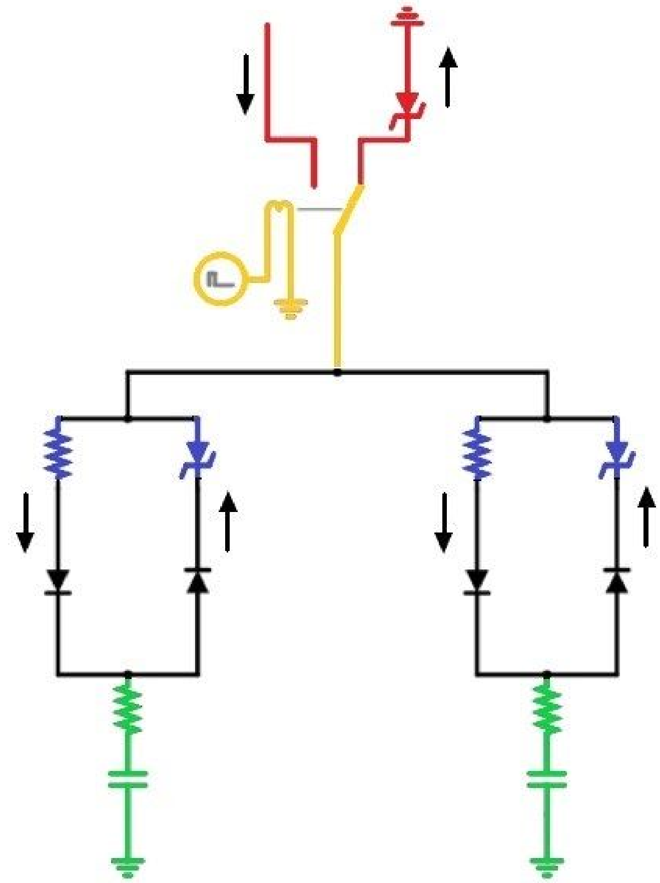

Figure 2. Electrical diagram of combined ventilation: in red, the "ventilator system"; in green, the "subject system"; in yellow, the connection between both systems through the relay; in blue, the DuplicAR system. The black arrows outside the circuit represent the direction of the current.

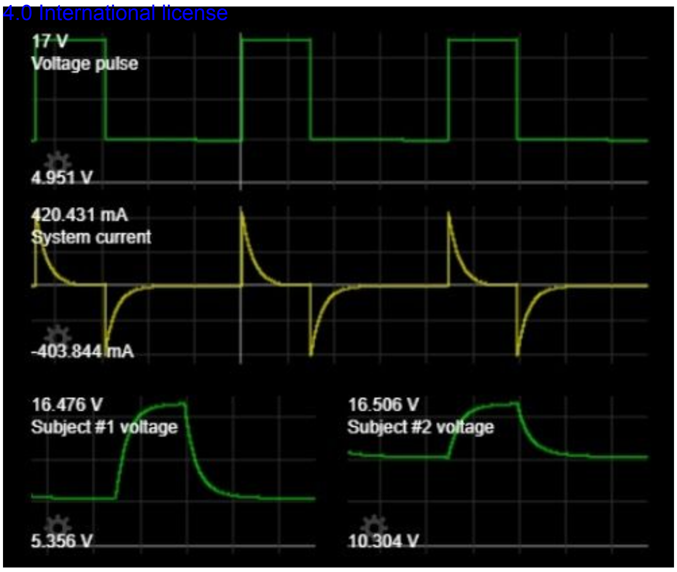

Figure 3. Records in a PCV simulation. All graphs show voltage or current as a function of time. Top: the voltage pulse generated by the "ventilator system". Medium: the current register in the system. Lower: the capacitor voltage that represents each subject, equivalent to the alveolar pressure in mechanical ventilation.

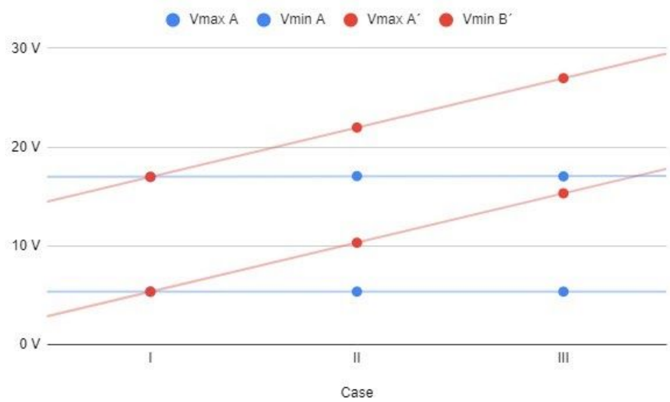

Figure 4. Stage 1 simulation, voltages registers under PCV mode.

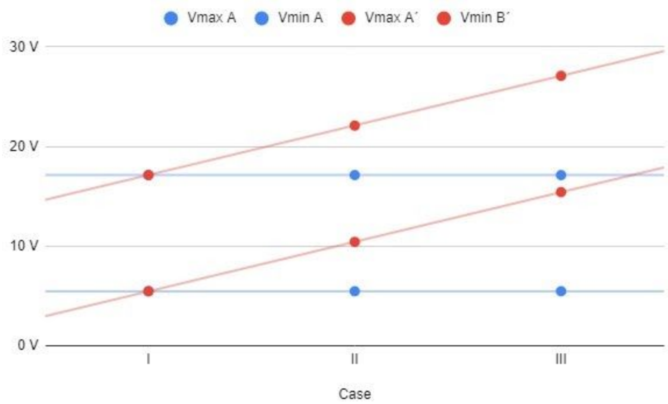

Figure 5. Stage 1 simulation, voltages registers under VCV mode.

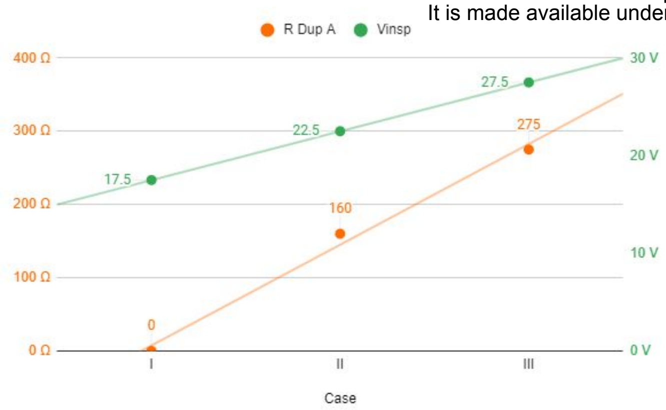

Figure 6. Stage 1 simulation, PCV mode. In green, power source voltage; in orange DuplicAR resistance of one RC circuit.

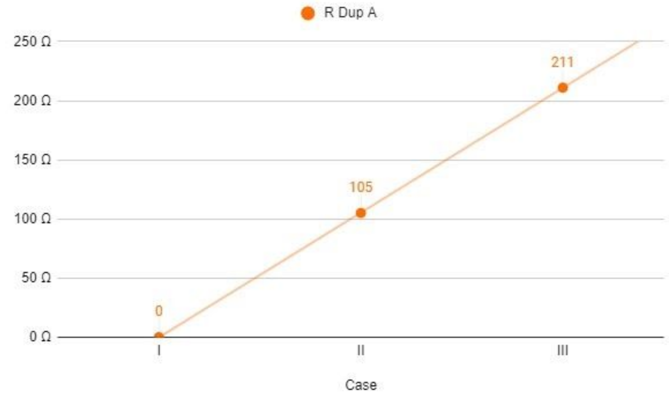

Figure 7. Stage 1 simulation, VCV mode. DuplicAR resistance of one RC circuit in each case.

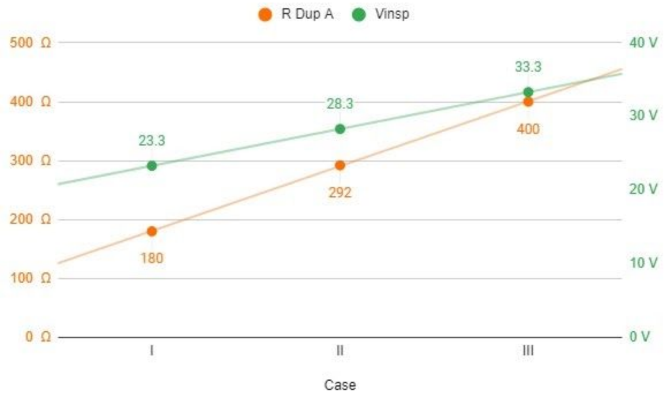

Figure 8. Stage 2 simulation, PCV mode. In green, power source voltage; in orange DuplicAR resistance of RC circuit A.

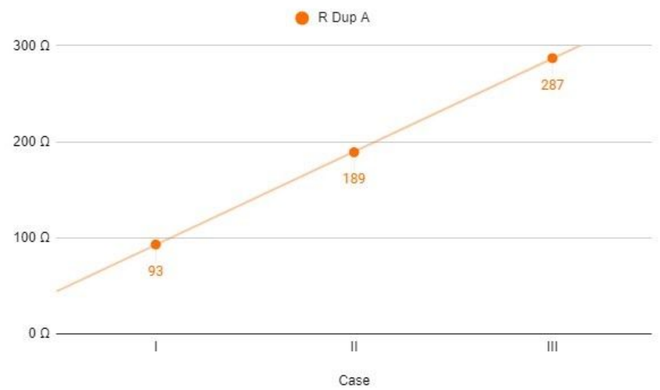

Figure 9. Stage 2 simulation, VCV mode. DuplicAR resistance of RC circuit A in each case.

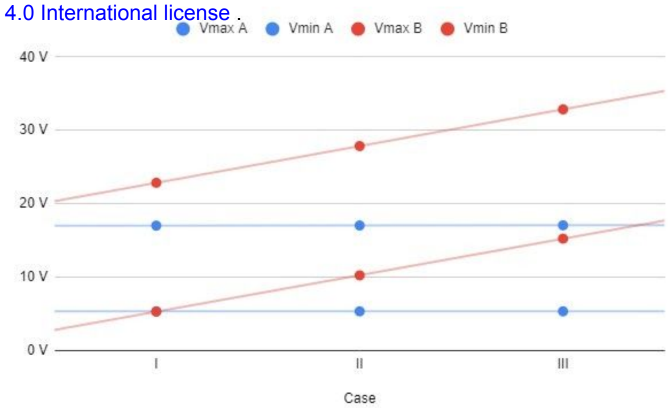

Figure 10. Stage 2 simulation, voltages registers under PCV mode.

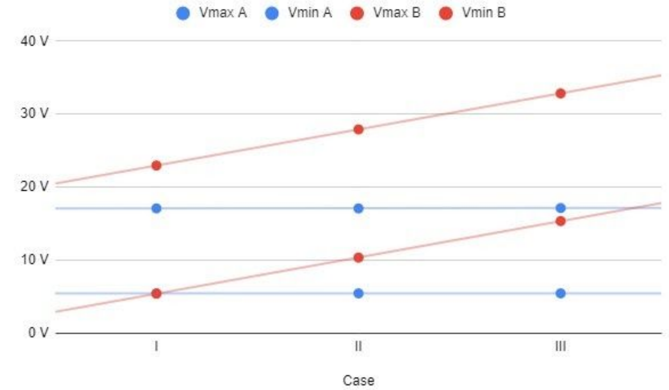

Figure 11. Stage 2 simulation, voltages registers under VCV mode.

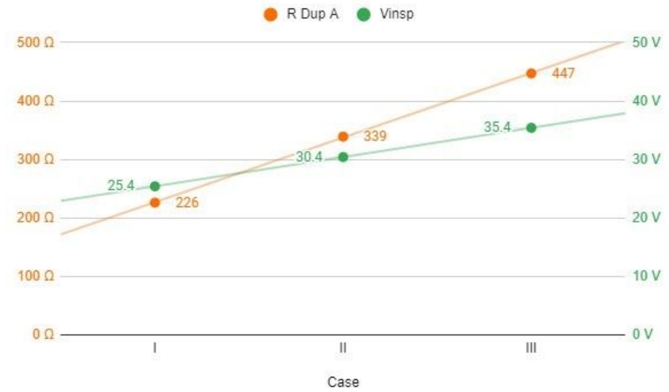

Figure 12. Stage 3 simulation, PCV mode. In green, power source voltage; in orange DuplicAR resistance of RC circuit A.

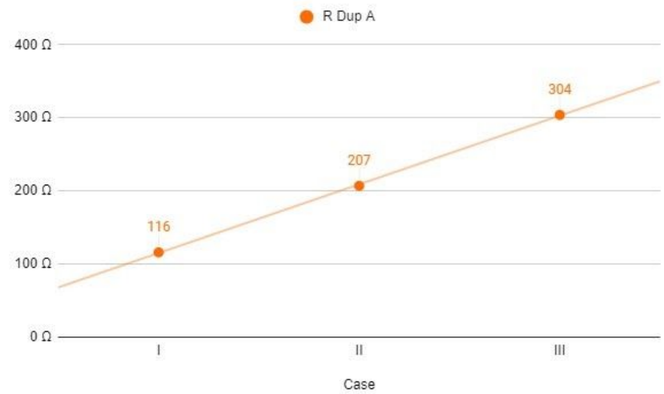

Figure 13. Stage 3 simulation, VCV mode. DuplicAR resistance of RC circuit A in each case.

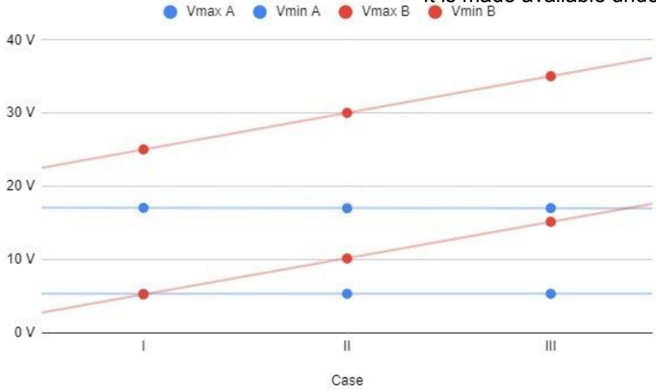

Figure 14. Stage 3 simulation, voltages registers under PCV mode.

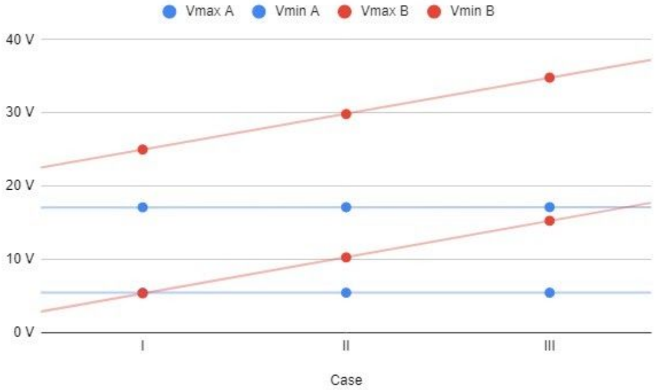

Figure 15. Stage 3 simulation, voltages registers under VCV mode.