AMO - Advanced Modeling and Optimization, Volume 17, Number 2, 2015

Modeling of the Optimized Control Cardioverter

A. A. Saraiva[1,2] F. M. de A. Araújo[2]

GALILEO - CEUMA, Teresina, Piaui, Brazil
LABIRAS - IFPI, Praça da Liberdade, 1597, 64000-040, Teresina, Piaui, Brazil

E-mail: aratasaraiva@gmail.com

Abstract:

The modeling of a single-phase cardioverter to be advantageous to minimize the energy required by the system is shown in this work. The method used to control the optimization is Linear-Quadratic Regulator, which results in better performance in systems that use external power to the control action

1. Introduction

Defibrillation is the application of an electric current in a patient by a defibrillator, an electronic device whose function is the reversal of cardiac arrhythmias by applying an electric current pulse of large amplitude in a short time. Passing through the heart, this current strength a simultaneous contraction of cardiac fibers, enabling the restoration of a normal rhythm.

The defibrillator is made up of two blades connected by cables to a device that transforms electrical energy into electric shocks. The intensity of the shock is adjustable and can reach up to 360 joules (360 W.s). The two blades is discharging shocks on the anterior chest wall.

The pulse of electrical current through the heart promote depolarization (contraction) of a large quantity of fibers were repolarizadas ventricular (relaxed) and prolongs the contraction of which were already committed. If a certain critical mass (75% to 90%) of the fibers simultaneously respond to this forced contraction, when they return to the idle state will be able to respond to the natural pacemaker of the body and with the timing, the pumping is restored.

The concept of defibrillation was introduced in 1899 by Prevost and Batelli after an experience where it was applied a high voltage shock to the heart of a dog with ventricular fibrillation [1].

In 1933, Hooker, Kouwenhoven and Langworthy published numerous successes internal defibrillation in dogs with AC and in 1936 Ferrie successfully performed defibrillation using alternating current directly from the power grid [2].

In 1947, Claude Beck reported the first successful defibrillation in humans, applying an alternating current with a frequency of 60 Hz during a surgery [3].

In later research, Edmark and her colleagues found that the direct current or defibrillation pulse was more effective and with fewer side effects than the alternating current and with the advent of electronics in the 60s, this current, technologically improved, it has become employed on a large scale.

2 Cardioversion

Electrical cardioversion involves the application of a continuous electrical current to shock, synchronized with the electrocardiogram QRS complex (ECG) on the chest to the reversal of cardiac arrhythmias. In defibrillation, the discharge is not synchronized. In these cases the heart is in operation when the discharge is made. The instant that the discharge is made must match the contraction of the ventricles, ie with the QRS complex of the electrocardiogram.

The cardioverter is a device that converts electric alternating current to direct and enables the application, via electrodes (blades) of a given amount of energy (charge) to the patient. Is coupled to an EKG recording system with which it is synchronized, in other words, it must recognize the R wave (or S) of the electrocardiogram and thus prevent discharge of energy in a vulnerable period of the ventricle, which could lead to trigger ventricular fibrillation.

The electric shock mechanism of action is the simultaneous depolarization of excitable cardiac fibers in that moment, which stops the reentry mechanisms responsible for a large number of arrhythmia, allowing the sinus rhythm is reestablished.

In ventricular fibrillation, a greater energy is required to cause a transient inhibition of the heterotopic focus. The suitable density of electrical current delivered through the heart is crucial for the success of cardioversion and depends on the selected energy and thoracic impedance. This is related to the size and the distance between the blades, the pressure exerted on them, the use of gel or saline and respiratory phase (decreases during expiration). The impedance also decreases after repeated shocks (8% in the second shock).

The immediate cardioversion is indicated in the sustained ventricular tachycardia and reentrant arrhythmias accompanied by chest angina, cardiac failure or hypotension.

Arrhythmias by increasing the automation does not respond to cardioversion. In emergency situations, when there is ventricular fibrillation caused by myocardial infarction, electric shock or other cause, immediate cardioversion allows to obtain excellent results. When used early, it is possible reversion of arrhythmia in over 90% of cases.

In other situations, the decision by cardioversion should be made taking an analysis of the clinical condition of the patient as a whole, the prior use of medication (digitalis, antiarrhythmics, etc.), the duration, the significance and the prognosis of arrhythmias, anticoagulation provided or the previous installation of pacemaker.

Defibrillators that have option for cardioversion have attached monitors or have input to EKG signal from an external monitor.

The defibrillators are also used during cardiac surgery with opening and manipulation of the heart. Accordingly, to maintain the physiological circulatory patterns, it uses a cardiopulmonary bypass machine and a blood oxygenator. After the surgery, a direct desfibrilatória jolt to the heart with the internal paddles with low energy stimulates the heart to resume its pumping function of blood.

2.1 Cardiac Electrophysiology

The heart pumping is started by a small electric current pulse, which spreads quickly through the heart, causing your muscles to contract. If all the muscles of the heart contracted at the same time there would be no pumping effect. Then the electrical activity starts at the top of the heart and spreads down and then starts a new round on top again [4].

The heartbeat occurs due to the fact that a group of specialized cells called pacemaker, modify their potential from positive to negative and positive again, quickly. The first electric wave of a contraction of the heart, to be raised on top of the heart, spreads based on the ability of cardiac cells to conduct this electric charge to adjacent cells, causing a chain reaction [5]. It is observed that the electric wave that propagates in the heart is resulting from a negative pulse causing depolarization, and occurs with the influx of positive ions ($Na+$) to the inside of cells, causing the potential outside the cells is more negative. Later, in a process called repolarization, the positive ions ($K+$) flow to the exterior of the cell causing the potential outside the cell stay positive [5].

Specialized cardiac muscle cells, which are self-excitable and impulse conductive, form the electrical conduction system and are divided basically into: sinus or sinoatrial node (NSA), internodais beams, beam of Bachman, atrioventricular node (NAV), His-Purkinje system (bundle of His and Purkinje fibers) (Malmivuo and Plonsey, 1995). As a statement of the system can be observed in Figure 1.

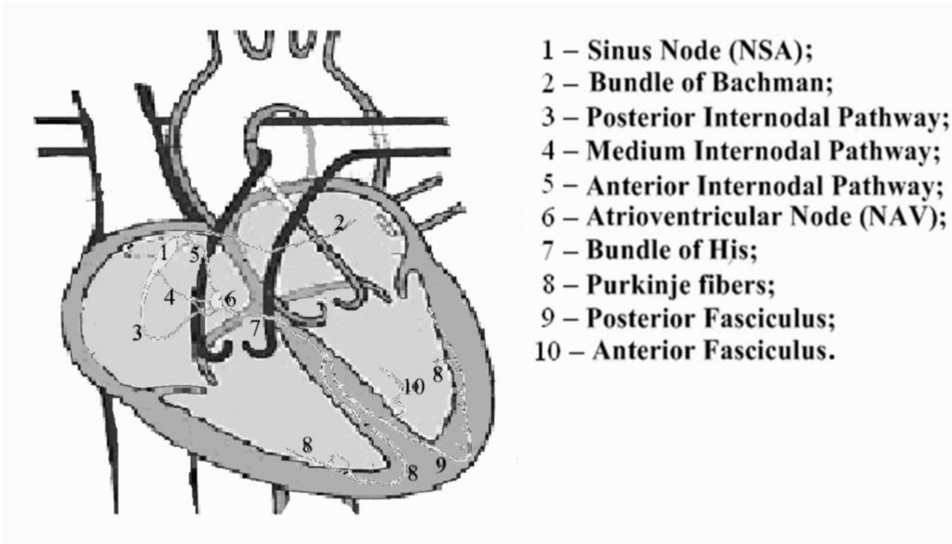

Figure 1. Heart structures involved with the cardiac electrical impulse. Source: Own authorship.

A simplified way to represent the electrophysiology of the heart, causing a relation to that presented in the electrocardiograph:

- The electric stimulus originated in the NSA, which receives the parasympathetic and sympathetic innervation, propagates the electric stimulus by internodais beams being formed basically by two types of special cells: nodal cells (P cells) that are centrally and are responsible for the cardiac automatism with the name P from the fact of being pale (cytoplasm poor in glycogen; transitional cells (T cells)) which involve the P cells and are intermediate, performing the transition with atrial muscle cells. This wave of depolarization, propagated by atria, is represented in the electrocardiogram as P wave [6].
- The internodals beams, in turn, carry out the connection of the NSA with NAV. They are consist of three avenues: earlier, middle and posterior. The earlier takes fiber to the left atrium communicating with the beam of Bachman, driving thrust of the right atrium to the left atrium and leads fibers to the NAV. The middle via average fibers to the left atrium and the NAV. And posterior via leads fibers to the right atrium and NAV [7].
- The electrical impulse then reaches the NAV, where there is little delay in driving (0.1 s). Is located exactly above the insertion of the septal tricuspid valve. Its function is to conduct the electrical impulses to the ventricles through the bundle of His, being divided into three areas: (a) that performs the transition of atrial myocardium with NAV, the compact area and which performs the transition from NAV with the bundle of His (penetrating) [8].
- From the NAV, the electrical impulse follows the interventricular septum, the bundle of His (atrioventricular) that splits into its left and right branches for their respective ventricles. The left bundle branch still forks in subdivisions: anterior and posterior fascicle [9]. The branches of the bundle of His end in numerous fibers called Purkinje fibers, which are distributed widely by the myocardium and determine their depolarization, the endocardium to the epicardial, sparking the ventricular contraction [10].

• The electrical impulse that propagates the NAV to the Purkinje fibers and myocardial cells is represented on the electrocardiogram by QRS complex, which therefore represents ventricular depolarization, which predates their contraction. Being that in Purkinje fibers the electrical impulse is preferred, varying the speeds of the stimulus in the myocardium so that they engage in much the same way and only penetrate to a third of the myocardium (subendocardial fibrosis region) [10].

The spread of electrical impulse through the heart depolarizing and repolarizing the specialized cells generating the action potential is shown in Figure 2. Being the sum of the potential actions of the ECG signal.

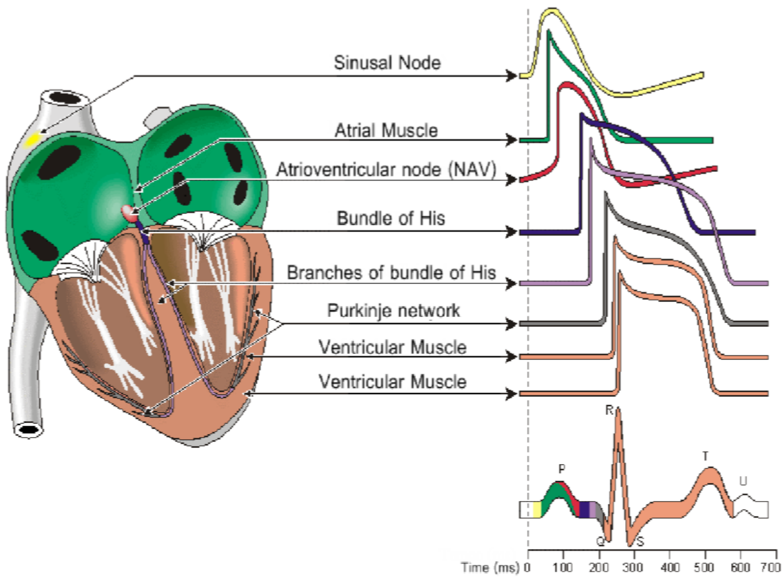

Figure 2. Spread of electrical impulse through the structures of the heart.. Fonte: Modified of the [11].

U wave may exist, a little wave observed after the T wave and before the P wave, representing a late repolarization of the ventricles, not possessing a lot of clinical relevance [11]. The basic properties of myocardial fiber are: Automation, driving, excitation and contraction [12]:

• In automation the heart is able to auto-excite due to the ausence of an external stimulus (nervous or otherwise) to perform the contraction. Is the ability of myocardial fiber has to generate the momentum that determines its contraction [13];

• The driving is the property of electric stimulus conduction from one fiber to other sectors of the myocardium [12];

• In the excitement the heart muscle is able to respond to certain stimuli electrical, mechanical, chemical (adrenaline, acetylcholine) or thermal generating action potentials and causing occurs the proportional contraction to the stimulus [12];

• On contraction when receiving external or internal stimuli myocardial fiber has the ability to contract [12].

2.2 System Leads

Cardiac electrical currents can be sequentially documented by electrocardiogram (ECG), which is up by a galvanometer [14]. To acquire the signals of electrical activity of the heart is needed attracting electrodes on the body surface in standard positions [13].

Electrocardiographic standardized is formed by a line joining two electrodes, corresponding to the record obtained by an electrode placed at any point of body. Usually they are placed on the surface of the chest and limbs, non-invasively. However, there are situations in which use electrodes inside the esophagus (esophageal bypass), within the heart (endocardial derivation) or the surface of the heart (epicardial derivation) [14].

The leads are classified into three types: bipolar, unipolar and precordial. The bipolar leads have been proposed primarily by Einthoven, and measure the difference in potential between the ends (I, II and III), two poles having a same lead. The unipolar leads were proposed by Wilson and feature 3-lead at the ends: aVR, aVL and aVF. And the chest leads (horizontal) from V1 to V6 having to analyze the electrical activity in a number of different angles [10], as shown in Figure 3.

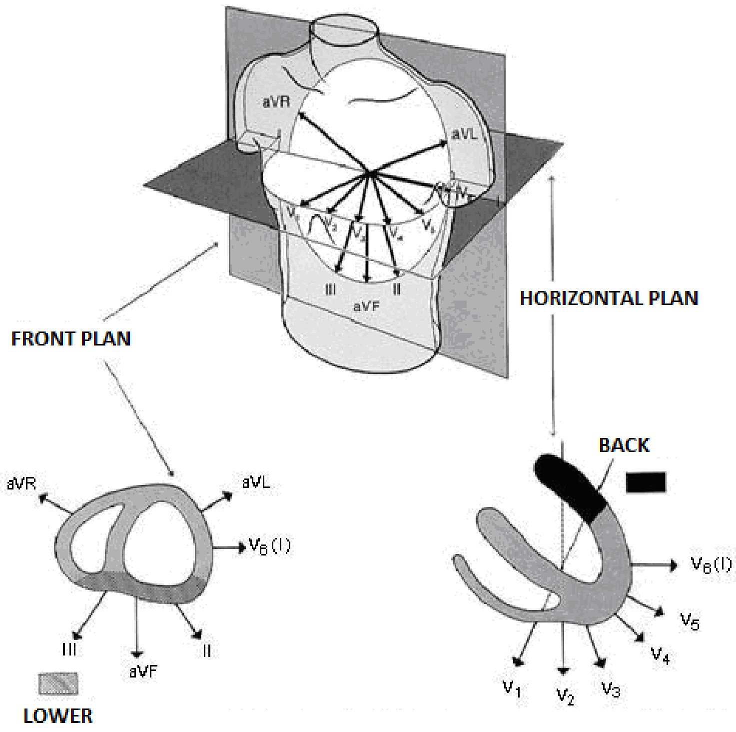

Figure 3. – Heart Derivations.

Source: Modified Malmivuo and Plonsey (1995) [10].

2.3 Cardiac Arrhythmias and Ventricular Fibrillation

Defibrillation, which is the termination of atrial fibrillation requires the re-synchronization of cardiac muscle cells. The electrical defibrillation operates by producing a uniform state of ventricular excitability. The electric shock defibrillation encourages all excitable tissue, leading to refractory state a critical mass of ventricular muscle by stopping fibrillation. Normally, the heart is able to beat synchronously after electrical defibrillation. For an individual cardiac cell, the shock need not be larger than the pulse generated by the pacemaker's own heart. In this sense, a defibrillator is a similar stimulating a heart pacemaker large scale, that should not cause damage to the heart.

To produce the same current density than the natural cardiac pacemaker, defibrillation requires properly positioned electrodes (see Figure 4.1) and large surface by which circulating currents of tens of amps, and most of these chains end up going around and not through the heart [15].

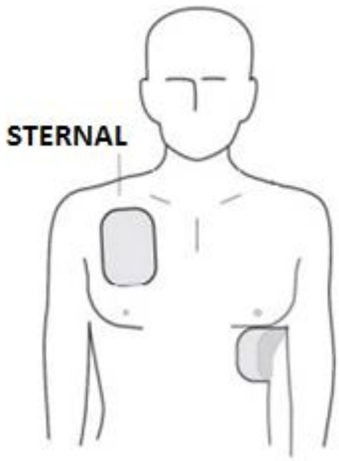

Figure 4. Proper positioning of the electrodes for transthoracic transmission of energy and effective defibrillation

E provided = Estored x[Rp / (Rp + Rd)] (4.1)

where: E = Energy provided provided Earmazenada = Stored energy Rp = patient Resistance Rd = Internal resistance defibrillator

The thoracic impedance adult men undergoing defibrillation in hospitals varies 25-105 ohms, and the mean value of 58 ohms, very close to the commonly employed as a reference value [15]. Studies with various waveforms point to two factors that play a fundamental influence on the outcome of defibrillation, intensity and duration of the shock.

The function that describes the commitments required between intensity and duration to produce an adequate defibrillation current is known as the strength-duration curve, as shown in Figure 5. In general, longer shocks require less current than shorter shocks.

Intensity and duration shocks located in the area above and to the right of the current curve (or above the power curve) have adequate charge to defibrillate while shocks below and to the left, not [16].

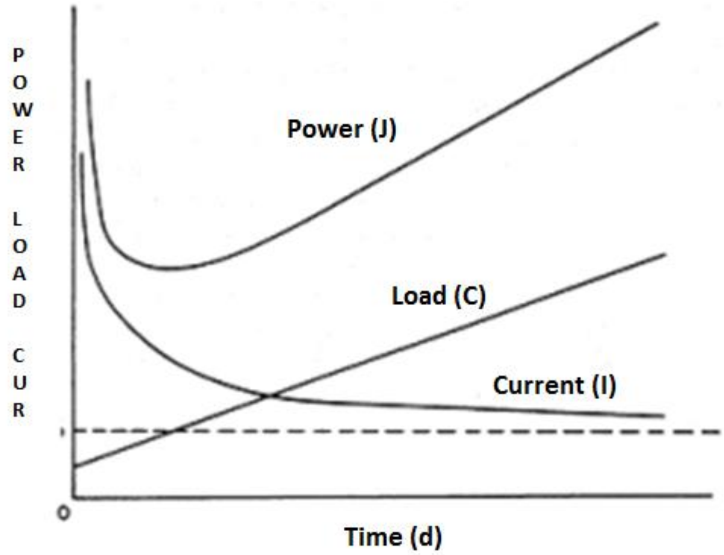

Figure 5. curve intensity-duration for energy load and current (Adapted from BRONZINO, JD, 1995 [16]).

Thus, for the majority of the waveforms, the minimum energy for defibrillation pulse released by 3 to 8 ms duration. In practice, a shock applied to the fixed electrodes on the skin surface of the patient's chest, the durations of the discharge is of the order of 3 to 10 ms and have an intensity of several kilovolts and tens of amperes [16].

2.4 Defibrillators Clinical

The development of defibrillators result of physiological and medical research as well as technological advances in hardware. All clinical defibrillators currently used store energy in capacitors. The energy stored by a capacitor is calculated using the following formula.

$$W = 1/2(CE2) \quad (1)$$

where: W = Stored energy, given in joules C = capacitance, in farads given E = applied voltage on the capacitor, given in volts.

The figure 6 shows the basic diagram of the defibrillator block. Most have one monitor and an integrated synchronizer, constituting what is called the monitor / defibrillator or cardioverter. The monitoring feature increases the diagnostic capability of fatal arrhythmias, especially as the electrocardiogram is monitored through the same electrodes that are used to deliver a shock.

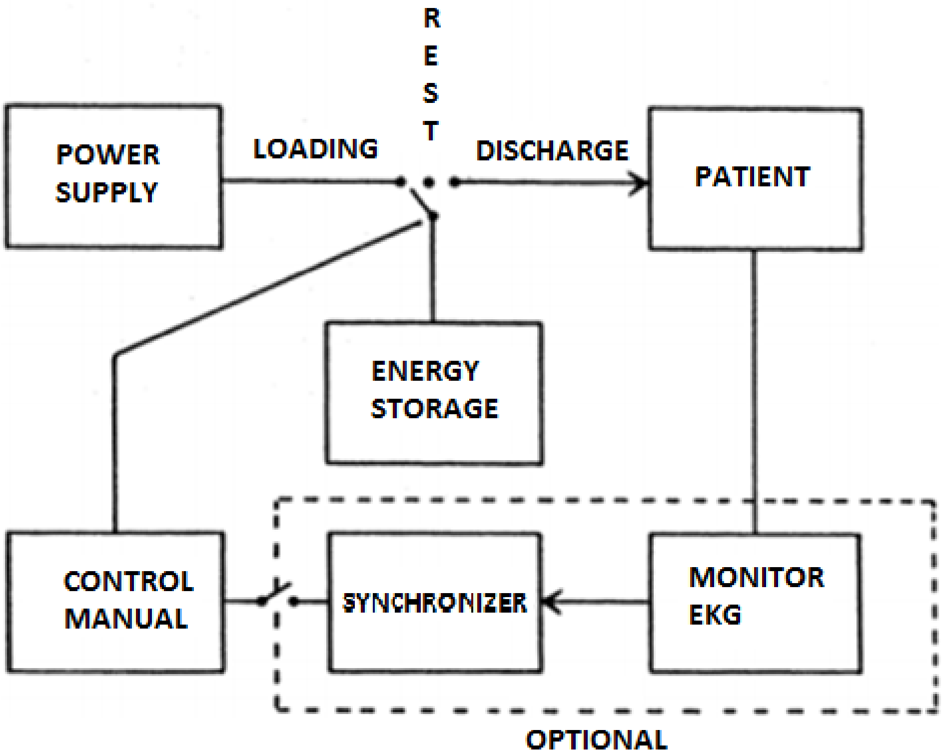

Figure 6. Diagram of a basic block defibrillator (Adapted from BRONZINO, JD, 1995 [16]).

2.5 Single-phase defibrillators

Most existing defibrillators on the market apply impact by means of damped sine wave produced by an LCR circuit or a truncated exponential wave form, as shown in the block diagrams of Figure 7. Defibrillators can deliver up to 360 joules of energy to the patient, within a range of values that starts at 5 joules. Thus, the device allows you to treat the full range of patients, from pediatric to adult obese. Whereas the patient's resistance varies within a range between 25 and 105 ohms and is part of the discharge RLC circuit, and the duration of the damping pulse energy also vary. The higher the patient's resistance, the longer will be damped and the discharge pulse [16].

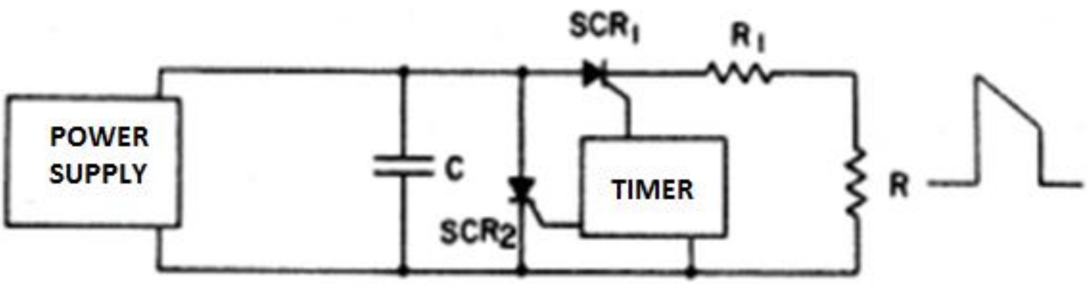

Figure 7. Defibrillator type form of truncated wave (Adapted from BRONZINO, JD, 1995 [16])

The figure 8 shows the form of damped sinusoidal monophasic, with its respective discharge circuit. It should be noted that in practice the damped sinusoidal wave single phase, there is a small polarity reversal at the end of the cycle due to the typical response of this type of circuit (RLC) [17]. A power source which loads the capacitor bank when the loading switch is closed. When the capacitors are charged, the load switch is opened and the discharge switch is closed The capacitor performs a quick and intense discharge of energy stored in the patient's chest through the inductor L. This creates a wave of Lown.

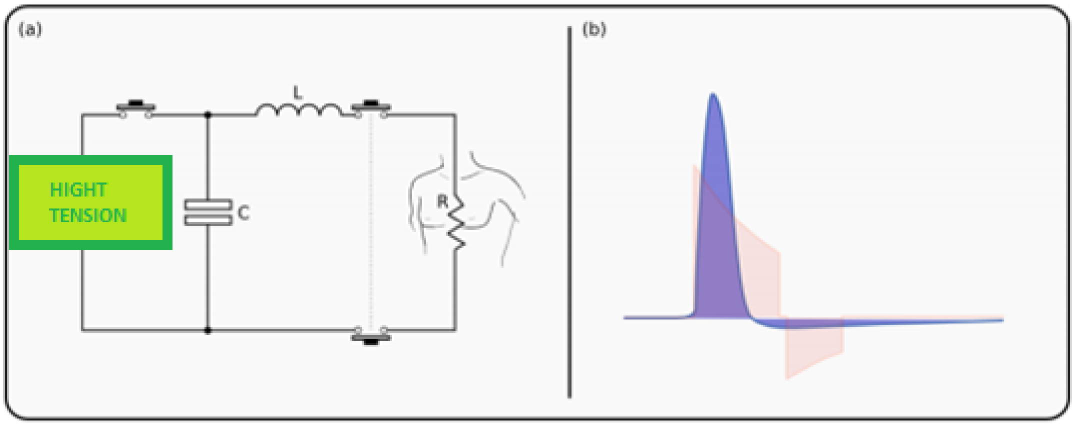

Figure 8. - Sine monophasic defibrillation pulse cushioned. In (a), RLC discharge circuit formed by the capacitor (C), inductor (L) and the patient (R), and it represents the resistive component of the circuit; in (b) defibrillation pulse measured in the patient, where the abscissa axis represents the pulse duration time and the ordinate axis represents the current flowing through the patient (Adapted from American Heart Association Guidelines for Cardiopulmonary Resuscitation, Field et al , 2010 [18]).

Main characteristics of the single-phase technology: - use of electro-mechanical components for switching, resulting in lower reliability; - Voltage levels and higher current; - Does not measure transthoracic impedance; - Valid therapeutic accuracy for transthoracic impedance of 50 ohm; - Larger physical dimensions and weight of the equipment.

Ideally, defibrillators should have sensitivity and specificity of 100%. However, this is not the reality because there is a balance (commitment) between sensitivity and specificity.

The performance evaluation is determined by regulations issued by the American National Standards Institute / Association for the Advancement of Medical Instrumentation, ANSI / AAMI (see below the list of applicable standards) [17].

According to these rules, the sensitivity for recognition of ventricular fibrillation at an amplitude of 200 microvolts or more should be greater than 90% in the absence of artifact.

2.6 System Modeling RLC Proposed

The proposed LCR system is presented in Figure 9. Since the state variables are the inductor current (i_L) is the tension on the capacitor (v_c). The output variable is the voltage across the capacitor.

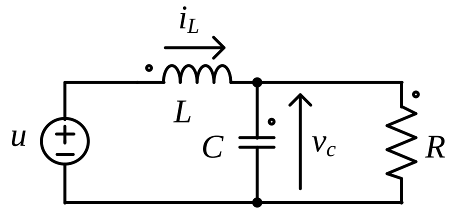

Figure 9. Proposed LCR system

where the specified values are:

$R = 50\Omega$
$C = 220\mu F$
$L = 886\mu H$
v_c (reference) = $50V$
u: supply voltage of a defibrillator. Possible values 0-100.

In order to obtain the model shown at (17).

$$\begin{bmatrix} \dot{v_c} \\ \dot{i_L} \end{bmatrix} = A \begin{bmatrix} v_c \\ i_L \end{bmatrix} + Bu \qquad (2)$$
$$y = C \begin{bmatrix} v_c \\ i_L \end{bmatrix} + Du$$

Circuit is obtained:

$$-v_c + Ri_R = 0$$

where $i_R = i_L - i_C = i_L - C\dot{v_c}$

$$-v_c + R(i_L - C\dot{v_c}) = 0 \Rightarrow \quad \dot{v_c} = -\frac{1}{RC}v_c + \frac{1}{C}i_L$$

Another mesh obtained:

$$-u + L\dot{i_L} + v_c = 0 \Rightarrow \quad \dot{i_L} = \frac{-1}{L}v_c + \frac{1}{L}u$$

The output is the voltage on the capacitor v_c. System obtained:

$$\begin{bmatrix} \dot{v_c} \\ \dot{i_L} \end{bmatrix} = \begin{bmatrix} -\frac{1}{RC} & \frac{1}{C} \\ -\frac{1}{L} & 0 \end{bmatrix} + \begin{bmatrix} 0 \\ \frac{1}{L} \end{bmatrix} u \qquad (3)$$

$$y = \begin{bmatrix} 1 & 0 \end{bmatrix} \begin{bmatrix} v_c \\ i_L \end{bmatrix}$$

2.6.1 **Model System Enhanced**

To guarantee the zero error, is needed an integrator in error.

$$e = r - y = r - Cx \qquad (4)$$

$$\dot{\int} edt = r - Cx$$

The enlarged model of the system is given by:

$$\begin{bmatrix} \dot{v_c} \\ \dot{i_L} \\ \dot{\int} edt \end{bmatrix} = \begin{bmatrix} A & 0 \\ & 0 \\ -C & 0 \end{bmatrix} \begin{bmatrix} v_c \\ i_L \\ \int edt \end{bmatrix} + \begin{bmatrix} B \\ 0 \end{bmatrix} u + \begin{bmatrix} 0 \\ 0 \\ 1 \end{bmatrix} r \qquad (5)$$

Performing the substitutions:

$$\begin{bmatrix} \dot{v_c} \\ \dot{i_L} \\ \dot{\int} edt \end{bmatrix} = \begin{bmatrix} -\frac{1}{RC} & \frac{1}{C} & 0 \\ -\frac{1}{L} & 0 & 0 \\ -1 & 0 & 0 \end{bmatrix} \begin{bmatrix} v_c \\ i_L \\ \int edt \end{bmatrix} + \begin{bmatrix} 0 \\ \frac{1}{L} \\ 0 \end{bmatrix} u + \begin{bmatrix} 0 \\ 0 \\ 1 \end{bmatrix} r \qquad (6)$$

Thus, one obtains increased system model:

$$x_{aum} = \begin{bmatrix} v_c \\ i_L \\ \int edt \end{bmatrix} ; A_{aum} = \begin{bmatrix} -\frac{1}{RC} & \frac{1}{C} & 0 \\ -\frac{1}{L} & 0 & 0 \\ -1 & 0 & 0 \end{bmatrix} ; \; B_{aum} = \begin{bmatrix} 0 \\ \frac{1}{L} \\ 0 \end{bmatrix} ; \; C_{aum} = \begin{bmatrix} 1 & 0 & 0 \end{bmatrix} ; B_r = \begin{bmatrix} 0 \\ 0 \\ 1 \end{bmatrix}$$

2.6.2 **Closed-Loop System**

Closed loop, it has to be $u = -k \cdot x_{aum}$. Thus, the system model is given by:

$$\dot{x_{aum}} = A_{aum} x_{aum} - B_{aum} k \cdot x_{aum} + B_r r \qquad (7)$$

Obtaining:

$$\dot{x_{aum}} = (A_{aum} - B_{aum} k) x_{aum} + B_r r \qquad (8)$$

Since the output array is constant:

$$y = C_{aum}x_{aum} \tag{9}$$

2.6.3 **Controllability**

The numerical value obtained for the matrices is given by:

$$A_{aum} = \begin{bmatrix} -90{,}0 & 454{,}55 & 0 \\ -1128{,}7 & 0 & 0 \\ -1 & 0 & 0 \end{bmatrix} \tag{10}$$

$$B_{aum} = \begin{bmatrix} 0 \\ 1128{,}7 \\ 0 \end{bmatrix}$$

The controllability matrix is calculated by:

$$\mathcal{C} = [B_{aum} \quad A_{aum}B_{aum} \quad A_{aum}^2B_{aum}] \tag{11}$$

If the matrix display full rank lines, then the system is controllable. The controllability matrix is given approximately by:

$$\mathcal{C} = \begin{bmatrix} 0 & 5{,}13 \cdot 10^6 & -0{,}4664 \cdot 10^9 \\ 1{,}128 \cdot 10^3 & 0 & -5{,}7904 \cdot 10^9 \\ 0 & 0 & -0{,}0051 \cdot 10^9 \end{bmatrix} \tag{12}$$

The rank of the matrix is complete (rank = 3). Thus, the system is controllable.

2.6.4 **LQR Controller Design Applied to the LCR System**

For project LQR controller, the values of Q matrix and constant R are obtained empirically until the driver reaches the specifications.

For the controller design, the following conditions were used:

- No over-elevation (saturation) in the *u* control action.
- Controller gain and reduced control efforts;
- Obtained maximum dynamic response.

Q = [1 0 0;0 100000 0; 0 0 5000000];
R = [800];

Based on the matrices Q and R were determined Kn gains and system poles:

K = [-0,0223 11,1723   -79,0569];
Poles = [-505 -12195].

2.7 **Simulação Do Sistema RLC**

The LQR controller design was made so that the output of the RLC system keep in 50V considering transient response in nominal conditions less than 60 ms and low control effort. The controlled circuit shown in **Erro! Fonte de referência não encontrada.** was compared with a circuit operating in open loop. The following tests were conducted to verify the driver's behavior:

- Response system to nominal conditions;
- Reduction of 50% of the rated load;
- Increase of 100% load,
- Variation in the controller reference signal.

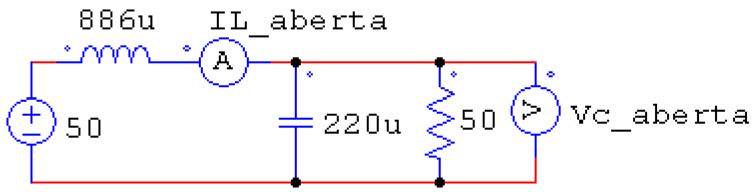

Figure 10. System simulated open loop

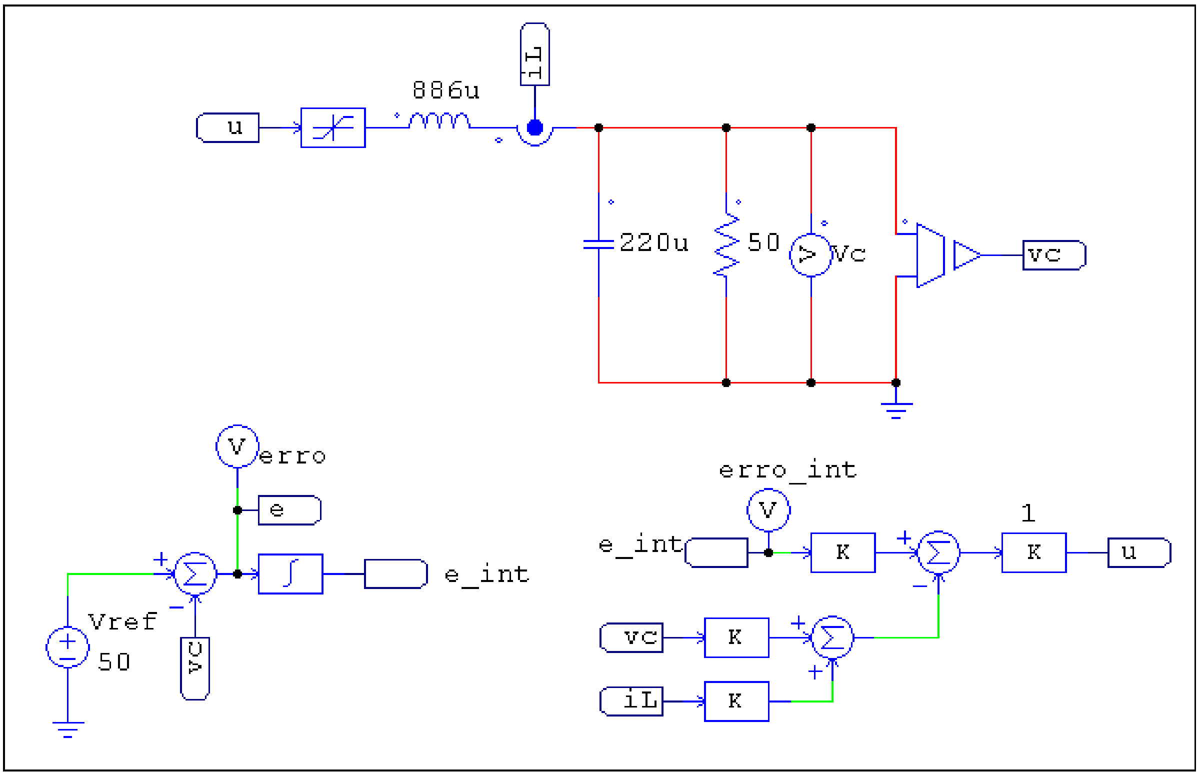

Figure 11. Simulated system with controller

2.7.1 **Simulation System for Nominal Conditions**

The designed system was simulated in Matlab and PSIM software. The system response obtained from software MATLAB is shown in Figure 12, and u resulting control system input is shown in Figure 13. The system frequency response in open loop and under the control action is shown in Figure 14 via its Bode diagram.

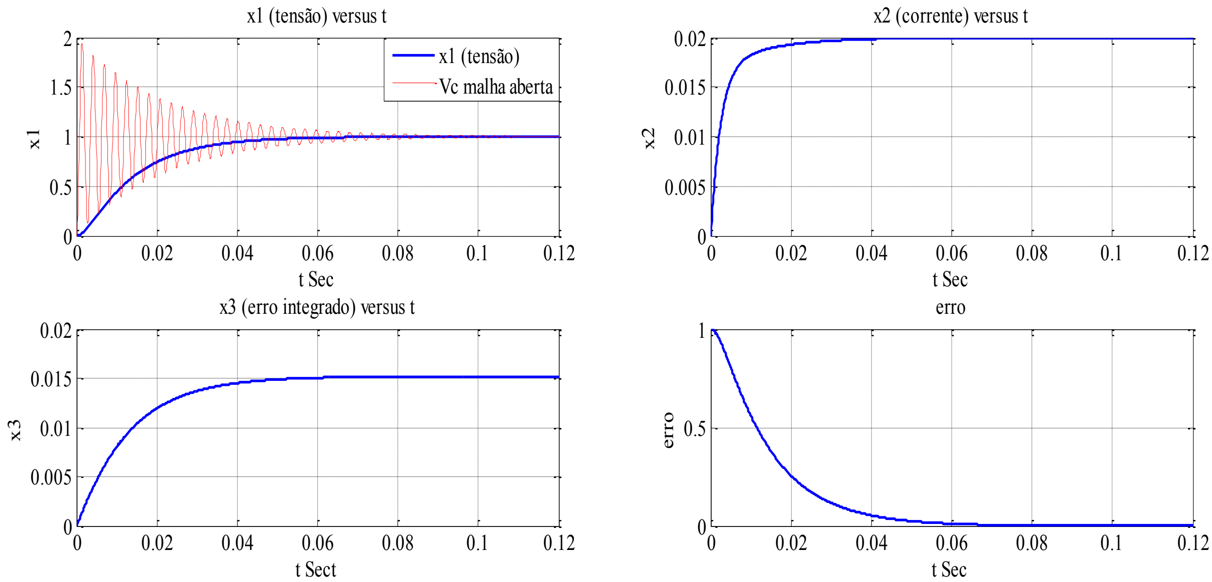

Figure 12. Response of the simulated system in Matlab

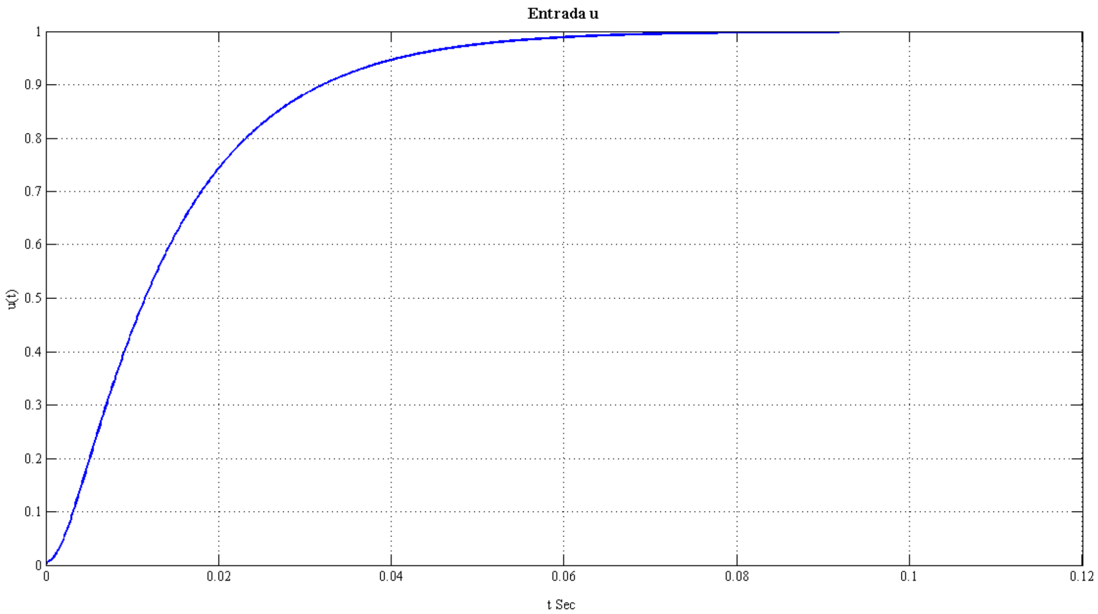

Figure 13. Input *u* resulting from the control system

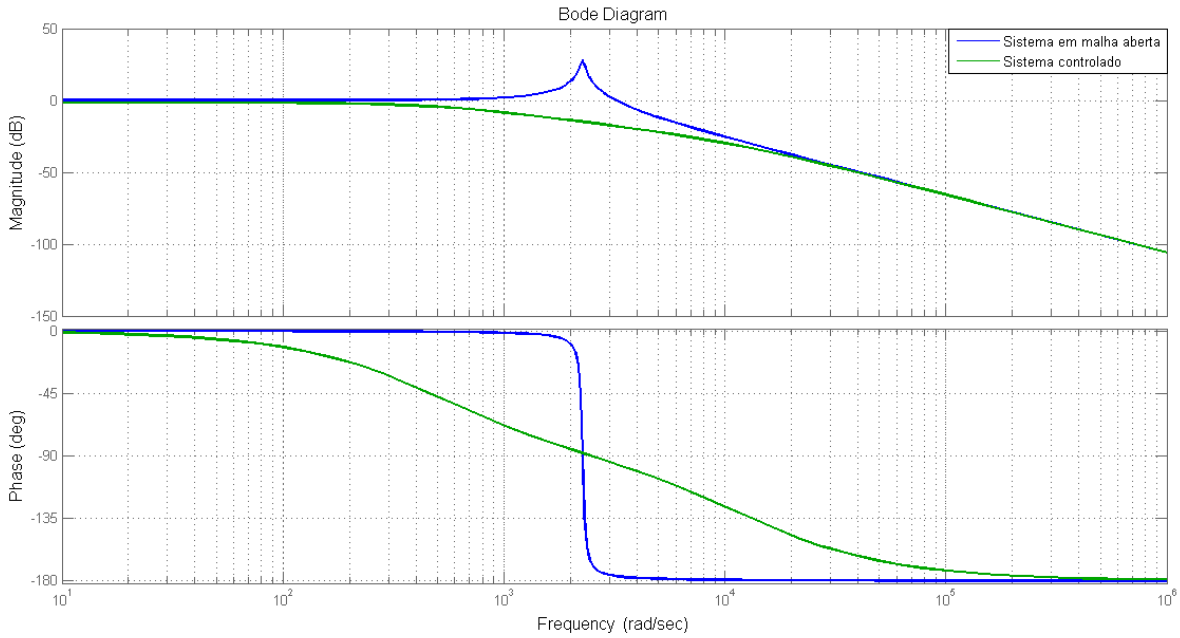

Figure 14. Bode diagram *Y(s)/U(s)* of the system open loop (blue) and controlled (green)

The response of the simulated system in PSIM software open loop is displayed in Figure 15.

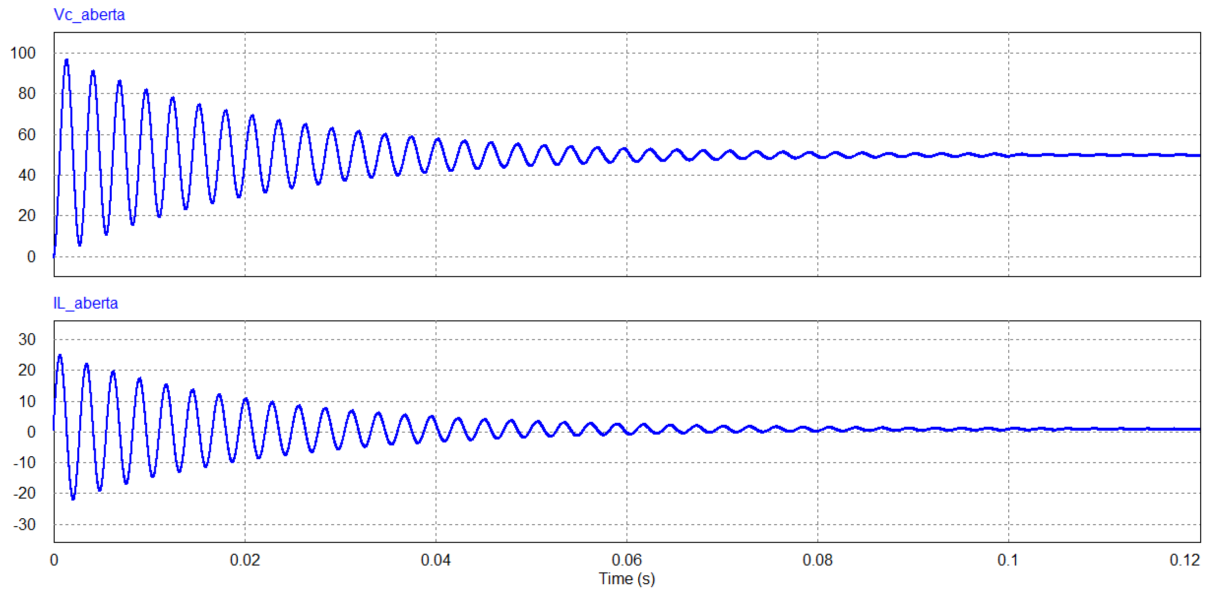

Figure 15. Response LCR open-loop system

Figure 16 shows the response of the controlled system. Figure 17 shows the error response of the designed system and the integrated error.

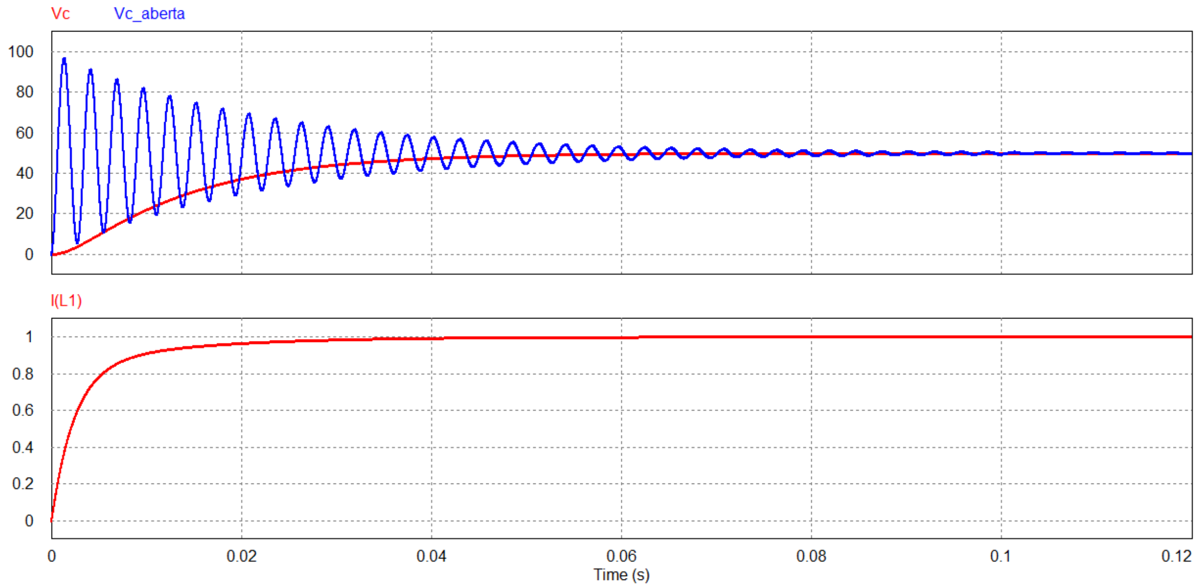

Figure 16. Comparison of the response of the system in open loop (blue) and the controlled system (red)

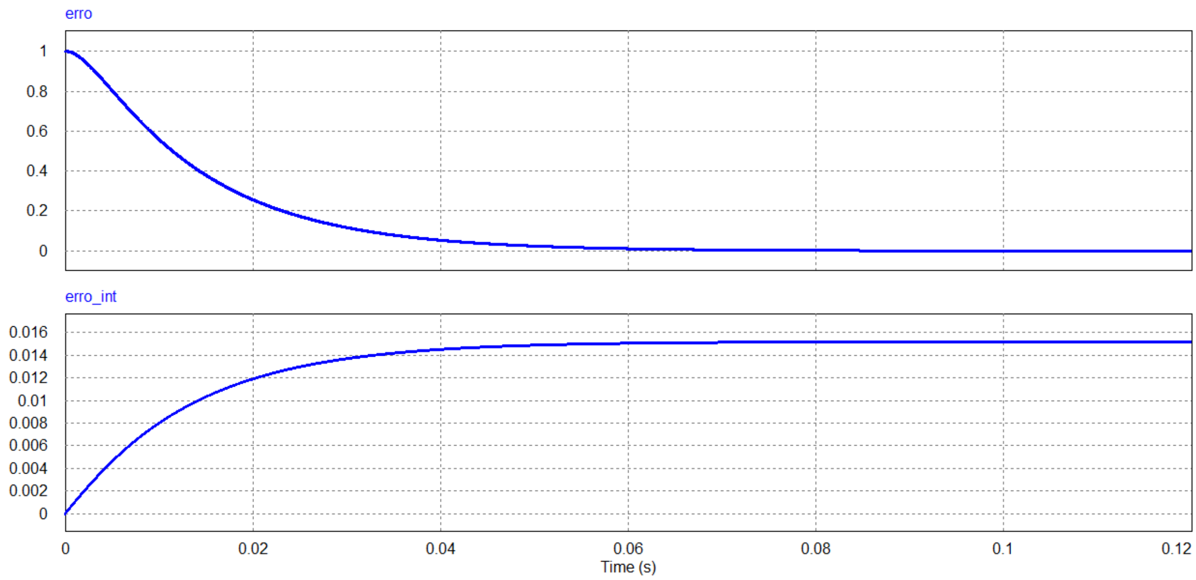

Figure 17. Percentage error and integrated controlled system error

2.7.2 **Simulation: Response to Load Reduction**

The system was simulated for a load reduction to 50% of nominal, and evaluated the possible saturation of the same, which was not detected for the controller designed for 100V input voltage levels and output voltage of 50V. The simulated circuit for evaluating the response to load reduction is shown in Figure 18. The system response to a load change is shown in Figure 19.

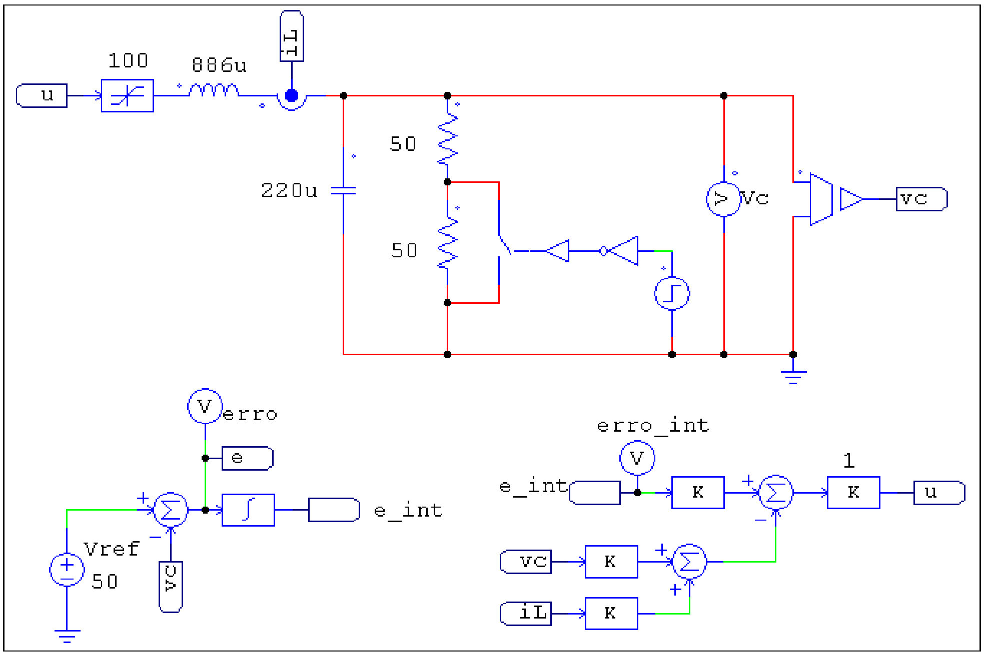

Figure 18. Evaluating the simulated circuit to the load reduction system

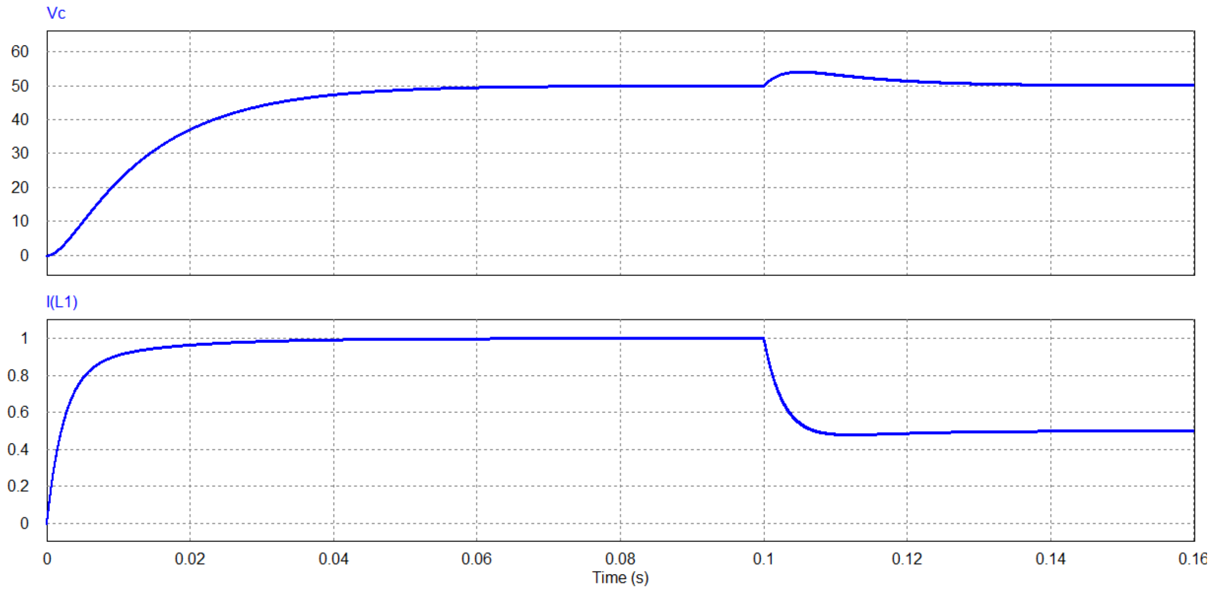

Figure 19. System Response to load reduction

2.7.3 **Simulation: Response to Charge Increase**

The system was simulated for a load increase to 100% of nominal, and evaluated the possible saturation of the same, which was not detected for the controller designed for 100V input voltage level and 50V output. The simulated circuit for evaluating the response to the load increase is shown in Figure 20. The system response to a load change is shown in Figure 21.

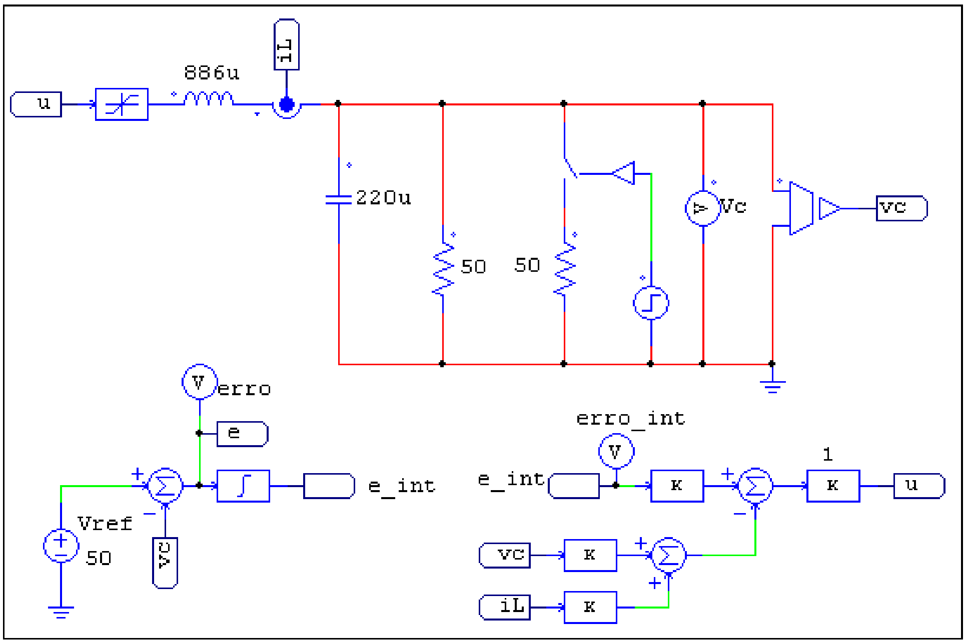

Figure 20. Simulated circuit to increased load evaluation in the system

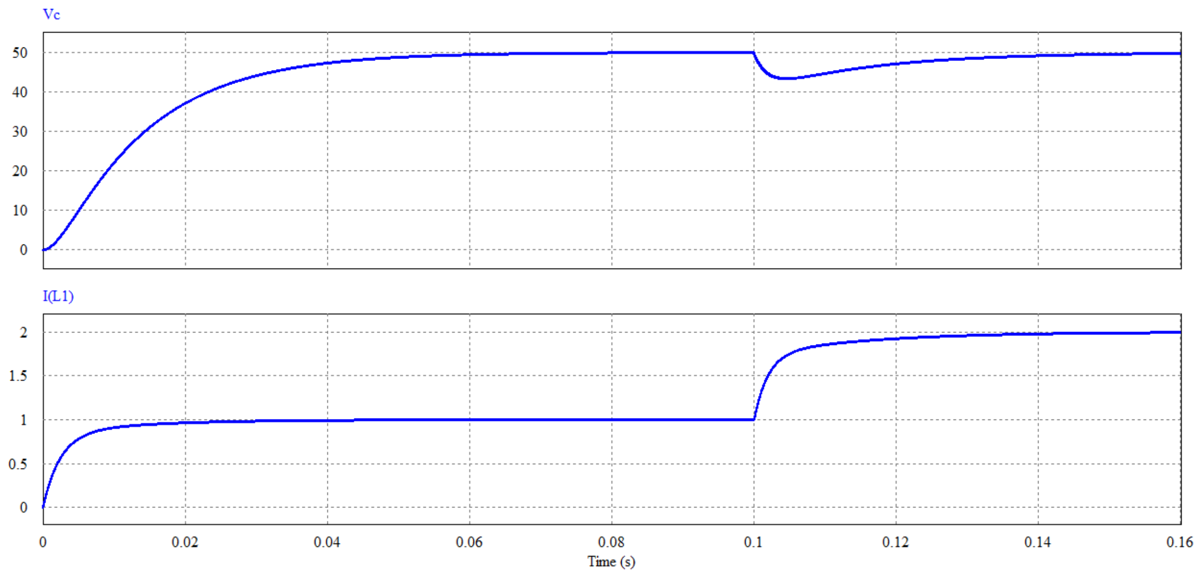

Figure 21. System response to increased load

2.7.4 **Simulation: Response to Change In Reference**

The system was simulated for a variation of the reference signal by increasing the reference signal 50V to 70V, in order to determine the same response behavior. The simulated circuit for evaluating the response to the variation of the reference signal is shown in Figure 22. The system response to a variation in the reference is presented in Figure 23.

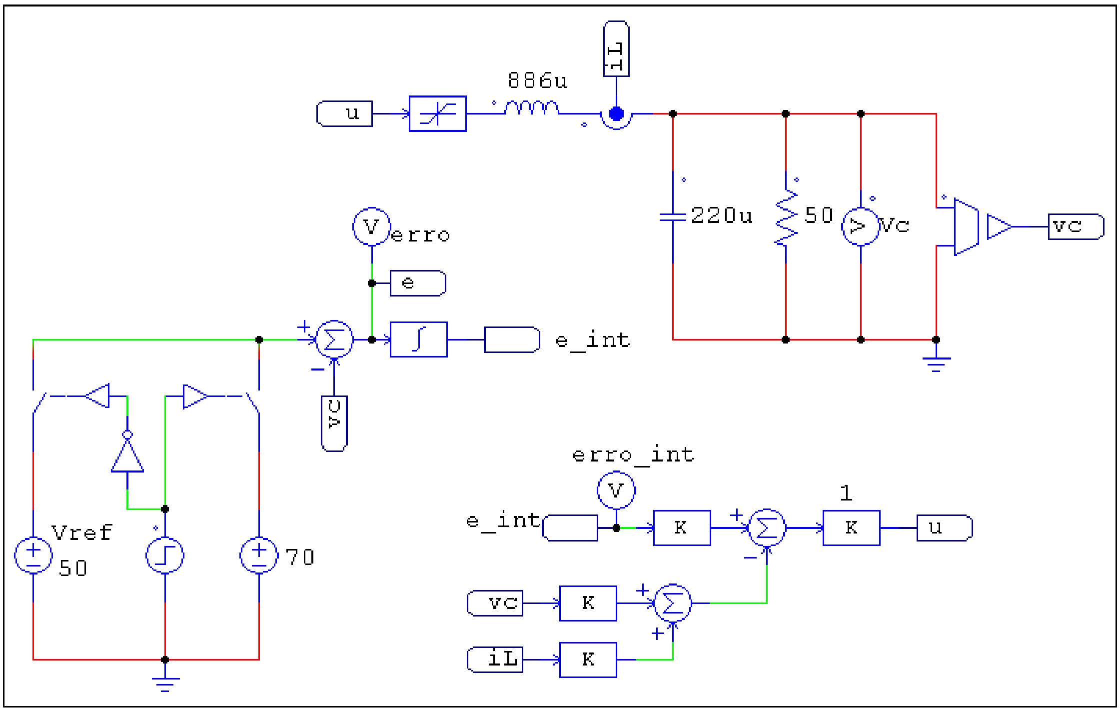

Figure 22. Simulated circuit to increased load evaluation in the system

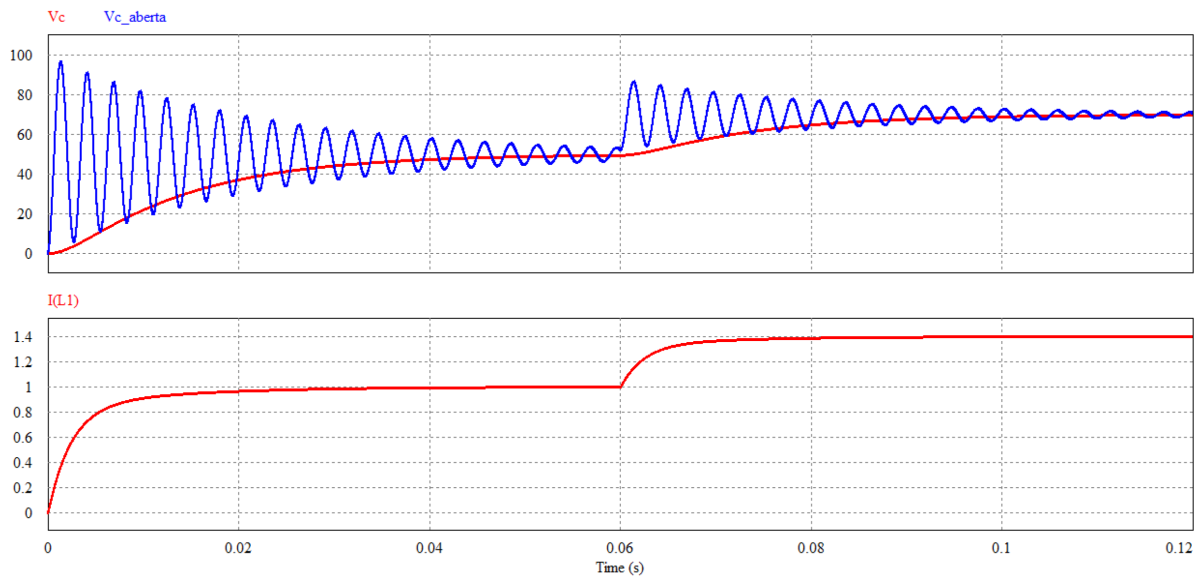

Figure 23. Response in open loop (blue) and closed loop (red)

2.8 **Calculation of Discharge Time**

In the case of a system of single-phase loads ranging from 0 to 360 Joules. It has been that: $360 \quad Joules = 360 \quad W.s$. And the calculation of power is given by equation 13

$$Pot = V^2 * R = 100^2 * 50 = 5 * 10^5 \qquad (13)$$

To obtain the discharge time follows equation 14.

$$
\begin{aligned}
360 &= Pot * t \\
360 &= 5*10^5 * t \\
t &= 7.2 \ \ ms
\end{aligned}
\tag{14}
$$

As the discharge time can vary from 3 to 8 milliseconds, it is observed that the time period is prescribed by the standard.

3 Conclusion

There are various techniques to develop the control systems by state feedback, this study applied the method of Linear Quadratic Regulator - LQR, in which the method is to obtain an optimal control in order to minimize the energy of the actuator on the plant to the desired response . However, the system's response depends on the projected values for the matrices Q and R, which determine the relative importance of the error and the amount of energy required in the control process. Such matrices are defined empirically, therefore, are subject to different answers, which will be as good as the higher the range of tested values, thereby set the configuration that best fits for a given project.

The controller design applied to the defibrillator aimed to get an answer to take the output voltage level to the desired value with reduced energy demand and rapid response of the system states. Therefore, searched an answer that was faster than the natural response of the system operating in open loop, but with possible gains to be implemented.

With the study of LQR controller can highlight the advantages of minimizing the energy demanded by the system, resulting in better performance on systems using external power to the control action. The disadvantage can cite the limitation of technique related to the random setting of the controller gains, making it difficult to define the optimal condition gains.

References:

1. Prevost JL, Batelli F: *Sur quelques effets des descharges electriques sur le coeur des mammiferes.* Acad. Sci. Paris, FR.: 1899; 129:1267–1268.
2. Hooker DR, Kouwenhoven WB, Langworthy OR. *The effect of alternating electrical currents on the heart.* Am J Physiology. 1933; 103:444.
3. Beck CS, Pritchard WH, Feil HS. *Ventricular fibrillation of long duration abolished by electrical shock.* JAMA.1947; 135:985–986.
4. Guyton, A.C. (1992) “Tratado de Fisiologia Médica” 8ª ed. Rio de Janeiro: Guanabara Koogan, 864p. Hoogkin, A.L., Huxley, A.F. (1952) *A Quantitative Description of Membrane Currents and its Application to Conduction and Excitation in Nerve*, Journal of General Physiology, pp. 117-500.
5. Lewis, T., Oppenheimer, B.S., Oppenheimer, A. (1910) *Site the Origin of the Mammalian Heart Beat: the Pacemaker in the Dog*, Heart, pp. 2-147.

6. James, T.N. (1963) *The Connecting Pathways Between the Sinus Node and the AV Node and Between the Right and Left Atrium in the Human Heart,* American Journal of Cardiology, pp. 66-498.
7. Carvalho, A.P., Hoffman, B. F., Carvalho, M.P. (1969) *Two Components of the Cardiac Action Potentials, Voltage Time Course and the Effect of Acelylcholine on Atrial and Nodal Cells of the Rabbit Heart,* Journal of General Physiology, pp. 54-607.
8. Hecht, H.H. (1973) *Atrioventricular and Intraventricular Conduction - Revised Nomenclature and Concepts,* American Journal of Cardiology, pp. 31-232.
9. Davies, M.J. (1971) *Pathology of Conducting Tissue of the Heart*, London Butterworths.
10. Malmivuo, J., Plonsey, R. (1995) *Bioelectromagnetism - Principles and Applications of Bioelectric and Biomagnetic Fields*, Oxford University Press, New York.
11. Machay, L. (2011) *Estudo de Caso Paciente com Distúrbio Cardíaco Denominado BAVT (Bloqueio Átrio Ventricular Total),* Dados colhidos no Hospital na unidade de terapia intensiva, Faculdade de Enfermagem Luiza de Marillac. União Social Camilliana. Rio de Janeiro.
12. Hoffman, B.F. (1960) *Cranefield, P. - Eletrophysiology of the Heart*, New York, McGraw-Hill.
13. Guimarães, J.I., Moffa, P.J., Uchida, A.H., Barbosa, P.B. (2003) "Normatização dos Equipamentos e Técnicas para a Realização de Exames de Eletrocardiografia e Eletrocardiografia de Alta Resolução" 58º Congresso da Sociedade Brasileira de Cardiologia, Arquivos Brasileiros de Cardiologia, v.80, n°.5 São Paulo.
14. Rieira, A.R., Filho, C.F., Subner, S., Schapachnik, E., Uchida, A. ; Moffa, P.J., Zhang, L., Luana, A.B. (2008) "Wellens syndrome associated with prominent anterior QRS forces: an expression of left septal fascicular block" Journal of Electrocardiology, v. 41, pp. 671-674.
15. Cook A. M., Webster J. G. Therapeutic Medical Devices Application and Design. Prentice-Hall. 1982.
16. Bronzino, J. D. The Biomedical Engineering Handbook. CR Press. 1995.
17. *Association for the Advancement of Medical Instrumentation. Cardiac defibrillator devices*. ANSI/AAMI DF80:2003. Arlington (VA): AAMI, 2003. American National Standard.
18. Field HM, Hazinski MF, Sayre MR, Chameides L, Schexnayder SM, Hemphill R, Samson RA, Kattwinkel J, Berg RA, Bhanji F, Cave DM, Jauch EC, Kudenchuk PJ, Neumar RW, Peberdy MA, Perlman JM, Sinz E, Travers AH, Berg MD, Billi JE, Eigel B, Hickey RW, Kleinman ME, Link MS, Morrison LJ, O'Connor RE, Shuster M, Callaway CW, Cucchiara B, Ferguson JD, Rea TD and Hoek TLV. Part 1: Executive Summary: 2010 *Cardiopulmonary Resuscitation and Emergency Cardiovascular Care Circulation* 2010;122;S640-S656 DOI: 10.1161/CIRCULATIONAHA.110.970889