MODELLING OF TRANSTHORACIC ELECTRICAL DEFIBRILLATION WAVEFORMS

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Abstract

Recent investigations related to designing of implantable defibrillators led to reconsideration of heart electrophysiology mechanisms of fibrillation and defibrillation. The data obtained raised questions on the efficiency and security of defibrillation pulses and resulted in reassessment of existing and advancing new types of waveforms.

Several models were advanced for designing intracardiac and transthoracic waveforms, comprising various electrode interface, generator and tissue resistances. We refrained from attempts at acquiring absolute values for the transmembrane voltage V_m induced by the applied voltage V_s waveform. Normalised relative changes of V_m time course were obtained by modeling the cell impedance Z_m with higher resistance and lower capacitance, resulting in a simplified equivalent circuit where the low total resistance between defibrillator electrodes is not shunted by Z_m.

We hypothesised that a simplified set of criteria for an efficient defibrillation pulse waveform would yield an approach for testing the efficiency of existing pulse waveforms. The feasibility of designing optimal waveforms by modeling was shown.

Introduction

The problem of designing efficient defibrillation impulse waveforms has been in the focus of many researchers since the first attempts at electrical therapy of fibrillation [16]. Various current waveforms were used: bursts of sinusoids, rectangular, triangular, exponential or truncated exponential pulses [1, 4, 15]. The early defibrillators using bursts of 50 or 60 Hz AC current were abandoned due to their heavy and bulky construction and to risk of dangerous post-shock arrhythmias. Defibrillators using direct capacitors discharge through the chest were much lighter and more compact, but the sharp leading edge was suspected to provoke myocardial damage [15]. Smoothing of the capacitor discharge waveform by inserting a series inductance was proposed by Edmark and Lown [16]. Thus the monophasic damped sinusoid waveform was developed. It was established as a standard in virtually all types of today's commercial defibrillators, with a maximum stored energy of 360 J [15]. However, Gurvich and Makarychev [7] used underdamped (biphasic) sinusoid requiring large inductance, but the maximum energy was only 190 J [1].

Recent investigations on implantable defibrillators yielded new cardiac electrophysiology data on thresholds and risk of refibrillation [2, 4, 15]. On the other hand, data of possible adverse effects with high energy transthoracic shocks were reported, leading to doubts whether the damped sinusoid was really an optimal defibrillation waveform [1, 3]. In order to cut the supposedly dangerous trailing edge of transmembrane voltage with damped sinusoid current, several types of truncated exponential impulses were suggested as more convenient [4], both for intracardiac and transthoracic application. For implantable devices these waveforms were preferred also for the possibility of building light and compact devices.

The newly obtained heart electrophysiology data resulted in attempts at modelling the cell transmembrane voltage waveform by electric circuits [2, 5, 10, 11, 13, 17]. Many researchers used the models to study biphasic pulse width, tilt and time constants of generator and cell [12, 13, 17, 19]; pulse amplitude [9, 18]; and phase separation [6]. The majority of these studies were directed specifically to intracardiac defibrillation. A tendency could be observed for testing separate waveform parameters, omitting to some extent their interdependence and interaction.

We attempted to model relative changes of the cardiac cell transmembrane voltage waveform during transthoracic defibrillation by taking into account just the generator internal resistance and the total resistance appearing between the electrodes. We modeled the cell time constant with a higher resistance and lower capacitance, rather than the opposite, thus avoiding the shunting effect of its impedance on the transthoracic resistance.

Method

Existing models take into consideration the generator parameters (V_c - initial voltage of the energy storage capacitor and Z_i - internal impedance), the electrode-tissue interface resistance R_e, one or several serial R_{st} and parallel R_{pt} resistive components of the thorax, heart tissue resistance R_h, and the cardiac cell impedance Z_m, represented as a first order R_mC_m network (Fig. 1).

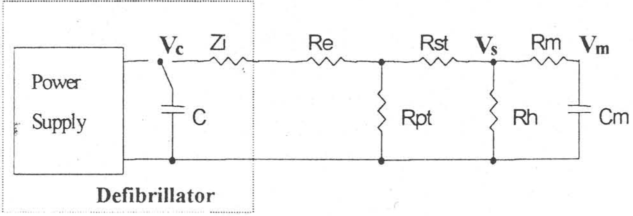

Fig. 1. Generalised model for defibrillation

We simplified the circuit in a way similar to that of Cleland [5], but selected to model the cell time constant with relatively high impedance Z_m (higher R_m and lower C_m) rather than the opposite. Thus we avoided eventual shunting effect of Z_m on the relatively low resistances of tissues. The latter were represented by an equivalent total thoracic resistance between the defibrillation electrodes R_{th} (Fig. 2):

$$R_{th} = R_e + \frac{R_{pt}(R_{st} + R_h)}{(R_{pt} + R_{st} + R_h)}$$

R_{th} is known to be in the approximate range of 30 to 100-120 Ω, with an accepted "standard" value of 50 Ω.

The cardiac cell time constant τ_m, measured during intracardiac studies, was reported by different authors to be in the range of about 2 to 5 ms [13, 17, 19]. The low value of R_{th} allows to set $R_m >> R_{th}$ ($R_m >> 100$ Ω) and $C_m = \tau_m / R_m$.

As the purpose of this study is to model different defibrillation pulse waveforms and compare their efficiency, we adopted normalised values instead of absolute values for V_m, and for the voltage applied to the cell V_s.

These considerations allowed us to use the simplified circuits of Fig. 2, where the V_s waveform can be easily obtained, as the generator parameters V_c and Z_i are known and the transthoracic resistance R_{th} is taken at its "standard" value R_{th}.=50 Ω.

Several types of most often used waveforms are considered: damped and underdamped sinusoids and some truncated exponential pulses.

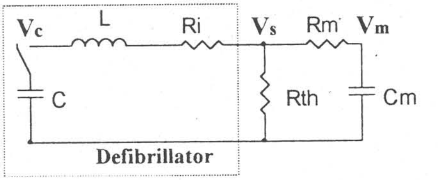

Fig. 2a. Model for damped sinusoid defibrillation

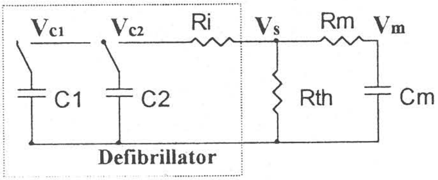

Fig. 2b. Model for biphasic pulse defibrillation

The transmembrane voltage V_m can be obtained from the applied voltage V_s using the following differential equation:

$$\frac{dV_m(t)}{dt} + \frac{V_m(t)}{\tau_m} = \frac{V_s(t)}{\tau_m}$$

Its general solution is: $V_m(t) = e^{-t/\tau_m}[\int V_s(t)\frac{e^{t/\tau_m}}{\tau_m}dt + c]$,

where c is an integration constant, depending on the initial conditions.

In modeling the "classic" damped sinusoid, (Fig. 2a) we selected the following parameters: C=30μF, L=20mH, R=(R_i+R_{th})=50 Ω. Then the voltage applied to the cell is:

$$V_s(t) = \frac{V_c R_{th}}{L(\alpha_1 - \alpha_2)}\left(e^{\alpha_1 t} - e^{\alpha_2 t}\right),$$

where $\alpha_{1,2} = -\delta \pm \sqrt{\delta^2 - \omega_0^2}$, $\delta > \omega_0$ $(R > 2\sqrt{L/C})$, $\delta = R/2L$, $\omega_0^2 = 1/LC$

An underdamped sinusoid, similar to the waveform of Gurvich and Makarychev [7], was modeled with parameters C=30 μF, L=200mH. Here the voltage applied to the cell is: $V_s(t) = \frac{V_c R_{th}}{L\omega'} e^{-\delta t} \sin \omega' t$,

where $\omega' = \sqrt{\omega_0^2 - \delta^2}$,$\omega_0 > \delta$ $(R < 2\sqrt{L/C})$, $\delta = R/2L$, $\omega_0^2 = 1/LC$

The next case relates to biphasic truncated exponential pulses (Fig. 2b). The capacitors are C1=C2=C=30 μF. Here

$$V_s(t) = V_c \frac{R_{th}}{R_{th} + R_t} e^{-t/\tau_s},$$

where $\tau_S = (R_t + R_{th})C$ - is the time constant of the defibrillator.

Results

The transmembrane voltage V_m obtained from a critically damped sinusoid is shown in Fig. 3. V_s and V_m are normalised and set to unity. These conditional units are used further in all cases under consideration, in order to facilitate comparisons. The cell time constant is assumed to be τ_m=4 ms.

The effect of an underdamped sinusoid is shown in Fig. 4, modeled by the circuit of Fig. 2a, where the following parameters were used: C=30 μF, L=200mH.

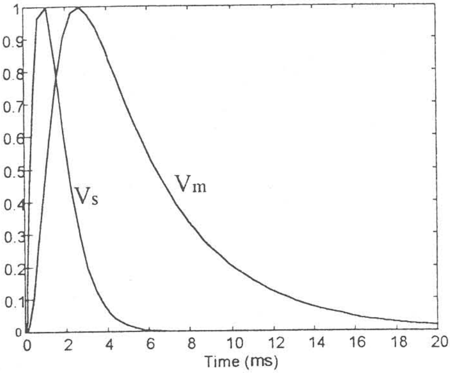

Fig. 3. Transmembrane voltage from a damped sinusoid pulse.

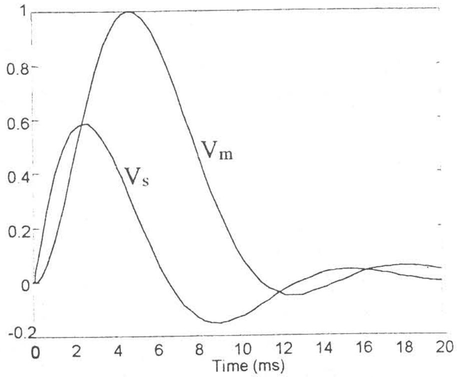

Fig. 4.A case with underdamped sinusoid

Modeling symmetrical biphasic pulses, according to the circuit of Fig. 2b, each of 4 ms duration with separation of 0.3 ms, resulted in overcompensated decay (Fig.5, V_{m3}).

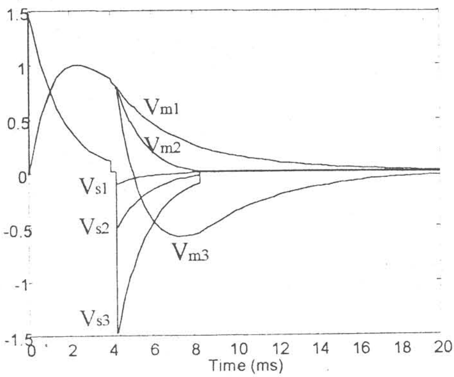

Fig. 5. The effect of different biphasic pulses

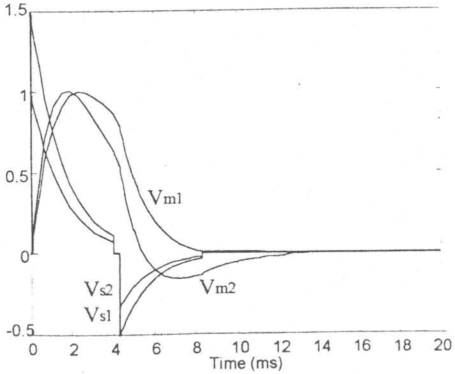

Fig.6. The Effect of cell time constant variation

Some implantable defibrillators use switching of one capacitor for generation of biphasic pulses, where phase 2 initial voltage (V_{s1}- Fig. 5) is equal to the end voltage of phase 1, in order to use at maximum the stored energy. In this case the transmembrane voltage V_{m1} has a long tail.

If a ratio of phase 1 and phase 2 voltages were adequately adjusted, an optimal of V_m time course could be reached. With a ratio of 1:3, the curve V_{m2} (Fig. 5) was obtained.

The effect of different cell time constants is shown in Fig. 6. Here the "optimal" V_{m1} corresponds to τ_m=4 ms and V_{m2} is with τ_m=2 ms. Both V_m were adjusted to the same threshold level, thus showing that higher voltage V_{s1} is needed for the larger time constant. On the other hand V_{m2} has an overcompensated decay.

Discussion

It was hypothesised that one possible cause of unsuccessful defibrillation, is that the transmembrane voltage V_m decays slowly after the depolarisation threshold was reached or exceeded. The trailing part of V_m may lead to repeated depolarisation of some cells whose refractory period had expired. Thus a new irregular depolarisation wave might arise, initiating in fact a refibrillation [8, 14].

Therefore, the purpose of defibrillation as formulated from Ideker et al [8] may be modified as follows: i) raising the transmembrane voltage V_m to its depolarisation threshold with a minimum of voltage or current, for a time interval compatible to the excitation process in the cell; ii) restoring the transmembrane voltage V_m to its initial level for a time interval compatible to the refractory period of the cell.

If the above modification was accepted, the results can be discussed as follows.

Using a damped sinusoid (Fig. 3), the second condition, namely an adequate decay of V_m, was not respected. V_m tends to return to its initial level as late as 12-14 ms after the peak.

The model with underdamped sinusoid (Fig. 4) showed an interesting result: V_m declined very well. This case could explain why the defibrillator of Gurvich and Makarychev [7] had a maximum energy of 190 J, while with the damped sinusoid 360 J were considered mandatory [1]. The question of how to obtain the waveforms V_s is not discussed here. It could only be mentioned that higher stored energy will be needed for the generation of the underdamped sinusoid.

The example with symmetrical biphasic pulses (Fig. 5, V_{s3}) showed that the high second phase amplitude could lead to hyperpolarisation and result in longer decline of the transmembrane voltage (V_{m3}), comparable to that with the damped sinusoid.

Reverse switching of a single capacitor for generating biphasic pulses leads to low second phase initial voltage V_{s1}, especially with lower load resistance (high tilt). The result is a long V_{m1} tail. It can be seen (Fig. 5) that this solution makes the second phase influence minimal. Selecting an adequate ratio of phase 1 and phase 2 voltages (Fig. 5, V_{s2}) it was possible to optimise the time course of V_{m2}.

Conclusion

Modelling the changes of the cardiac cell transmembrane voltage provoked by defibrillation pulses offers possibilities for detailed comparison of different waveforms. It allows for designing optimal pulse shapes, but expected variations in the cell time constants should be taken into consideration.

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