ENERGY MEASUREMENT OF BIPHASIC EXPONENTIAL HIGH FREQUENCY CHOPPED DEFIBRILLATION PULSES

Ivo Iliev#, Vessela Krasteva*

Technical University of Sofia
*Centre of Biomedical Engineering - Bulgarian Academy of Sciences
E-mail for correspondence: izi@alpha.vmei.acad.bg

Abstract

A previous study on optimization of defibrillation impulse waveforms led to the development and application of high frequency chopped (50% duty cycle) biphasic pulses. Thus the need arises for performance extension of the existing energy measurement defibrillation analyzers. A new measurement method is proposed, implemented by an additional analogue preprocessing module including a sample-hold circuit. Its purpose is to obtain an envelope signal that follows the input voltage during the pulse active phases and holds it during the pauses. For the working chopping frequency interval of 1-10 kHz, we selected a sample period of 50μs. The sample-hold circuit output signal is then digitized with a frequency of 10kHz. The measurement error of the method is independent of the input signal frequency and is virtually constant in the working frequency range, with a mean value of 1.6%. The accuracy obtained guarantees the patient safety and treatment efficiency in evaluating the delivered energy during defibrillation.

Introduction

Recent studies were directed to the development of more efficient defibrillation pulse waveforms, leading to reduction of the energy required to terminate fibrillation of the heart, compared to the conventional monophasic damped sine wave [2-6]. The objective is technological implementation of the new type biphasic capacitor discharge pulses in building light and compact public-access automatic external defibrillators (AEDs). The first commercially available low-energy biphasic waveform AED was introduced in 1998 [7].

Defibrillators are commonly calibrated either in joules of energy initially stored in the capacitor or in energy which would be delivered in a load of 50Ω. The patient safety and the defibrillation success depend mainly on the accurate delivery of the pre-set (by the operator or automatically) energy. The existing standards and the available energy analyzers used for testing of defibrillators [8, 9] are designed to measure the parameters of the standard monophasic damped sine wave pulses. They are inappropriate for measurement of biphasic and high frequency chopped waveforms.

The method and the analogue preprocessing module proposed here open the possibility for extending the range of tested defibrillators, irrespective of the type of waveform used.

Method

The new type of defibrillators [10, 11] use capacitor-discharge waveforms that are 50 % on-off "chopped" at a high frequency f_{HF}=(1 ÷10) kHz, into a series of short pulses, with an overall duration of several milliseconds.

The energy delivered to the patient is described by equation (1) and depends on the output circuit parameters:

- V_{c1}, V_{c2} - initial voltages of the energy storage capacitors C_1 and C_2
- R_{int} - defibrillator internal impedance
- R – the resistance between the defibrillator electrodes
- N_1, N_2 - number of HF pulses during the first and second phase respectively

$$E_{HF} = \sum_{i=1}^{2} \frac{1}{2} C_i V_{ci}^2 \left(1 - e^{-\frac{N_i}{f_{HF}(R_{int}+R)C_i}}\right) \tag{1}$$

All parameters of a defibrillator under test may not be available. Therefore, this equation could be practically unusable for calculation of the delivered energy.

The conventional method for energy measurement of sampled continuous pulses proposed in a previous work [1], calculates the sum of the momentaneous energies dissipated in the load R for the shock duration as follows:

$$E_{HF_1} = \frac{T_s}{R} \sum_{i=1}^{Ns} U_i^2 \tag{2}$$

Here T_s is the sampling period, N_s is the number of samples counted during the shock and U_i are the discrete values of the acquired pulse voltage, applied across the standard-value resistance $R=50\Omega$. The disadvantage in direct application of this method for HF chopped waveforms is the variation of the measurement error, which depends on the ratio of the chopping frequency f_{HF} and the analogue-to-digital conversion (ADC) rate – *Fig.3b-(1)*.

The proposed method requires to use a continuous signal that follows the input voltage during the active phase of the pulse and holds the signal level during the pause, thus obtaining the chopped pulse envelope. Then the envelope voltage is AD-converted with a sampling frequency F_s and the total energy delivered to R is calculated using the following equation:

$$E_{HF_2} = \frac{T_s}{2R} \sum_{i=1}^{Ns} U_i^2 \tag{3}$$

Here Ns *is* the number of acquired voltage samples and $Ts=1/F_s$ is the sampling period. The dependence of the method error on the chopper frequency is presented in *Fig3b-(2)*. It is constant and of negligible value in the entire 1 to 10 kHz range of HF working frequencies.

The measurement error was calculated according to the following equation:

$$Err = \frac{E_{HF} - E_{HF_{1,2}}}{E_{HF}} 100 \ [\%] \tag{4}$$

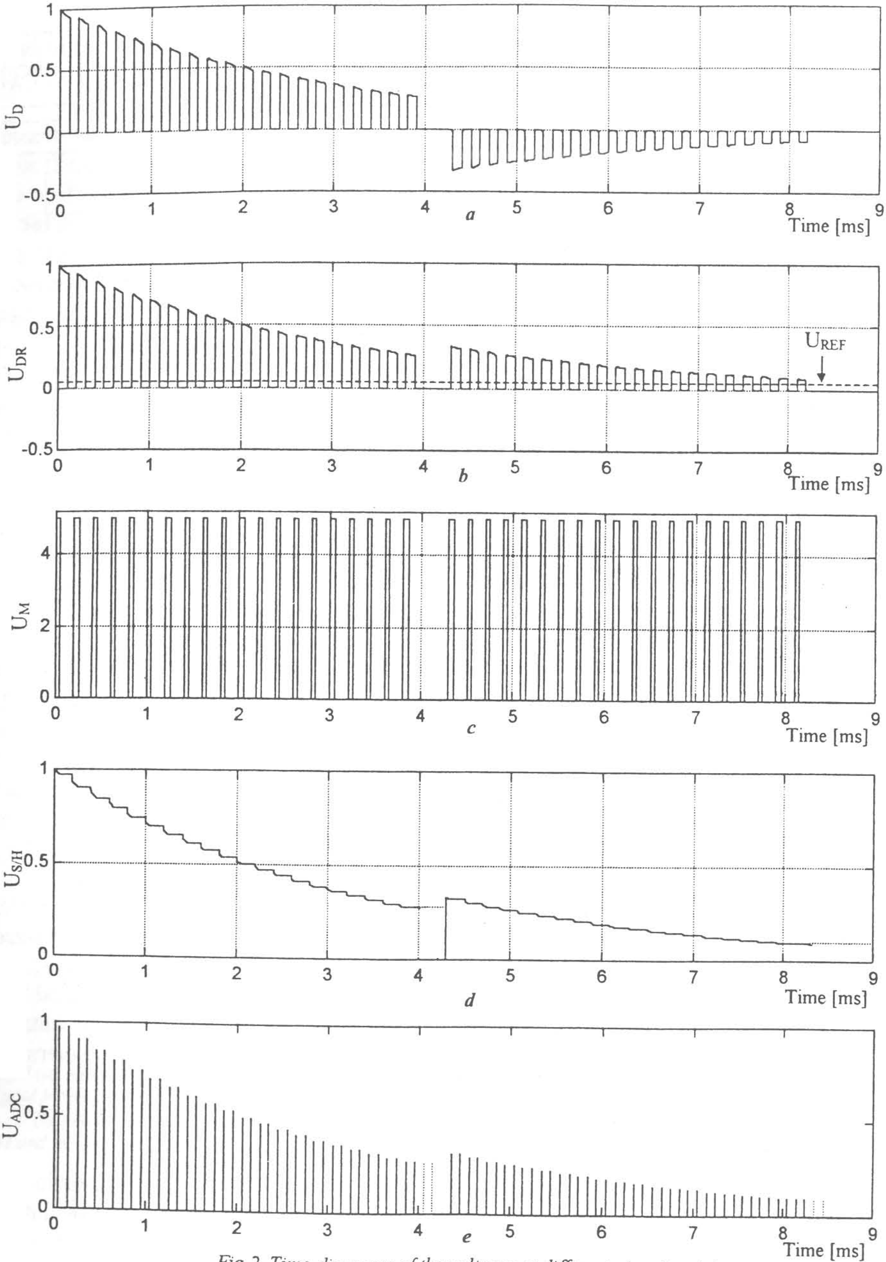

Fig. 2. Time-diagrams of the voltages at different circuit points

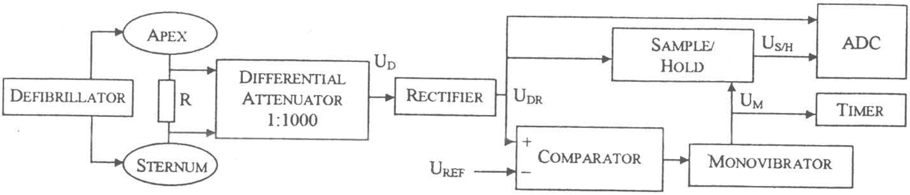

Fig.1 Block-circuit of the analogue preprocessing module

Hardware and Software solution

The detailed basic block-diagram of the defibrillator tester was described in a former work [1]. Here the additional analogue module is proposed (*Fig. 1*), which implements an analogue preprocessing of the biphasic series of high frequency voltage applied across the standard resistance R=50 Ω. The initial amplitude may be as high as 5 kV. The signal U_D after the Differential Attenuator 1:1000 passes through a full- wave rectifier, converting both phases in a positive voltage U_{DR}. The U_D and U_{DR} signals are shown in *Fig2 a* and *b*, respectively. In order to detect the front edge of each HF pulse, we used a comparator with open collector and reference voltage applied to its inverting input U_{REF}=0.06V. This value referred to the input of the circuit corresponds to 60V, which is well below the lowest possible level of a defibrillation voltage. The pulses obtained after the comparator are chopped with the input signal frequency and are normalized to a constant 5V amplitude. They are fed to a monovibrator, activated by each HF pulse front edge, which produces standardized fixed duration pulses U_M (Fig. 2c) for controlling the Sample-Hold (S/H) circuit. Thus the S/H output becomes a non-interrupted signal $U_{S/H}$ *(Fig. 2d)*. It follows the input voltage U_D during the active phase of the pulse for the time interval set by the monovibrator and holds the signal level during the pause. The S/H output signal is then digitized with a sampling rate of 10 kHz – U_{ADC} *(Fig. 2e)*, by the 8 bit ADC integrated in the used microcontroller 68HC11. The amplitude resolution referred to the input level is equal to 20V/bit. In order to detect properly the pause between both HF series and the end of the entire pulse, the initial edge of the monovibrator output signal U_M is also acquired, using the Timer system. A software method for input capture interrupt counts the frequency of the input HF signal. If the next initial edge of U_M do not come for a time interval equal to twice estimated HF period, the last two U_{ADC} samples are discarded because it is assumed that the voltage level of U_{DR} is zero (dotted samples in *Fig 2e*). Then the analog to digital conversion is prohibited until the next front edge of the U_M signal.

Results

The capacitor discharge chopped high frequency pulses with frequencies in the range 1-10kHz are generated by an in-house built experimental model defibrillator. The energy storage capacitors for both phases are $C_1=C_2=30\mu F$ and the internal resistance is $R_i=6\Omega$. The duration of the two HF series is 4ms, with a pause between them of 300µs. The reference energy E_{HF} delivered to the load R according equation (1), was calculated from the pre-set capacitor voltages for both phases U_1 and U_2.

We selected an adequate duration of the monovibrator output pulses U_M thus determining the optimum time for following the input signal during the active phase. Thus the error of the measured energy with respect to the calculated reference energy E_{HF} will be minimal (equation 4). The frequency of the input signal was selected to be 5kHz (the approximate mean value in the working range). The results are shown in *Fig. 3a*. It is evident that the minimal error is obtained when the sample time is between 50 and 60 µs. Further in our experiments we set a 50 µs duration of the monovibrator pulses.

The second aim was to determine the variation of the measurement error when the input frequency changes between 1 and 10 kHz. We studied the two methods described by equations (2) and (3). In the first case directly the rectified signal U_{DR} was examined and the corresponding error is shown in *Fig3b-(1)*. The measurement error of the signal after analogue preprocessing using equation (3) is presented in *Fig3b-(2)*.

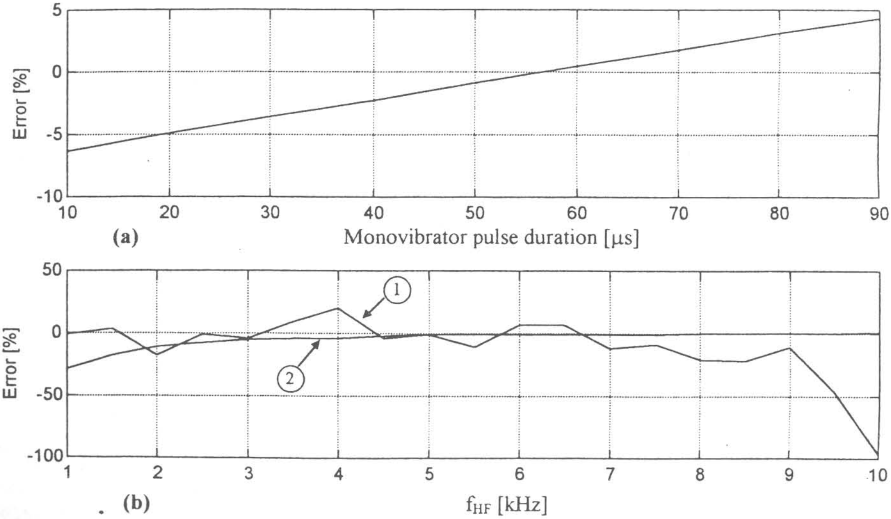

Fig.3 (a) –Relation between U_M pulse duration and the measurement error of the proposed method for input signal frequency 5kHz and sampling frequency Fs=10kHz.

(b) –Relation between the input signal frequency and the measurement error for the standard method (1) and the new proposed method (2) with sample period 50 µs and sampling frequency Fs=10kHz.

Discussion

The existing method for evaluating the energy of continuous defibrillation pulses is not applicable for the energy measurement of HF-chopped signals. This could be seen in *Fig.3b-(1)*, where the measurement error varies considerably with the HF pulses frequency f_{HF} and depends on the ratio f_{HF}/F_s.

The approach in evaluating the HF pulses energy using equation (3) is very simple and gives the opportunity to perform this operation by a single-chip microcontroller. Thus a portable defibrillator tester could be build or the performance of the existing defibrillator analyzers could be extended. An additional analogue preprocessing module must be implemented, in order to obtain the envelope of the 50% duty cycle HF-chopped biphasic defibrillator pulses.

The measurement error of the proposed method, with its mean value of 1.6%, is independent of the pulse chopping frequency in the range of 2 to10 kHz - *Fig3b-(2)*. This low error value satisfies the requirements for correct measurement of the energy delivered to the patient during defibrillation procedures. Thus the treatment efficiency and the patient safety should be guaranteed. This testing equipment is aimed to applications in hospitals, emergency teams, service and maintenance units.

Conclusion:

The proposed method for energy measurement of high frequency-chopped biphasic defibrillation pulses can be implemented by an analogue preprocessing module to be added to the input of a conventional defibrillator-analyzer. The measurement error of the method is negligibly low and constant in the chopping frequency range of 1 to 10 kHz, obtained by selection of an appropriate analogue-to-digital conversion frequency and duration of the sample period. The method is reliable and accurate, representing a simplified mean for energy testing of the newly developed and used defibrillation pulses. It can extend the performances of conventional defibrillator-analyzers.

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