

A SIMPLE BIOELECTRICAL SIGNAL SIMULATOR FOR MEASUREMENT DEVICE TESTING

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Abstract – A very simple PC based system for simulation of bioelectrical signals is presented. In addition to a regular PC with an audio card and the constructed software only few passive components are needed to accurately mimic the waveforms of the original recorded bioelectrical signals stored in the PC's hard drive. Alternatively, the simulated signals that are generated from the original recordings by Matlab can be played back with an mp3-player. Due to the nature of our method and the bandwidth limitations of PCs' audio card, the system is not suitable for simulation of high frequency signals but for bioelectrical signals like electrocardiogram (EKG) the method suits very well.

Keywords: measurement device testing, EKG simulator

1. INTRODUCTION

In the development of measurement systems for bioelectrical signals, especially if the system features an automatic signal processing, it would be useful to have a method for validating the whole system objectively. Commonly the operation of the different parts of the measurement system; measurement hardware, software, and the signal processing algorithms in the system are verified separately. However, being able at once to verify and validate the whole measurement chain would increase the confidence of the validation process. This requires that the example bioelectrical signals are fed to the system as they are originally recorded and the output of the entire system is observed for validity. One possibility of doing this would be to use a fairly expensive measurement card controlled from a computer. If the system is being developed, validated, manufactured, and tested in several places simultaneously, the cost of several test setups will increase quite high.

Our method of implementing the test setup for validation is extremely simple and inexpensive. The already recorded EKG or other signal is used to amplitude modulate (AM) a higher frequency sinusoidal carrier signal. This is done to shift the signal being simulated temporarily to higher frequency for enabling the use of computer's audio card or other audio device for signal playback. Before feeding the signal to the device that is being tested, it is demodulated with a simple envelope detector.

There exists a variety of devices made for testing the principle operation of EKG monitors. However, at least the

basic models of these devices do not verify the operation of the signal processing algorithms commonly integrated into the EKG devices. [1, 2]

Research has also been done with artificially generating EKG waveforms [3]. These systems are useful also for verifying the basic operation of signal processing algorithms alone i.e. how accurately the algorithms are recognizing the simulated arrhythmias etc. It is, however, difficult to artificially produce such signal artefacts and other unidealities that would efficiently demonstrate the real use case.

2. SYSTEM DESCRIPTION

In this section, the operation principle of our biosignal simulator is explained in detail. First the original signal to be simulated is retrieved from a database. Currently the data format of MIT-BIH database [4] and our own proprietary data format are supported. In reading the data from the MIT-BIH arrhythmia database's 212 format to Matlab we are using a script written by James Lamberg [5].

2.1. Scaling

After importing, the signal is scaled to the correct amplitude. In scaling two things need to be taken into account. First, for playing the signal with a soundcard, the whole signal has to be within the limits between -1 and 1 to avoid clipping. Second, the modulation depth or modulation index (the amplitude of the modulating signal, e.g. EKG, compared with the carrier amplitude) has to be quite small to reduce distortion. The distortion occurs partly because the forward voltage of the rectifying diode depends on the diode current and the current then depends on the voltage over the resistor of the envelope detector. Also the discharging speed of the envelope detector's capacitor depends on the voltage of the resistor, which is another source of distortion.

On the other hand, the modulation depth affects the amplitude resolution of the resulting signal. It also needs to be taken into account that the lowest amplitude of the modulated signal should not go below the threshold voltage of the rectifying diode. For these reasons we have selected to scale the 16-bit signal in our main application between values 0.5 and 1. This sets the dynamic range of the simulated signal to 14-bit when the whole -1 to 1 range is 16 bits.

After scaling, the signal is interpolated and modulated before it is ready to be played e.g. with a computer audio card.

2.2. Signal interpolation and modulation

The reason for signal interpolation is that the sinusoidal carrier signal has to have more data points than the original modulating EKG or other signal. The amount of data is therefore substantially increased, which is a concern when working with long recordings.

To decrease the distortion caused by the demodulation (the envelope detector), the carrier signal should have more than one period per one sample of modulating signal. Also every period of the carrier signal must consist of at least few data points. We have used settings in which the frequency of the carrier signal is five times the sample frequency of the original signal and each carrier period has five data samples. With a computer soundcard that is able to reproduce around 20 kHz frequencies at maximum, the highest sample frequency for the signal that is simulated can then be 4 kHz. However, if the signal to be simulated in this case consists of high frequency components, the properties of the envelope detector need to be considered carefully.

In amplitude modulation we use a double sideband transmitted carrier (DSB-TC) method. The reason for choosing this method is that it is very simple to demodulate because the phase of the carrier signal does not change and therefore no synchronous detector is needed [6]. Figure 1 shows an image of the modulated signal.

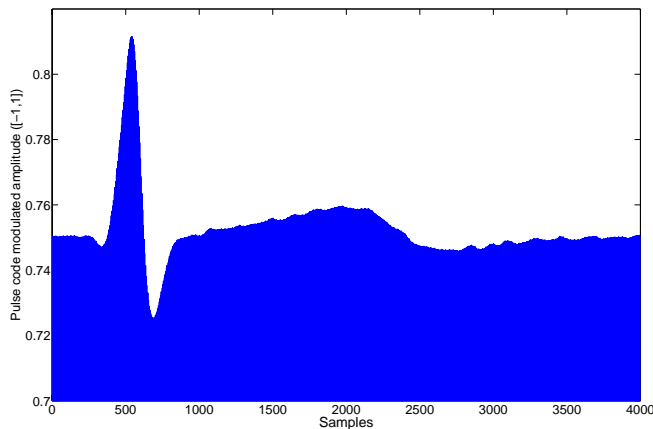


Figure 1. The carrier signal amplitude modulated by EKG signal.

2.3. Signal playback

The modulated signal can either be played with Matlab or saved as .wav file and played later with e.g. Winamp®. To reduce the size of the file it is also possible to encode it to .mp3 or .wma format and play with a portable mp3-player. Besides an affordable price and small size, the benefits of using mp3-player as the playback device also include the fact that this way an electrically floating signal source is easily achieved. When using a battery operated portable device for playback it should still be noted that these devices are normally not able to produce as high output amplitudes as computers or audio amplifiers. This

may cause restrictions to the component values of the envelope detector and amplitude that can be simulated.

One benefit of our method also is that there are two separate audio channels in normal stereo recordings, which allow two channel EKG recordings being included into one file.

2.4. The envelope detector

Two basic possibilities for easily demodulating the (DSB-TC) AM-modulated output signal are a simple passive envelope detector made with a diode, capacitor, and resistor and a precision rectifier circuit, which besides the aforementioned components, includes an operational amplifier.

When using the diode detector, it should be considered that the modulation depth and the minimum amplitude of the modulated signal are limited due to nonzero forward voltage and the current-voltage dependence of the diode. The precision rectifier on the other hand does not have these limitations but it requires more components and a power source. The passive version of the envelope detector gave adequate results in our tests and therefore we ended up using it.

After demodulating with the envelope detector, the signal is scaled down with a resistor division. Even though the modulation depth has to be small, the signal amplitude still needs to be higher than the original one in order to keep the diode in its operational region. Also the source impedance of the simulator can be set close to the original source (human) with the resistor division.

The components of the RC-filter for the envelope detector need to be selected so that as little distortion as possible is generated by the filter. The determinants are the highest frequency of the modulating signal and the frequency of the carrier signal. On one hand, the descending slew rate of the filter should not be lower than the highest derivative of the simulated signal. On the other hand the slew rate should be as small as possible to minimize the saw wave type of distortion between the peaks of the carrier signal. [6]

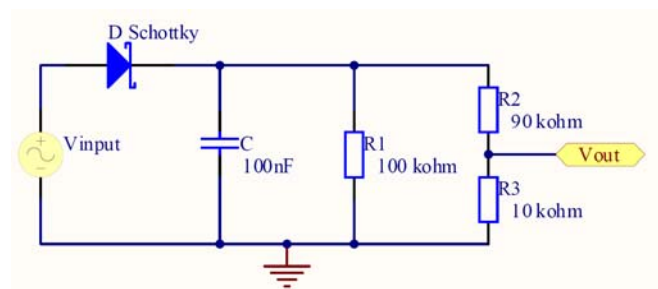


Figure 2. Schematic diagram of the envelope detector for one channel.

2.5. Calibration of the signal amplitude

An important aspect in simulation of the bioelectrical signals with our method is the setting of the correct output amplitude. The reason is that computers or mp3-players lack an absolute adjustment for the output amplitude because in

normal use the users adjust the level based on the audible feedback from the speakers. For this reason the sound level settings need to be calibrated and set for every computer or other playback device separately and every time when the sound volume settings might have been adjusted between using the simulator.

In calibration, we are using 10 Hz sinusoidal signal with known amplitude. The modulated signal is played with the computer or an mp3-player and the output amplitude of the envelope detector is measured with an accurate multimeter.

Then the sound level is adjusted from the computer until the output amplitude reaches the desired value. The amplitude can also be observed from the screen of the device that is used to measure the simulator's signal if such display exists and if it has a visible scale.

As explained in Section 2.1, the whole signal is scaled between -1 and 1 when creating wav-files. Because the simulated signal is used as a modulator in AM output and the DSB-TC modulation method is used, the simulated signal has to be between 0 and 1. When selecting the actual lower limit for the simulated signal, the Schottky diode's forward voltage has to be considered.

In our main application, EKG signal simulation, we have selected to bind ± 10 mV input signal to 0.5–1 span. When the amplitude of the unattenuated output signal of the envelope detector is set correct, the DC-level of the unattenuated signal is $30 \text{ mV} \cdot 10 = 300 \text{ mV}$. The measurement device that is being used with the simulator therefore has to have at least 30 mV dc-offset voltage tolerance. The RMS voltage measured over the resistor R1 with the multimeter should now be $10 \text{ mV} / \sqrt{2} \cdot 10 \approx 70.7 \text{ mV}$ when the output amplitude is set correct.

The peak-to-peak amplitude of QRS complex is normally between 1 and 5 mV depending mainly on the electrode locations. Therefore some measurement devices may have less than ± 10 mV input range. In these cases calibration signal with smaller amplitude can be used.

A drawback of this adjustment method is that due to the nature of the human's aural perception (changes of less than ± 3 dB in the sound intensity are basically not noticeable by the ear), the adjustment of the soundcards' or mp3-players' output amplitudes are very coarse and a combinations of different sound volume controllers (master volume, preamplifier, equalizer, etc.) has to be used in order to achieve a correct setting. There is also a change that setting the amplitude exactly enough with these controllers is still impossible.

To improve solving this problem the resistor R2 could have few alternative values set by a switch. Using a potentiometer to set the R2-R3 voltage division would enable an exact tuning of the amplitude but this way the relation between the measured voltage and the actual output would be lost.

3. RESULTS AND DISCUSSION

3.1 Analysis of the signal distortion

Two main mechanisms exist, which cause distortion to the simulated signal. One is the diode forward voltage's dependence on the diode current. In general the forward

voltage of a diode increases roughly 20 mV for each doubling of the diode current according to equation 1 [7].

$$\Delta V = V_T \log_e \frac{I_2}{I_1}, \quad (1)$$

where V_T is the thermal voltage of the diode and is usually approximately 25.85 V at room temperature.

Defining the diode current at certain moment of time and its effect on the envelope detector's output is not straight forward. For simplicity, we assume that the effect is directly proportional to the capacitor's discharge current and hence its voltage. In our measurement setup, the capacitor's voltage can vary between 200 mV and 400 mV. The amplitude distortion caused by this is 17.9 mV, and after scaling with voltage division 1.79 mV.

Another main mechanism is the discharging velocity of the capacitor, which depends on its capacitance, the voltage over it, and the parallel connection of the resistors:

$R1 \parallel (R2 + R3)$ in Figure 2. The discharging causes a saw wave type of distortion to the demodulated signal. However, if the sampling of the demodulated signal is synchronous with the carrier signal ($f_c = n \cdot f_s$), the effect of the saw wave itself is eliminated. What then causes the distortion is that the discharging velocity of the capacitor is higher when its voltage is higher. Equation 2 shows the change of the capacitor voltage as a function of time after the capacitor has been recharged by the input signal. In the case of synchronous sampling, where t is constant, the amplitude of distortion that is caused by the discharging is directly proportional on the initial voltage V_0 . The difference between two ΔV_C calculated at the highest and lowest phase of the modulating signal is the amplitude distortion caused by the discharging.

$$\Delta V_C(t) = V_0 \left(1 - e^{-\frac{t}{RC}} \right). \quad (2)$$

The effect of discharging in case of synchronous sampling also depends on the constant time t that is randomly set at the beginning of a measurement. If t is in the middle of two periods of the carrier ($t = 0.4$ ms in our application), the scaled amplitude distortion is 1.54 mV. In the worst case, when the sample is taken right before the new recharging of the capacitor ($t = 0.8$ ms), the scaled distortion is 2.96 mV.

Figure 3 shows a simulation result from the operation of the envelope detector. The larger sinusoidal wave is the AM modulated input signal and the saw tooth wave is the unattenuated output. The distortion caused by the variation of the diode voltage is seen as the non-constant cap between the input signal and saw wave spikes (i.e. diode voltage). The distortion caused by the discharging of the capacitor is seen as the variation of the amplitude of the saw wave.

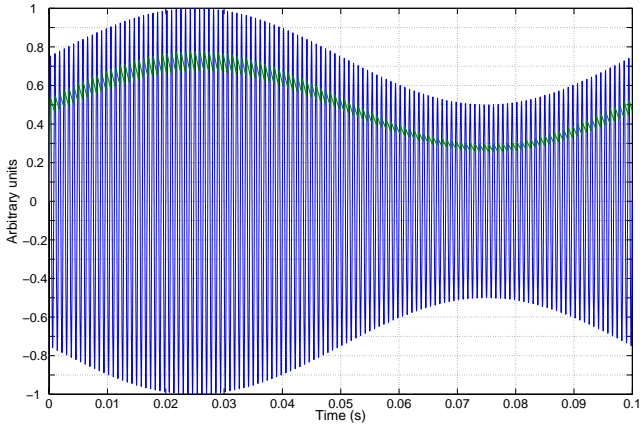


Figure 3. Simulated input (blue) and output (green) of the envelope detector.

In total, the two distortion effects may produce as much as 4.75 mV of the attenuation which is 23.8 % of the 20 mV amplitude span. However, because we set the input amplitude by measuring the output of the demodulator, the amplitude distortion does not affect that much the amplitude of the simulation result but it affects slightly its waveform. Figure 4 shows the original Matlab generated sinusoidal calibration signal compared with the same signal recorded from the simulator. As seen from the figure, the upper part of the simulation result is wider than in the original sinusoidal signal, which is a result of attenuation of higher amplitudes. The distortion of the waveform also affects a little the amplitude when calibrating using a multimeter because the relation of the amplitude and the RMS value of the distorted waveform is not exactly the same as with sinusoidal signal. $A \neq \sqrt{2} \cdot V_{RMS}$

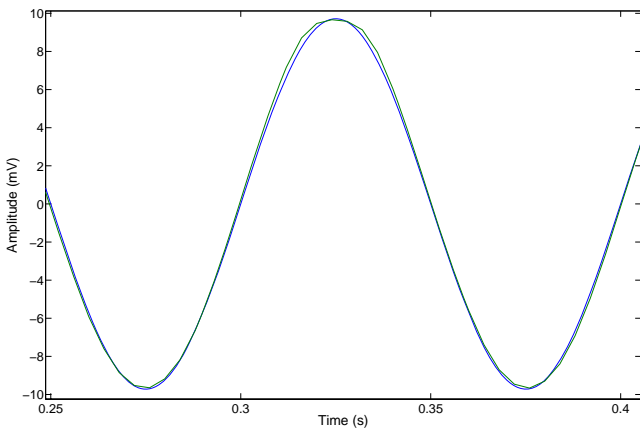


Figure 4. The simulation result (green) compared with the original sinusoidal signal (blue).

3.2 Results of biosignal simulation

As stated, the simulation of EKG signal has been our primary target when defining modulation and scaling parameters and the component values for our envelope detector. Figure 5 shows the output signal generated by the simulator (green) in comparison with the original EKG signal (blue). The sample is a part of the recording number

100 of MIT-BIH arrhythmia database. As seen from the figure, the simulated signal corresponds well with the original one. Two main differences between the original and simulated one are the increased noise and slightly lower amplitude in the simulated signal. Both of the effects can be decreased by better shielding and more accurate tuning of the simulator as seen from the coming figures.

The simulation result has been recorded using 250 Hz sample frequency and amplifier that has high-pass filter at 0.016 Hz and a low-pass filter at 125 Hz. The original MIT-BIH data has a pass-band from 0.1 Hz to 100 Hz. Strangely, even it is claimed that the MIT-BIH-data has a high-pass filter at 0.1 Hz it is seen in the image that the data has a dc-offset of around -0.3 mV. Because our measurement device used to record the simulation result actually has a high-pass filter, this offset is removed. Accurate simulation of the signal with a dc-component is not possible directly with our simulator because it adds its own dc-component (30 mV with our current settings) to the output. This dc-component however is constant so it can be removed afterwards if the measurement device is capable of recording it flawlessly, i.e. is dc-enabled.

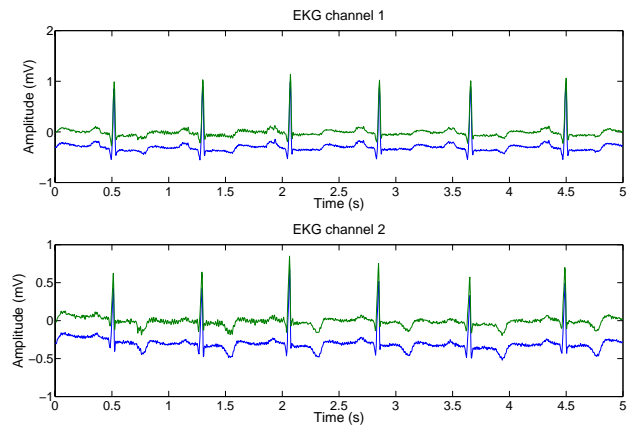


Figure 5. The output signal of the simulator (green) in comparison with the original signal (blue).

Figure 6 shows a result of an EOG simulation. EOG signal includes very low frequency components that result from a static direction of gaze and high frequency components called saccades which are a result of changing the direction of the gaze.

The horizontal EOG is measured between two electrodes that are placed on the temples and the vertical EOG is measured from the electrodes placed above and below the eye. Eye blinks are another source of high frequencies in the signal and are seen as spikes in the vertical EOG.

The recording of the original signal was made with the same device as the recording of the simulated EKG but for the recording of the simulated signal we decreased the high-pass filter's cut-off frequency to 0.0080 Hz. It should be noted that if the simulated signal is measured with the same device that is originally used to collect the signal, the signal will be distorted in the measurement because it is then re-filtered from the same frequencies.

The dc-levels of the signals in Figure 6 are altered to draw the signals apart and to get them more comparable.

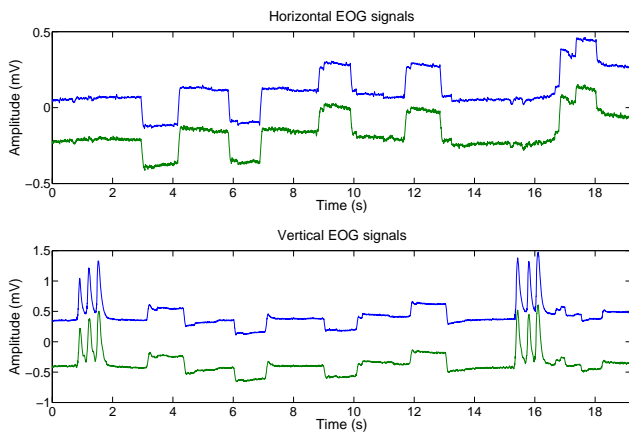


Figure 6. The output signals of the simulator (green) in comparison with the original EOG signals (blue).

Figure 7 shows a simulated EMG signal compared with the original one. The EMG is recorded from the same electrodes as horizontal EOG in the previous figure. The sources of the EMG are muscles responsible for occlusion.

As seen, also the EMG signal that consists of high frequencies can be adequately simulated with our method.

Also here the dc-levels of the signals are artificially altered for clarity.

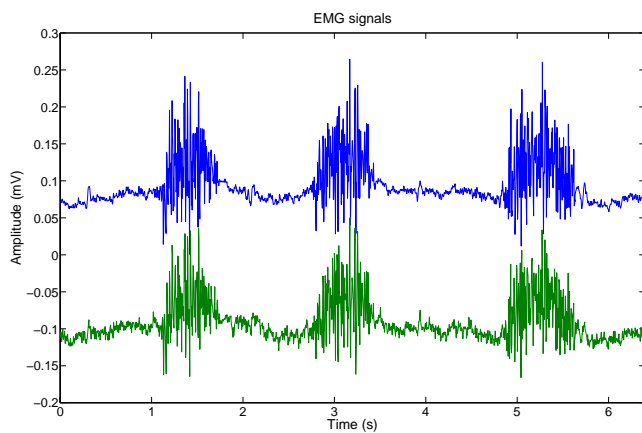


Figure 7. The output signal of the simulator (green) in comparison with the original signal (blue).

It should also be noted that portable mp3-players are usually not able to produce as high output voltages as for example computers' audio cards. Therefore it may be

necessary to change the scaling of the output signal (voltage division) smaller. If only computers or devices with separate power amplifier enabling higher output voltages are used for playback, it may be beneficial to set the scaling larger (e.g. 1/100 instead of current 1/10). This would decrease the distortion caused by the diodes forward voltage variation.

4. CONCLUSION

We have developed an extremely cost efficient way of simulating bioelectrical signals in device testing purposes. The method also suits for simulating other signals up to around 1 kHz. The maximum sample frequency of the signal that is being simulated is 4 kHz. The method of the simulation set limits to the amplitude of the original signal.

As seen from the figures 5, 6, and 7, the simulated EKG, EOG, and EMG signals are reproduced fairly accurately by our simulator.

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