A NOVEL ULTRASOUND BASED AUTOMATED PULSATILE FLOW DETECTION METHOD FOR RESUSCITATION

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ABSTRACT
Early defibrillation is critical for successful resuscitation of a sudden cardiac arrest patient. Defibrillators currently do not assess the patient’s heartbeat or blood circulation and the responder has to manually check for the pulse, a procedure known to be very inaccurate. In this work an ultrasound Doppler based approach to determine the presence of blood flow in the context of resuscitation is proposed. A new pulsation index based on a spectral analysis of the Doppler signal is proposed for automated flow assessment. The method was demonstrated on data from pigs undergoing cardiac arrest and resuscitation.

KEY WORDS
Ultrasound, Doppler, cardiac arrest, resuscitation, pulse detection.

1. Introduction
Sudden cardiac arrest (SCA) is the leading cause of death in the US and countries with equivalent lifestyles. In the majority of cases, the initiating event of a sudden cardiac arrest is either a pulseless ventricular tachycardia (VT) that degenerates rapidly to ventricular fibrillation (VF) or primary VF [1]. For such patients, early defibrillation applied within a few minutes after cardiac arrest is the only method of successful resuscitation. Given that a majority of sudden cardiac arrests occur out of the hospital environment, the use of automated external defibrillators (AEDs) in public places by lay responders assumes great importance [2].

Defibrillators analyze the ECG of the SCA patient and recommend a therapeutic shock if VT or VF is present. However in as many as 30 to 60% of cases, the patient is in a state of pulseless electrical activity (PEA) or asystole after the defibrillation shock is applied [3, 4]. PEA is the condition where the patient’s ECG has a normal rhythm but the heartbeat, blood circulation and pulse are absent. Because of the possibility of PEA or asystole, the protocol requires that the rescuer manually check the pulse of the patient after defibrillation by placing his/her fingers on the neck at the location of the carotid artery.

Recently many studies have questioned the adequacy of responders to manually assess the pulse [5, 6]. The Eberle study [6] specifically showed that the accuracy of the manual pulse check procedure of lay responders (first responders and emergency medical technician students who have undergone a 4 hour basic life support course) was about 53% even when they took more than the recommended time of 10 seconds to make their decision. Moreover, only about 6% were able to make a decision within 10 seconds. For trained personnel, (paramedics) the corresponding numbers were about 82% and 33% respectively. Because of these limitations, the recent guidelines recommended the elimination of pulse check by lay responders [7].

The consequence of incorrectly concluding that the pulse is present when it is actually absent (probably due to feeling one’s own pulse) is that the responder fails to perform CPR or other life saving measures, which leads to lack of perfusion to the body’s vital organs, especially the brain, and eventually death. Conversely, the consequence of incorrectly reporting that the pulse is absent when it is actually present is that the rescuer initiates CPR, which may impede recovery of the heart [6]. Therefore, a reliable, noninvasive, automated method to ascertain the presence of blood flow and therefore pulse would be of great benefit in providing care to a sudden cardiac arrest patient. Pulse check is also required during the initial assessment before defibrillation to ascertain whether the patient has suffered sudden cardiac arrest. Additionally, such a method would be useful in other related situations such as non-invasive monitoring of a patient’s vital signs in the intensive care unit, emergency room or operating room without the need for invasive arterial lines.

One possible approach for pulse detection is measurement of thoracic electrical impedance, which is linked to changes in thoracic volume [8]. However this approach has not been studied in the context of post-defibrillation to evaluate the sensitivity for cases in which victims are recovering from initially low but gradually increasing arterial pressure. Another approach is pulse oximetry, which is a non-invasive and optical method for measurement of arterial oxygen saturation and pulse rate. However, in the case of cardiac failure, a pulse oximeter
may not be accurate since peripheral pulsations are minimal or absent due to peripheral vasoconstriction. Other techniques for pulse detection include passive methods such as measurement of tissue displacement from arterial wall pulsatile movement, or pressure propagating through the tissues to the skin surface and having their source from inside the body. Such techniques are limited to vibrations that propagate to the skin surface with sufficient intensity and are affected by breathing or motion artifacts.

In this work, we propose a continuous wave (CW) ultrasound Doppler based approach for assessing the pulse of a patient by determining the presence or absence of blood flow in the carotid artery, since the pulse is a direct consequence of the pulsatile blood flow in the artery. The CW Doppler approach was chosen because the carotid artery depth in a patient is not precisely known. The pulsed wave (PW) approach would have the added complexity of searching for the artery at several depth gates. The technical challenge is that, unlike in a typical ultrasound examination in cardiology or general imaging, the decision making process in the present situation needs to be fully automated. This is especially important for lay responders using AEDs in public places under time-critical conditions. In this work, we propose a new quantitative index that can be used for automated decision-making and demonstrate initial results based on animal trials.

2. Limitations of previous quantitative Doppler measures

Before we describe the new method for automated pulsatile flow evaluation, it is instructive to briefly review other existing methods that are presently used in commercial ultrasound systems.

One possible measure is the total Doppler power integrated over the useful frequency range. If this quantity exceeds a pre-determined threshold, one could conclude blood flow. However this quantity is strongly influenced by the placement of the transducer on the patient’s skin surface. In the present application, the lay user can only be expected to place the transducer approximately near the carotid artery, and therefore the computed Doppler power would vary depending on the position of the artery in the transducer’s beam pattern. Also, because of wide inter-patient variability it is not feasible to determine a universal threshold that can adequately cover the variety of human physiologies. Furthermore, the Doppler power does not determine the presence of an organized pulsatile flow, but only the presence of any flow be it pulsatile or non-pulsatile. For that reason, its value could be strongly affected by motion artifacts, e.g. while the patient is in an ambulance.

The intensity weighted mean Doppler frequency or velocity is another parameter that is commonly used in conventional Doppler displays. A possible method would be to conclude that blood flow is present if the estimated mean Doppler frequency is above a pre-determined threshold. However when a CW Doppler system is used, the presence of flow in the nearby veins in the opposite direction to that of the arteries would reduce the mean Doppler velocity. This is because the CW system does not have the range discrimination to separate signals from veins and arteries. Also, in the case of turbulent flow, e.g. due to stenosis, both positive and negative Doppler shifts are possible even for flow in a single artery. This could lead to mean velocities that are close to zero, similar to the one for the no flow case. Moreover if a simple demodulation (without quadrature demodulation) were used, then the mean Doppler frequency when there is no flow would be quite high, possibly at the center of the useful BW, and comparable to that of the case when there is blood flow. Hence this quantity does not have a good capability to separate the flow and no flow situations.

Pulsatility and resistive indices are also commonly used in quantitative Doppler analysis [9]. These quantities require the determination of the peak and minimum velocities in the Doppler spectrum. The peak velocity is typically estimated using a threshold procedure [10]. The threshold is based on an estimate of the noise floor plus an additional level, e.g., 6 dB. The method marches from the Nyquist frequency towards zero and then chooses the first frequency where the Doppler power exceeds the threshold. The additional 6 dB level is needed to ensure that the algorithm reliably picks a frequency where the signal is well above the noise. However, when there is no flow, the method might fail to yield a peak frequency measure because it is possible that there is no frequency for which the power exceeds the noise plus 6 dB threshold. In other words, the peak velocity estimation is meaningful only when there is a priori knowledge that there is flow. Furthermore, the wall filter setting can significantly affect the pulsatility index and resistive index values making them inappropriate for the present application where the user cannot be expected to optimize the wall filter settings.
Therefore we developed a new method for pulsatile flow assessment explicitly addressing the possibility that the heartbeat and blood flow might be absent. This method is described in Section 4 using a typical data set collected from a pig where data acquired before and after defibrillation were used to test the method.

Figure 2: Ultrasound Doppler, ECG and ABP signals recorded during a typical experiment. VF was induced at about 18 seconds and the animal was subsequently resuscitated using the defibrillator at about 32 seconds.

3. Experimental Methods

Figure 1 shows a schematic of the CW experimental setup that was designed and built for the experiments. A 5-MHz single element transducer (Panametrics, Waltham, MA; Model A309S) is excited by an arbitrary waveform generator (Wavetek/Fluke, Everett, WA; Model 295) in a CW mode. Another transducer identical to this transmit transducer collects the Doppler shifted backscattered echoes. The received signal is amplified using two low noise pre-amplifiers (Minicircuits, Brooklyn, NY; Model ZFL-500LN) each having 28 dB of gain. The signal after pre-amplification is sent to a mixer (Minicircuits; Model ZP-3MH), which also receives part of the excitation signal from the Wavetek generator. The output of the mixer contains a signal that has the sum and difference frequencies of the excitation signal and the received signal. A low pass filter (Minicircuits; Model BLP-1.9) removes the signal at the sum frequency leaving the Doppler signal at the difference frequency to pass through.

Figure 3: Auto correlation of the Doppler power in four frequency bands and their Fourier Transforms (power spectra). The auto correlations were computed using a 5 second window whose end point is shown on the x-axis. The y-axis is the lag time at which the correlations were computed. The color/gray level of the image represents the actual correlation value. For the power spectra, the x-axis is the same as that for the auto-correlations whereas the y-axis is the frequency, and the color/gray level indicates the strength of the spectral component.

The Doppler signal is further amplified and filtered by a two-channel tunable filter and amplifier (Krohn-Hite Corporation, Brockton, MA; Model 3382). A PCMCIA card DAQ card (National Instruments; NI 6062E) was used for the laptop based data acquisition. A Labview based graphical user interface was used to extract data from the DAQ card and display on the screen in real time. During the experiments, the ECG and arterial blood pressure (ABP) signals were recorded together with the ultrasound Doppler data at a sampling rate of 20 kS/s.
Figure 4: Fourier Transform of auto correlation, i.e. power spectrum, of the Doppler power in the 1150 to 1350 Hz band for the (a) initial state during normal cardiac activity and (b) VF state. For normal pulsatile flow, the Doppler power is concentrated around a specific frequency (and its harmonics), whereas for the VF state, such a behavior is absent.

Animals were used in this study since tissue-mimicking flow phantoms would not capture the complex behaviour of the heart after resuscitation. Domestic swines, weighing about 30 kg were anesthetized and intubated. A sternotomy was performed to expose the heart, and VF was induced electrically using a low voltage electrical shock at the surface of the heart. After about 15 seconds of VF, a defibrillating shock from a defibrillator was applied using hand-held surgical electrode paddles. The ultrasound transducers were held on top of the carotid artery location using mechanical clamps and coupled to the skin using conventional ultrasound gel. The ECG and ABP signals were also continuously monitored to serve as ground truths for the electrical and mechanical activities of the heart. The open heart was also visually monitored for its mechanical beating activity. A minimum of three minutes was allowed to elapse for stabilization of the animal’s heart before another experiment was done. About 28 experiments were done for each animal.

4. New approach: Pulsation index

In this Section we use a typical data-set to illustrate the development of a new metric, the pulsation index. Figure 2 shows the data recorded with our system during a typical experiment. The central idea is to pick a specific Doppler frequency band and analyze the Doppler power within that band as a function of time. This would imply taking a horizontal band across the spectrogram image in Figure 2, and analyzing its temporal variations. When a pulsatile flow is present, this quantity would show a periodic behavior that represents the changes from systole to diastole within a cardiac cycle. Depending on the flow condition, certain frequency bands would emphasize this pulsatile behavior better than others. For instance, for normal flow conditions, the low velocity portion of the flow would be more or less constant and would not change from systole to diastole. Hence the low Doppler frequency bands would show more or less a constant (not an obvious pulsatile) behavior over time. On the other hand, at higher frequency bands, only the systolic phase would show sufficient Doppler power and the diastolic phase would not. Hence, the power within these frequency bands would be pulsatile.

Figure 5: Proposed pulsation index for the data shown in Figure 2. For color copy: The color of the graph indicates the frequency band for which the index was the maximum - Black: 225-425 Hz, Red: 650-850 Hz, Green: 1150-1350 Hz, Blue: 1650-1850 Hz.

To illustrate the idea, the Doppler power in four frequency bands were selected for analysis: 225 to 425 Hz, 650 to 850 Hz, 1150 to 1350 Hz, and 1650 to 1850 Hz. The unbiased auto-correlation of these Doppler powers within sliding 5-second windows and their Fourier Transforms (power spectra) are shown in Figure 3. The pulsatile nature of the Doppler power during the initial state (up to about 18 sec) and recovery state (after about 32 seconds) is evident from the periodic nature of the autocorrelations and the peak in the power spectra. It can also be seen that some of the frequency bands (e.g., 1150 to 1350 Hz) expose the periodic nature better than the others. However, during the VF state between 18 and 32 seconds, such a periodic behavior is absent.

Figure 4 shows the power spectra in the 1150 to 1350 Hz band obtained from Figure 3 at two specific time instants. The two time instants correspond to the cases when the 5 sec windows used in the auto correlation ended at 10 and 30 seconds respectively. The former corresponded to the
initial state of the heart before fibrillation and the latter to the VF state. It can be seen that during the initial state, the Fourier Transform (FT) showed a peak at a frequency of about 2.58 Hz, which corresponded to a heart rate of 155 beats per minute, the same rate measured by the defibrillator monitoring the ECG signal. In this particular case, a significant second harmonic is also seen at twice the fundamental frequency. However during the VF state, the FTs do not show the presence of a strong peak.

Based on the results shown in Figure 3 and Figure 4, it is possible to compute a pulsation index measure that represents a pulsatile behavior of the flow. As described above, the Doppler power in several frequency bands is computed as a function of time, followed by the computation of the auto-correlations and power spectra. A peak-searching algorithm then determines the frequency at which the power spectrum is a maximum. The fraction of the total power contained within a narrow band around this frequency is determined. For the case of normal pulsatile flow one would expect that a significant portion of the total power is present in this narrow band whereas it would not be the case when pulsatile flow is absent. Hence a pulsation index based on this fraction could be used to differentiate the presence or absence of a pulse. We propose the ratio of the power in a narrow band around the peak frequency to that of the total power excluding the second harmonic to be the pulsation index. This quantity is close to zero for no flow and close to unity when there is a periodic, pulsatile flow.

A priori assumptions based on physiology could be used to restrict the search space for the location of the peak in the power spectrum. For instance, for the data recorded from pigs, it could be assumed that during normal flow in the carotid, the heart rate would be between 40 and 240 beats per minute. Thus the algorithm would search for the maximum of all the local maxima between 0.67 and 4 Hz.

We now address the issue of which Doppler frequency band to use for the computation of the pulsation index. The frequency band that captures the pulsatility information best depends on several factors such as the transmit frequency, angle, and the blood flow conditions. Since the flow conditions are not known a priori, it is not possible to set one optimal frequency band. Instead we propose that the pulsation index be computed for several bands, and the maximum among the values in all frequency bands be determined. Such a quantity would automatically identify the most optimal band. Henceforth, we will refer to the maximum pulsation index from all the frequency bands simply as the pulsation index.

5. Results

Figure 5 shows the pulsation index for a period of 60 seconds computed for the same experiment shown in Figure 2. For comparison purposes, the ECG and ABP are reproduced. It can be seen that the pulsation index is high for the case of normal and recovery states, and is low for the VF case. The pulsation index closely follows the trend in the ABP signal.

Figure 6 shows the pulsation index computed for another experiment in which the animal was initially put in VF for about 5 minutes, and defibrillated. The ECG returned to normal state, but the ABP did not. For color copy: The color of the graph indicates the frequency band for which the index was the highest - Black: 225-425 Hz, Red: 650-850 Hz, Green: 1150-1350 Hz, Blue: 1650-1850 Hz. The pulsation index is essentially the same during the VF and PEA periods.

![Figure 6: Pulsation index for the case of VF and transition to PEA. At about 308 seconds, a defibrillating shock was applied and the ECG returned to normal state, but the ABP did not. For color copy: The color of the graph indicates the frequency band for which the index was the highest - Black: 225-425 Hz, Red: 650-850 Hz, Green: 1150-1350 Hz, Blue: 1650-1850 Hz. The pulsation index is essentially the same during the VF and PEA periods.](image)

Figure 7: Histogram of pulsation indices for all experiments done on two animals.
an organized rhythm, but the ABP did not, indicating the presence of a Pulseless Electrical Activity (PEA). It can be seen that both during VF and during the transition to PEA, the pulsation index was low, showing the potential for this index to be indicative of the pulse state.

Figure 7 shows histogram of the pulsation index values computed for multiple experiments for two animals. The flow data measurements were obtained from several experiments and at several time points from each experiment during the normal and recovery periods. The no flow data measurements were obtained from several time points from a single experiment where the animal was put in a state of VF for several continuous minutes. For Animal A, the flow and no flow cases could be easily distinguished. The data were sufficiently separated to enable differentiating between the two cases. Similar results were seen for Animal B, although some overlap between the two cases was seen.

6. Conclusions and future work

In this work we proposed an ultrasound-based solution to replace the unreliable manual pulse check used in resuscitation. Limitations of previous Doppler measures were described and a new pulsation index measure was proposed and shown to work in animal studies. The pulsation index described in this work is very promising as a measure of the pulsatile behaviour of the flow, and consequently as a relevant index for assessment of pulse state. We expect that the pulsation index would be a robust measure for assessing the flow condition.

For simplicity we considered only sinusoidal type of pulsatility measure. Therefore there exists potential to improve the method to detect a more general (non-sinusoidal) periodicity in the Doppler power within a frequency band.

The proposed method needs to be tested in more animal studies, from which statistical analysis in terms of sensitivity and specificity would be evaluated.

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References