A System for Continuous Estimating and Monitoring Cardiac Output via Arterial Waveform Analysis

Vakily A.[1], Parsaei H.[1]*, Movahhedi M. M.[1], Sahmeddini M. A.[2]

[1]Department of Medical Physics and Engineering, Shiraz University of Medical Sciences, Shiraz, Iran
[2]Shiraz Anesthesiology and Critical Care Research Center, Shiraz University of Medical Sciences, Shiraz, Iran

ABSTRACT

Background: Cardiac output (CO) is the total volume of blood pumped by the heart per minute and is a function of heart rate and stroke volume. CO is one of the most important parameters for monitoring cardiac function, estimating global oxygen delivery and understanding the causes of high blood pressure. Hence, measuring CO has always been a matter of interest to researchers and clinicians. Several methods have been developed for this purpose, but a majority of them are either invasive, too expensive or need special expertise and experience. Besides, they are not usually risk free and have consequences.

Objective: Here, a semi-invasive system was designed and developed for continuous CO measurement via analyzing and processing arterial pulse waves.

Results: Quantitative evaluation of developed CO estimation system was performed using 7 signals. It showed that it has an acceptable average error of (6.5%) in estimating CO. In addition, this system has the ability to consistently estimate this parameter and to provide a CO versus time curve that assists in tracking changes of CO. Moreover, the system provides such curve for systolic blood pressure, diastolic blood pressure, average blood pressure, heart rate and stroke volume.

Conclusion: Evaluation of the results showed that the developed system is capable of accurately estimating CO. The curves which the system provides for important parameters may be valuable in monitoring hemodynamic status of high-risk surgical patients and critically ill patients in Intensive Care Units (ICU). Therefore, it could be a suitable system for monitoring hemodynamic status of critically ill patients.

Keywords

Arterial Pulse Waveform, Cardiac Output Estimation, Stroke Volume

Introduction

Cardiac output (CO), expressed in liters/minute, is the amount of blood pumped by heart in one minute. Mathematically, CO is the product of heart rate and the stroke volume [1-8]. The heart rate, in beats/min, is the number of beats per minute and the stoke volume is the volume of blood, in milliliters (mL), pumped out of the heart on each beat.

Cardiac output depends on cardiovascular system parameters and cardiac or even extra–cardiac factors. Also, CO is a fundamental determinant of oxygen delivered to tissues and is an essential indicator of how well the heart can meet the demands of the body. Therefore, CO is one

*Corresponding author: H. Parsaei
Department of Medical Physics and Engineering, Shiraz University of Medical Sciences, Shiraz, Iran
E-mail: hparsaei@sums.ac.ir

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of the most important parameters for monitoring cardiac function, estimating global oxygen delivery and for understanding the causes of high blood pressure. In other words, CO is a critical factor for monitoring hemodynamic status of patients [1-11]. Ultimately, measuring cardiac output has always been a matter of interest to researchers and clinicians.

The methods developed so far for measuring CO can be categorized as : Flowmetry [12, 13 and 14], Fick [13-16], the relative exhaled carbon dioxide [15, 16 and 17], Thermo dilution [12, 13, 14 and 18], Esophageal Doppler ultrasound [26, 28], Esophageal Doppler [17, 18, 19, 27-31], bio-electrical impedance [32-34], plethysmography impedance [33] and arterial analysis method [12, 13, 14, 35-43]. Most of these methods are either invasive or too expensive and need special expertise or experience; hence, they are not risk free. For example, Fick and Thermo dilution methods are both clinically possible, but they are invasive methods and could only be conducted in an equipped environment such as Intensive Care Units (ICUs) and the cardiac catheterization laboratories. In addition, these methods require the injection of cold saline or dye into a big vessel in which the whole cardiac output is flowing in. In this method, Swan - Ganz Catheter is used which enters the right side of heart through the inferior vena cava and then the pulmonary artery, but it is both invasive and needs special expertise and experience. In Fick method, the measurement of mixed venous oxygen needs blood sample from the pulmonary artery which is an invasive method. Doppler ultrasound method is non-invasive and accurate, but needs expensive equipment and the operator must be an expert. Therefore, to facilitate the conduction and to prevent the side effects of the invasive methods for measuring the cardiac output, a system was designed to measure the cardiac output in a non-invasive way (with the least invasion) through the arterial pulse wave. The procedure and the system evaluation method have been explained below.

Methods

The developed CO estimation system is generally based on the theory that blood pressure in an artery is related to the Stroke Volume (SV). The relation between SV, blood pressure and vascular resistance (Z) is as follows [12, 13, 14, 35-43]:

$$SV = \frac{\int_{ejection} \left[P_A(t) - P_D \right] dt}{Z} \tag{1}$$

Where $P_A(t)$ is the arterial pressure at time t and P_{TD} is the arterial pressure at the end of diastole. In fact, this integral calculates the area under systolic portion of the arterial pressure waveform, from the end-diastole to the end of the ejection phase; this corresponds to stroke volume. Figure 1 shows an example of the area under the curve calculated by the integral used in Equation 1.

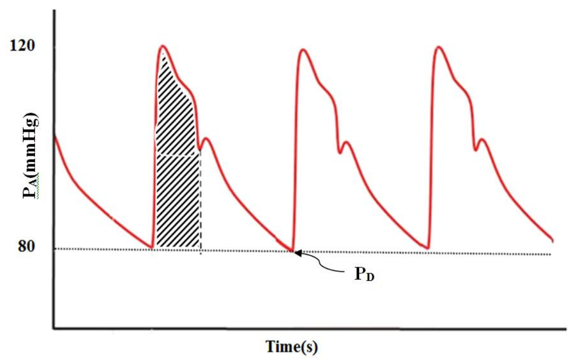

Figure 1: The area under the wave of the arterial pressure curve for estimating the stroke volume

Pressure changes in the artery are obtained via analyzing arterial pulse; details will be explained below. Through beat–to–beat analysis of the arterial pressure waveform, the cardiac output can be measured continuously. An important advantage of this approach is that the patient does not require endotracheal intubation or sedation to be monitored.

Most steps involved in the developed system are processing and analyzing the arterial pressure waveform. Figure 2 illustrates the main steps of the developed system. In short, from a complex arterial pulse, we find the starting point and the Dicrotic notch corresponding to the systolic portion of the arterial pressures, then, the area under the curve between these two points is calculated. The area under the curve is proportional to the stroke volume; if it is multiplied by the heart rate, then we can calculate the cardiac output. Detailed development of each step is given below.

Signal Acquisition

The signal is recorded using a blood pressure transducer which is attached to the arterial lane (radial artery, brachial or femoral). In this method, at first an arterial route is created by a catheter 22 (arterial line), and then the arterial lane is connected to the transducer. Then, mechanical pulses of the heart beat change into electrical signals through the diaphragm and the life signals are transmitted to the monitor. Finally, by connecting the monitor to a computer, the signals are stored.

Signal Pre-processing

Although estimating the area under curve seems easy, the accurate estimation of CO is difficult due to many artifacts that can affect the signal. Arterial waveform of the signal often interferes with different noises such as the power line noise, base line wandering noise and EMG signals. These noises are often created as a result of electrical signal fluctuations, motion or breathing artifacts [11, 14 and 15]. Extracting information from the arterial waveform relies on a clean and noise-free signal; otherwise, it can lead to inaccurate results and may mislead physicians. Therefore, before using any algorithms to extract clinical information from the arterial waveform, it is necessary to remove the unusual waveforms and noise signals. The purpose of this step is to remove noise from an acquired arterial pressure signal.

Baseline wondering noise is removed from a given signal by polynomial fitting technique; first the noise in the signal is estimated and then the estimated noise is subtracted from the signal. To estimate the noise level, a polynomial curve was adapted to representative time samples of the signal. The polynomial estimating the baseline was fitted by requiring it to pass through the sample of the signal. Ultimately, the baseline wondering noise was removed from the signal by simply subtracting the obtained polynomial curve from the signal. An example of removing baseline wondering noise is shown in Figure 3.

Other noises and artifacts in the signal are removed by filtering; various low pass filters including median filtering, the Kalman filter and Savitzky-Golay filter were studied. Frequency limit of the arterial waveform is up to 100 Hz range; thus, the filter cutoff frequency was also adjusted in this range. The best and most efficient filter was chosen using a simulation experiment.

Feature Extraction

The objective in this step was to extract im-

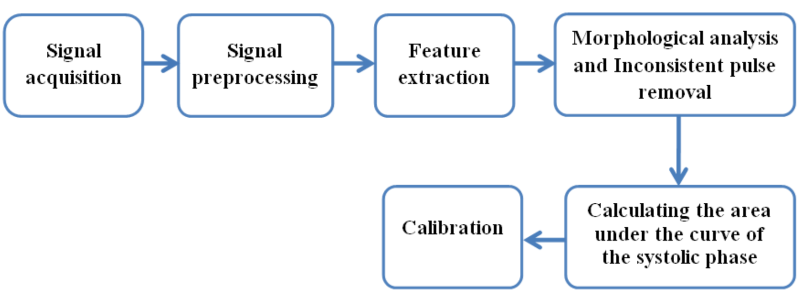

Figure 2: Steps of the developed cardiac output estimating system

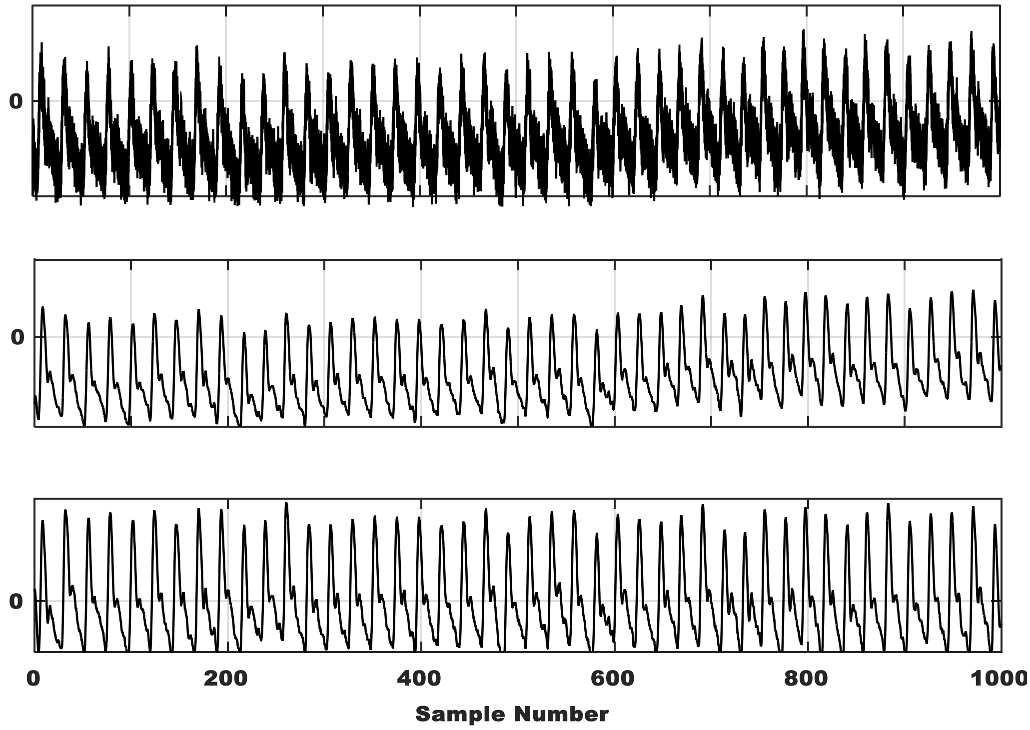

Figure 3: An example of removing noise from an acquired arterial pressure signal. (First row) Raw signal. (The second row) Signal after removing noise using Savitzky-Golay filter, (The thrid row) Signals after base line correction.

portant features for each individual beat of arterial pulse waveform for next steps. The following three features were employed (Figure 4): total pulse duration (T), systolic amplitude (AS) and diastolic amplitude (AD) of the pressure waveform.

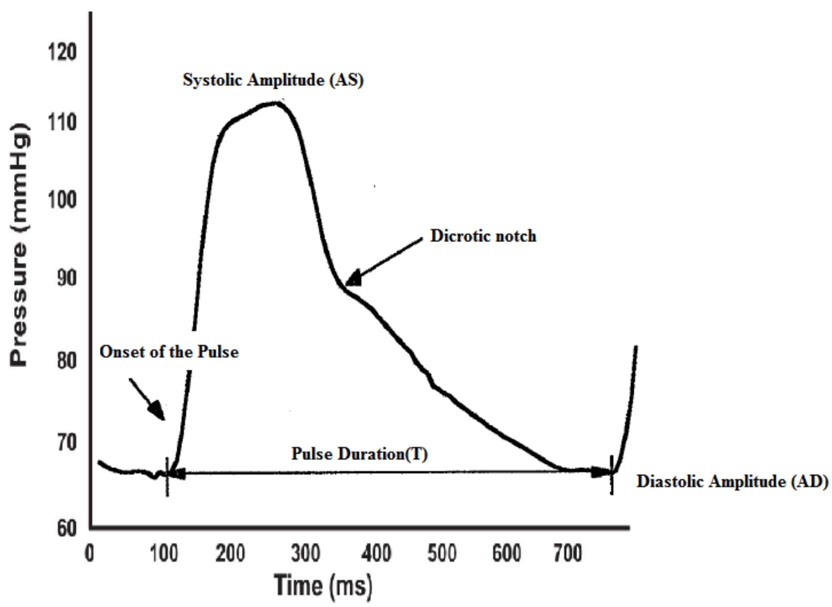

Figure 4: The characteristics of the arterial pulse wave used in this study modified from [35]

Total Pulse Duration (T) is the distance between the beginning and the end of pulse waveform. Onset of the pulse was determined using the first and second derivatives of the signal as it was discussed in [35, 44 and 45]; the first derivative of the pulse was scanned and the onset of the signal was considered as the first zero crossing point from negative to positive that led to the maximum point on the first derivative curve (see [35, 45] for more details).

Systolic Amplitude (AS) is the peak pressure of the waveform which is simply determined by scanning the pulse to find its maximum value.

Diastolic Amplitude (AD) is the trough of pressure waveform that is determined by finding the minimum value of the pulse

Morphological Analysis and Inconsistent Pulse Removal

The objective in this step is to flag inconsistent beats and remove them from further analyses. The inconsistent pulses are pulses that are not consistent with physiological characteristics of the cardiovascular system. For instance, the systolic domain could never be three times more than the average systolic domain [1-3]. Likewise, diastolic amplitude of a pulse should be positive. The criteria used in this study for identifying inconsistent beats are listed in Table 1. This step is completed using the morphological features calculated for each pulse in the previous step.

Calculating the Curve of Systolic Phase of Arterial Waveform

In this step, the area under the systolic portion of the arterial pressure waveform is calculated. The systolic portion is from the end-diastole to the end of the ejection phase; therefore, for estimating this area, the location of ventricular ejection and the closure of the aortic valve on each pulse should be identified. The location of ventricular ejection corresponds to the onset of signal that was specified in sub–section 2–3. The location of the closure of the aortic valve on arterial pressure waveform corresponds to a point called dicrotic notch. Dicrotic notch point in each pulse was determined by using the algorithm discussed in [35]. It is considered to be at the first point after the position of the peak at which the sign of the 3rd derivative of the pulse changes. Finally, after identifying the onset point and the dicrotic notch point, the area under the curve was calculated using the trapezoidal method.

To decrease the effect of variation in the arterial pressure waveform on estimating CO, area under the curve was estimated for a template of pulses detected over 10-sec cycles. The pulses were detected every 10–sec of the signal and it was segmented, aligned based on their peaks and then averaged to estimate a representative of these pulses.

Table 1: Criteria for identifying abnormal pulses (unacceptable). *AS* is the systolic domain (peak of the pulse wave form), *AM* is the average systolic domain, *AD* is the diastolic domain (least diastolic pulse), *T* is the width of the pulse and *TM* is the mean pulse width.

Feature	Criteria
Systolic Amplitude (AS)	AS > 3AM
Mean Systolic Amplitude (AM)	AS < 0.5AM
Diastolic Amplitude(AD)	AD<0
Pulse Width (T) Mean Width of all pulses(TM)	T>1.5TM

Calibration

As it was mentioned in equation (1), the stroke volume has an inverse relation with the coronary vascular resistance. Therefore, in addition to the area under the systolic portion of the arterial pressure waveform indicated in Figure 1, the vascular resistance is required for calculating the stroke volume and the cardiac output. This parameter varies from patient to patient, thus it should be set for each patient individually. However, direct measurement of this parameter is difficult. In this study, this parameter was taken into account as a calibration coefficient (K), as follows:

$$CO = K \times HR \times SV \quad (2)$$

Where HR is the number of heart rate per minute and SV is the stroke volume which is calculated by using equation 1, and the methods explained in sub–sections 1 to 6.

Based on equation 2, the developed system must be calibrated for each patient. In other words, parameter K has to be estimated for each patient, individually. In this study, this parameter was estimated in the first few seconds of recording the signal using CO value estimated via thermodilution technique.

Software Development

Our goal in this work was to present an interactive system that works as an interactive environment for continuously measuring CO through developing a MATLAB interactive software package. The purpose of interaction in the developed system is to facilitate the use of a computer for estimating, analyzing and storing CO and several parameters that may be valuable in monitoring hemodynamic status of high-risk surgical patients and critically ill patients in Intensive Care Units (ICU).

The interactive environment consists of tasks involving the user and the interactive system. These tasks can take place in a sequential order. The developed software package provides graphical user interfaces (GUIs) to filter an acquired arterial pressures signal, extract relevant features for each beat, estimate important parameters, calibrate the system, store data and reload and reanalyze a signal.

Using this system, arterial pressure signals are acquired from a patient monitor used to screen patient parameters and then several parameters such as the cardiac output, systolic blood pressure, diastolic blood pressure, mean blood pressure, heart rate and the stroke volume are continuously estimated and displayed as a numerical value. In addition, this software provides a time series graph for each of these parameters that assist in tracking changes of each parameter and ultimately better management of patient (see Result section for example outputs of the software).

In this software, a color alarm system is provided so that the background color of the monitor changes to red when the cardiac output is less than 2 liters /minute, to yellow when the cardiac output is between 2 and 5 liters, and it changes to green if the cardiac output is more than 5 liters. Thus, this option enables the operator to use his/her vision in addition to the sense of hearing in noisy environment. Figure 5 shows an example output of the designed software.

Results and Discussions

The performance of the designed CO estimation system was evaluated using MIMIC data available at MIBH website [46]. This database includes arterial pressures signal and CO of several patients. The reported CO values were estimated using thermodilution method. The provided values for CO were used as gold standard in this study.

The measured CO values and CO values estimated using developed system are summarized in Table 2. In this table only the results for the signals, that the CO values estimated by using thermodilution method were reported. As it can be seen, this system with an average error of 6.5% has an acceptable efficacy in estimating cardiac output. Statistical analysis using student t-test (α=0.05) also proved that there was no significant difference between CO values estimated using the developed system and those measured using thermodilution method.

The average estimation error of the developed system is negative which shows that the system underestimates CO; nevertheless, overall there is no significant difference between the estimated CO values and gold standard values.

The estimation error for signal #4 is relative-

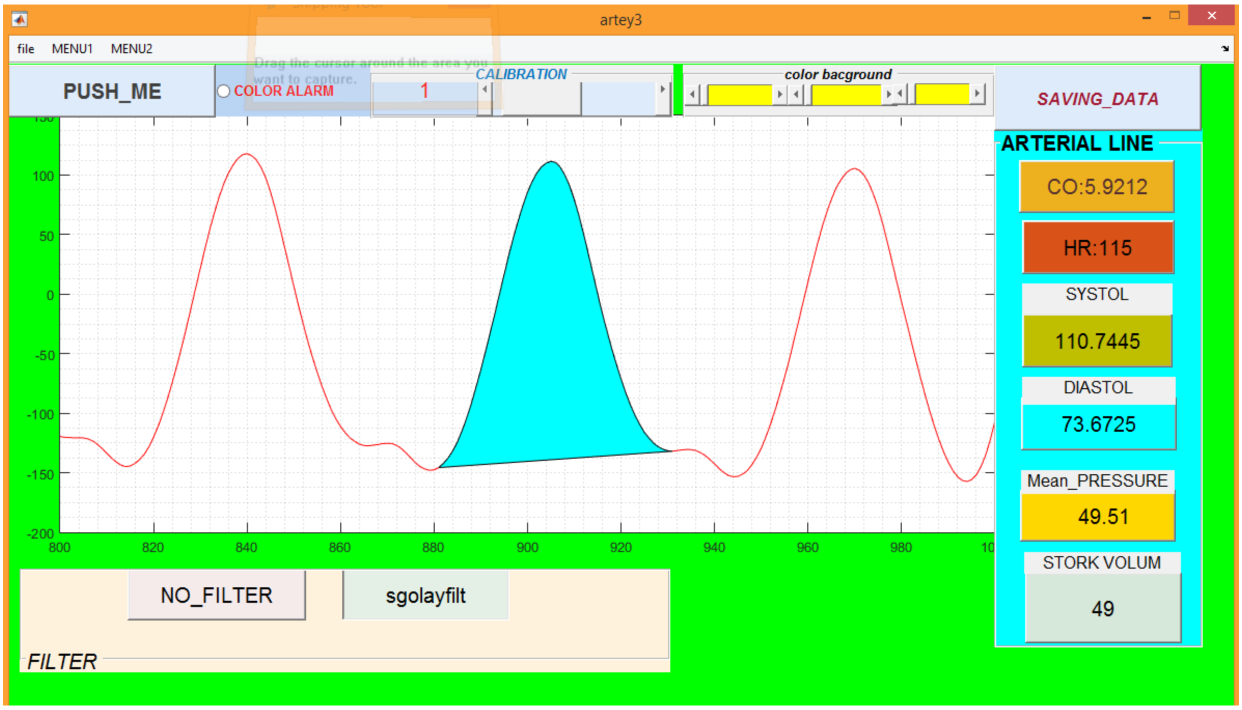

Figure 5: Example output of the developed CO estimation system. The values estimated for each critical parameter are the right column. The method for calculating the area under the curve can be seen.

Table 2: Cardiac output values estimated using the developed system (estimated CO) compared with those measured using the thermodilution method (CO).

Signal#	CO	Estimated CO	Error(%)
1	4.7	4.8	1.7
2	4.3	4.2	-1.4
3	4.5	4.5	-1.6
4	3.6	4.5	26.1
5	5.0	5.5	9.3
6	3.2	3.5	9.0
7	7.1	7.9	11.1
Mean	4.6	4.3	-6.5
STD	1.3	1.4	

ly large, this may be due to the fact that this signal is very noisy and the system could not detect Dicrotic notch correctly.

As it was discussed earlier, by using this system, we can track the changes in cardiac output, systolic blood pressure, diastolic blood pressure, average blood pressure, heart beat and the stroke volume of a patient via time series graphs provided for these parameters. A sample of these results is presented in Figure 6.

Conclusion

Cardiac output is a critical factor for monitoring hemodynamic status of high-risk surgical patients and critically ill patients in Intensive Care Units (ICU), because it can be used for monitoring cardiac function, estimating global oxygen delivery and understanding the causes of high blood pressure. In this paper, a system for estimating and monitoring CO is presented. The system consists of five main steps: signal acquisition, signal preprocessing, feature extraction, morphological analysis and inconsistent pulse removal, area under the systolic portion calculation and calibration. Performance analysis using 7 real signals demonstrated that CO values provided by the system

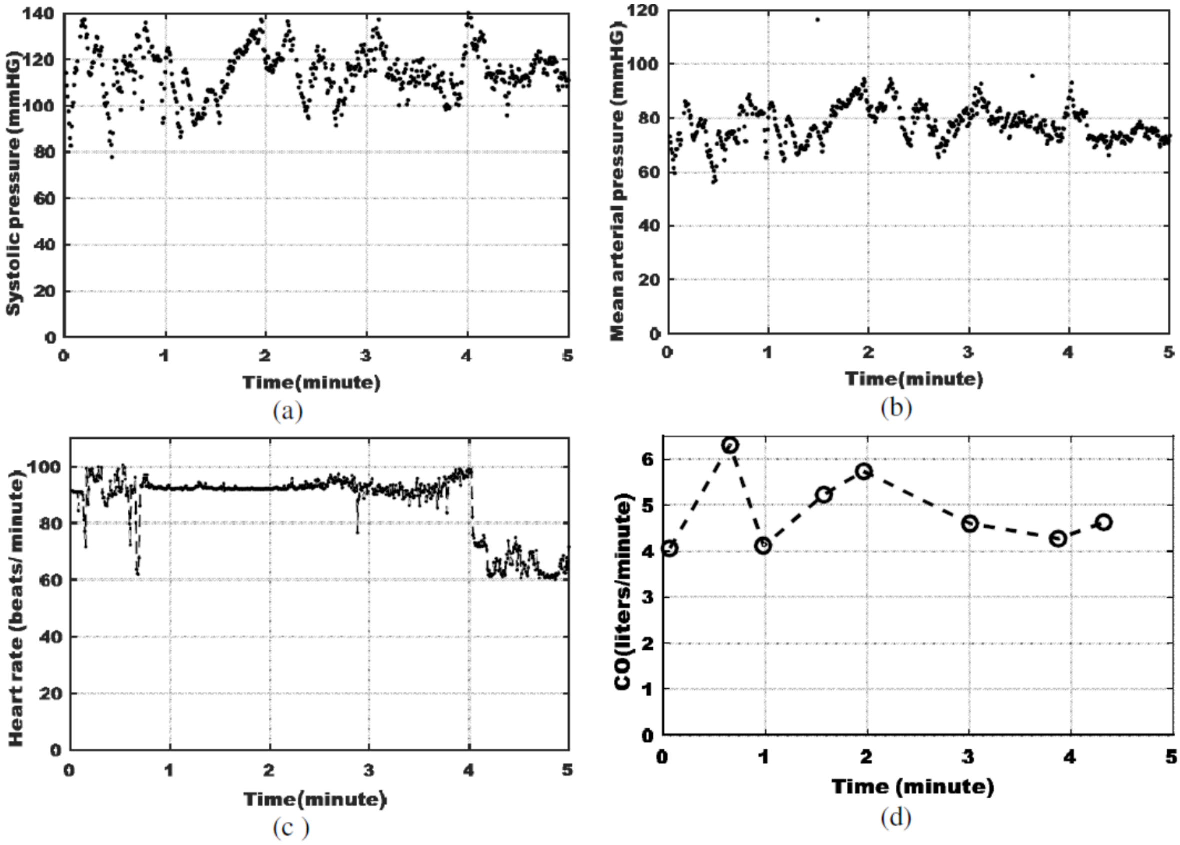

Figure 6: An Example output of the developed system that can be useful for tracking a patient important parameters. Systolic pressure (a), mean arterial pressure (MAP) (b), heart rate (c), and cardiac output (d) versus time.

are not statistically different from the values obtained using thermodilution method. The system has several advantages including being minimally invasive, easy to use, not expensive and finally providing continuous recording of CO and several important parameters for tracking these parameters and ultimately patient's conditions. Consequently, the developed system could be a suitable system for monitoring hemodynamic status of high-risk surgical patients and critically ill patients in Intensive Care Units (ICU).

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Conflict of Interest

None

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