Continuous Cardiac Output Monitoring by Peripheral Blood Pressure Waveform Analysis

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Abstract

We introduce a novel technique for continuously monitoring changes in cardiac output (CO) by mathematical analysis of a single peripheral arterial blood pressure (ABP) waveform. In contrast to all previous techniques, our technique analyzes ABP variations over time scales greater than a cardiac cycle in which wave reflections are attenuated. To validate the technique, we performed six swine experiments in which peripheral ABP waveforms and gold standard CO via an aortic flow probe were simultaneously measured over a wide physiologic range. We report an overall CO measurement error of only 15.0%.

1. Introduction

Cardiac output (CO) is perhaps the most fundamental index of cardiovascular function. Numerous methods have been developed for its measurements. However, all of the conventional methods have proven to be far from ideal ($e.g.$, highly invasive, inaccurate, discrete, and/or expensive) [1]. On the other hand, several ideal, or near ideal, methods exist for continuously measuring peripheral arterial blood pressure (ABP) such as radial artery catheterization.

Since 1904, investigators have proposed analysis techniques to monitor CO from ABP waveforms [2]. Much of the earlier work assumed that the arterial tree is well represented by a two-parameter Windkessel model accounting for the compliance of the large arteries (AC) and the total peripheral resistance (TPR) of the small arteries. While techniques based on this model generally failed when applied to ABP waveforms measured centrally in the aorta ($e.g.$, [3]), Bourgeois *et al* demonstrated that their technique yielded a quantity which varied linearly with aortic flow probe CO over a wide physiologic range [4]. The key concept of their technique is that, according to the Windkessel model, ABP should decay like a pure exponential during each diastolic interval with a time constant (τ) equal to the product of TPR and AC. Since AC is nearly constant over a wide pressure range and on the time scale of months [4, 5], CO could then be measured to within a constant scale factor equal to 1/AC by dividing the time-averaged ABP with τ. Thus, the technique of Bourgeois *et al* involves fitting an exponential function to the diastolic interval of an ABP wavelet so as to measure τ.

Bourgeois *et al* were able to validate their technique with respect to central ABP waveforms, because the diastolic interval of these waveforms can *some times* resemble an exponential decay (Figure 1A). However, central ABP is rarely measured clinically because of the difficulty in inserting a catheter retrogradely via a peripheral arterial blood vessel and the risk of blood clot formation and embolization. Moreover, in readily available peripheral ABP waveforms, an exponential diastolic decay is not apparent (Figure 1B). The reason is that the arterial tree is not a lumped system like the Windkessel model suggests but rather a distributed system with impedance mismatches throughout the system due to vessel tapering, bifurcations, and caliber changes. Thus, the diastolic (and systolic) intervals of peripheral ABP waveforms are corrupted by wave reflections that occur at every site of impedance mismatch. The technique of Bourgeois *et al* thus cannot be applied to readily available peripheral ABP waveforms.

More recent techniques have attempted to monitor CO from ABP waveforms measured peripherally. Techniques based on an adaptive aorta model, which require ABP waveforms measured at two peripheral sites, have been proposed ($e.g.$, [6]). Learning techniques, which require large training data sets consisting of simultaneous measurements of CO and ABP waveforms obtained over the entire physiologic range, have also been suggested ($e.g.$, [7]). Finally, Wesseling *et al* [8] and Linton *et al* [9] have proposed techniques which require only the analysis of a single radial ABP waveform. However, Linton *et al* showed that their heuristic technique was accurate only over a narrow physiologic range, and several studies have demonstrated the inadequacy of the technique of Wesseling *et al* ($e.g$, [10]).

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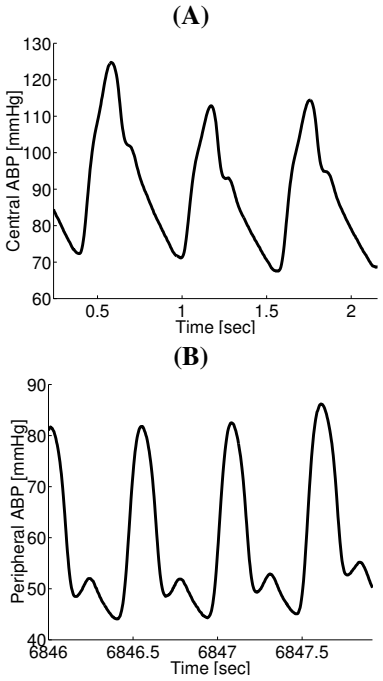

Figure 1. Examples of swine (A) central and (B) "radial" ABP waveforms.

In this paper, we introduce a fundamentally new technique for continuously monitoring CO changes from a single peripheral ABP waveform. To demonstrate the validity of the technique, we present results from six swine experiments in which peripheral ABP waveforms and gold standard CO via an aortic flow probe were simultaneously measured over a wide physiologic range.

2. The technique

The common feature of all previous techniques for monitoring CO from continuous ABP is that the waveform analysis is employed only on time scales within a cardiac cycle. Because of the presence of wave reflections at these time scales, these techniques were limited in that they 1) could only be applied to central ABP waveforms in which the cumulative effects of the wave reflections may be attenuated; 2) necessitated multiple peripheral ABP waveform measurements; 3) required a large training data set obtained over the entire physiologic range, and/or 4) are accurate only over a narrow physiologic range. However, the contribution of wave reflections to ABP waveforms diminishes with increasing time scales. That is, the wave reflections significantly corrupt peripheral ABP waveforms on short time scales (high frequencies) leaving the waveform on longer time scales (low frequencies) relatively undisturbed. For example, consider the limiting case in which the time scale is sufficiently long such that the wavelengths of the propagating waves are much larger than the dimension of the arterial tree. At such time scales, the arterial tree acts as a single blood reservoir, and the Windkessel model is therefore valid. This important concept implies that if pulsatile activity ceased, then a peripheral ABP waveform may eventually decay like a pure exponential as soon as the faster wave reflections vanish. Furthermore, the arterial tree is being continuously excited at time scales greater than a cardiac cycle ($e.g.$, breathing, baroreflex). Thus, there is significant long time scale information in all ABP waveforms.

Our technique therefore analyzes a peripheral ABP waveform (sampled at 90 Hz) over six-minute intervals so as to establish the pure exponential decay that would eventually result if pulsatile activity ceased. More specifically, the ABP response to a single, solitary cardiac contraction is estimated from the ABP waveform. Then, τ is measured by fitting an exponential to the tail end of this response once the faster wave reflections have vanished. Finally, proportional CO is computed via Ohm's law. The determination of τ, which is pictorially represented in Figure 2, is specifically achieved as follows.

First, a cardiac contractions signal is constructed through the formation of an impulse train in which each impulse is located at the onset of upstroke of an ABP wavelet and has an area equal to the ensuing pulse pressure (PP). PP is determined as the maximum ABP value minus the ABP value at the onset of upstroke. Then, the relationship between the cardiac contractions signal ($x(t)$) and the ABP waveform ($y(t)$) is characterized by estimating an impulse response function ($h(t)$) which when convolved with $x(t)$ "best" fits $y(t)$. By definition, the estimated $h(t)$ represents the ABP response to a single, solitary cardiac contraction (normalized approximately by the average PP). The impulse response function is specifically estimated according to the following autoregressive moving average equation:

$$y(t) = \sum_{k=1}^{m} a_k y(t-k) + \sum_{k=1}^{n} b_k x(t-k) + e(t), \qquad (1)$$

where $e(t)$ is the unmeasured residual error, the parameters $\{a_k, b_k\}$ completely specify $h(t)$, and m and n limit the number of parameters (model order) [11]. The parameters are estimated from $x(t)$ and $y(t)$ through the least-squares minimization of $e(t)$ which has a closed-form solution, while the model order is selected by minimizing the Minimum Description Length criterion [11]. Finally, τ is determined by fitting an exponential function with two adjustable parameters (amplitude and τ) over the interval of $h(t)$ ranging from two to four seconds following the time of its maximum value. In theory, accurate determination of τ is achieved by virtue of $h(t)$ coupling the long time scale variations in $x(t)$ to $y(t)$.

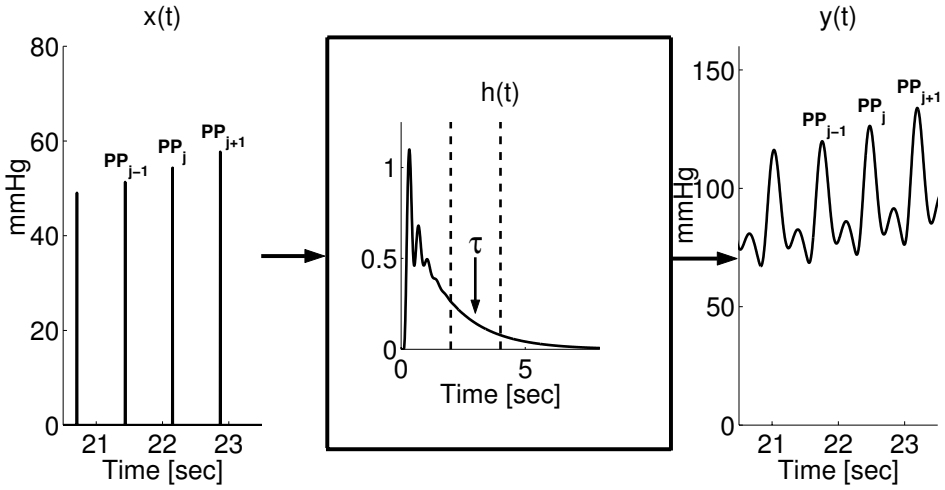

Figure 2. An illustration of how the proposed technique determines τ from a peripheral ABP waveform.

3. Methods

To validate the technique, we conducted experiments in six Yorkshire swine (30-34 kg) under a protocol approved by the MIT Committee on Animal Care. The animals were anesthetized with inhaled isoflorane, and mechanical ventilation at a rate of 10-15 breaths/min and a tidal volume of 500 ml was employed. Catheters were placed in the femoral artery for femoral ABP and as distal as possible to the brachial artery for "radial" ABP. The chest was opened with a midline sternotomy, and an ultrasonic flow probe was placed around the aortic root for gold standard CO. In each animal, a subset of the following interventions was performed over the course of 75 to 150 minutes to vary CO and other hemodynamic parameters: infusions of volume, phenylephrine, dobutamine, isuprel, esmolol, nitroglycerine, and progressive hemorrhage. To compare the proportional CO values estimated from the ABP waveforms with the absolute, gold standard CO values determined from the aortic flow probe, we first scaled the former values to have the same mean as the latter values for each animal. We then computed the standard deviation (SD) of the CO errors (normalized by the gold standard CO values and given in percent) as a metric for comparison.

4. Results

Table I summarizes the "radial" ABP waveform results for each animal, and Figure 3 is an example of the corresponding trends for animal 4. The table and figure illustrate the wide physiologic range considered and strong agreement between the estimated and gold standard CO trends (in terms of CO Error SD and visually). This strong agreement was confirmed by a high overall correlation coefficient (scale-invariant metric) between the estimated and gold standard CO trends (0.84 ± 0.07). Moreover, the CO errors were largely uncorrelated with the values of CO ($\rho = 0.11$), mean ABP (MAP) ($\rho = 0.47$), and heart rate (HR) ($\rho = -0.08$). Similar results were obtained with the femoral ABP waveforms.

Animal	CO Range [L/min]	MAP Range [mmHg]	HR Range [bpm]	CO Error SD [%]
1	1.6 - 5.2	29 - 100	96 - 180	19.1
2	2.3 - 4.2	54 - 127	101 - 204	16.0
3	1.9 - 5.8	70 - 120	96 - 186	16.7
4	1.3 - 4.3	27 - 106	103 - 198	12.3
5	2.4 - 5.0	65 - 118	91 - 198	8.0
6	2.3 - 5.6	52 - 108	109 - 177	14.7
Total	1.3 - 5.8	27 - 127	91 - 204	15.0

Table 1. Summary of results.

5. Summary and conclusions

We developed and validated a novel technique for continuously monitoring CO by the analysis of a single peripheral ABP waveform. In contrast to all previous techniques, our technique analyzes ABP variations over time scales greater than a cardiac cycle in which the wave reflections are attenuated. The technique specifically measures CO to within a proportionality constant and may be utilized to monitor quantitatively changes in CO. The proportional CO values may be calibrated, if desired, with a single, absolute measure of CO ($e.g.$, thermodilution). With further testing, the technique may potentially be employed in intensive care units and surgical suites in which invasive radial ABP waveforms are routinely measured or in primary care settings, emergency rooms, and regular hospital beds in which non-invasive measurements of ABP waveforms ($e.g.$, arterial tonometry) or related signals ($e.g.$, fingertip photoplethysmography) could easily be obtained. Finally, the technique is general and may be applied to

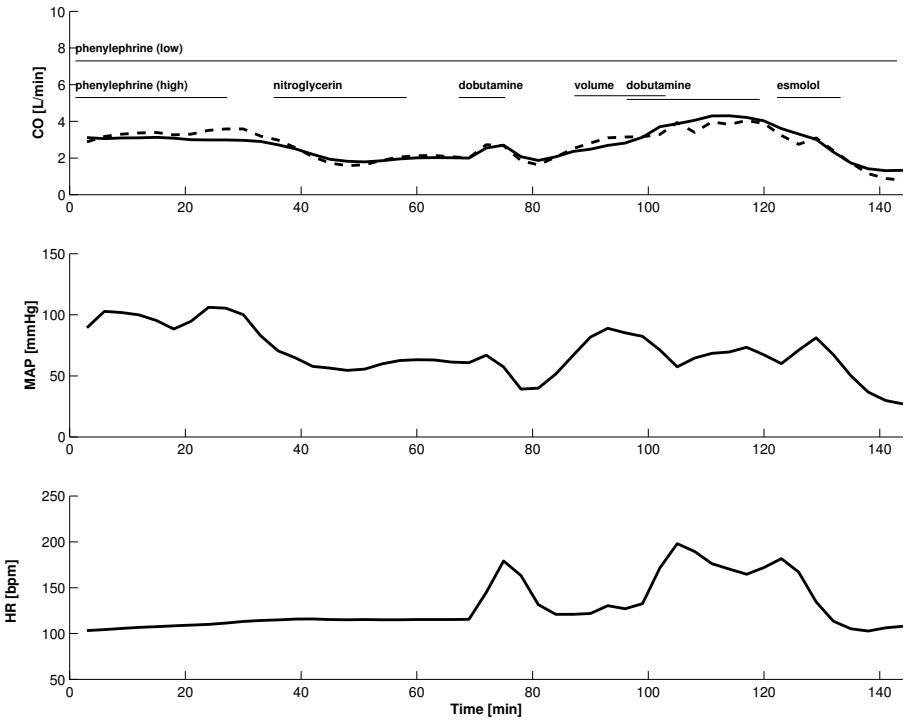

Figure 3. Results from animal 4. The top panel illustrates the trended CO determined with an aortic flow probe (solid) and estimated (*and calibrated*) from the "radial" ABP waveform (dash) along with the experimental intervention and duration (underline). The bottom two panels depict the corresponding trended MAP and HR.

ABP waveforms measured at any site in the systemic or pulmonary arterial tree.

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