Proceedings of the 28th IEEE
EMBS Annual International Conference
New York City, USA, Aug 30-Sept 3, 2006

A Novel Approach to Pulse Pressure Variation Estimation

Daniel Austin, *Student Member, IEEE,* Christian Staats, Mateo Aboy*, *Member, IEEE*

***Abstract*— We describe a novel algorithm to estimate the pulse pressure variation index (PPV) from arterial blood pressure signals (ABP). PPV has been shown to be one of the best predictors of fluid responsiveness in mechanically ventilated subjects. Our PPV algorithm uses a non-linear technique for envelope estimation, eliminating the need for automatic beat detection. Additionally, the algorithm makes use of nonparametric spectral techniques to extract the respiratory rate, and a median filter for artifact removal. The algorithm was validated against the continuous PPV output obtained from the commercially available PiCCO® system and gold standard expert PPV manual annotations. The data consists of ABP taken from subjects who experienced rapid changes in hemodynamics. This data comprised over six hours of continuous ABP monitoring.**

***Index Terms*— Pulse pressure variation index (PPV), stroke volume variation index (SVV), PiCCO® , pulse contour analysis, hemodynamic monitoring, fluid responsiveness.**

I. Introduction

We describe a new algorithm for PPV estimation from ABP signals. PPV has been shown to have strong correlation to fluid responsiveness in mechanically ventilated subjects with acute respiratory failure related to sepsis [1]. Another study indicated that PPV was a good predictor of change in stroke volume index related to fluid replacement in subjects before off pump coronary artery bypass grafting [2]. Numerous other studies have indicated that PPV is one of the most reliable predictors of fluid responsiveness in mechanically ventilated subjects [3]–[11].

Currently, there are few algorithms publicly available for PPV estimation. In [12] a method is proposed, but this algorithm was validated only on synthetic signals. In addition, this method was based on automatic beat detection algorithms. The other available options for PPV estimation are commercial devices that offer pulse contour analysis and continuous hemodynamic monitoring. Unfortunately, the algorithms used in these devices are proprietary and lack the capability of analyzing data that has been prerecorded.

In this paper we describe an algorithm that avoids the use of automatic beat detection algorithms [13]. We compare our algorithm against both "golden-standard" data points annotated by trained experts and the continuous PPV output of the PiCCO® system. We show that during periods of static hemodynamics, the PiCCO® output and the proposed algorithm are in agreement. However, during periods of rapid hemodynamic changes, the commercially available system fails to correctly track PPV. Our algorithm is shown to remain effective during periods of abrupt hemodynamic changes and robust to artifacts.

D. Austin, C. Staats, and M. Aboy are with the Electronics Engineering Technology at Oregon Institute of Technology, Portland, OR 97006 USA. *Asterisk indicates corresponding author.* (E-mail: mateoaboyieee.org) .

II. Algorithm Description

In this section we provide a detailed description of the proposed PPV estimation algorithm.

A. Step 1: Envelope Extraction

The ABP signal is preprocessed to remove the mean and shift the signal according to the following relationships,

$$y_1(n) = x(n) - \frac{\sum_{k=l}^{L} x(k)}{(L-l)(1-\alpha)} \tag{1}$$

$$y_2(n) = x(n) - \frac{\sum_{k=l}^{L} x(k)}{(L-l)(1+\alpha)} \tag{2}$$

where $w = L-l$ is the user–specified window and $0 \leq \alpha < 1$ is a user specified parameter, corresponding to a shift about the x-axis. After the mean is removed, a nonlinear filter is applied to extract the envelopes. This nonlinear filter $\zeta\{\cdot\}$ is specified as follows,

$$e_u(n) = \begin{cases} 0 & \text{if } y(n) \leq 0 \\ y_1(n) & \text{if } y(n) \geq y(n-1) \\ y_1(n-1) & \text{otherwise} \end{cases} \tag{3}$$

$$e_l(n) = \begin{cases} 0 & \text{if } y(n) \geq 0 \\ y_2(n) & \text{if } y(n) \leq y(n-1) \\ y_2(n-1) & \text{otherwise.} \end{cases} \tag{4}$$

This nonlinear operation has many applications. It can be used to eliminate the dependence on automatic beat detection algorithms in situations where the exact time location of the detection point is not required. In our case, we used this nonlinear operation to estimate the upper and lower envelopes without automatic beat detection. The peak and trough values in the ABP signal are detected indirectly, removing the dependence on beat detection. This is because our algorithm does not require knowledge of the exact time indices corresponding to these points. Fig. 1 illustrates the application of this nonlinear operation. The next step involves adding the local mean according to:

$$e_u(n) = e_u(n) + \frac{\sum_{k=l}^{L} x(k)}{(L-l)(1-\alpha)} \tag{5}$$

$$l_e(n) = l_e(n) + \frac{\sum_{k=l}^{L} x(k)}{(L-l)(1+\alpha)}. \tag{6}$$

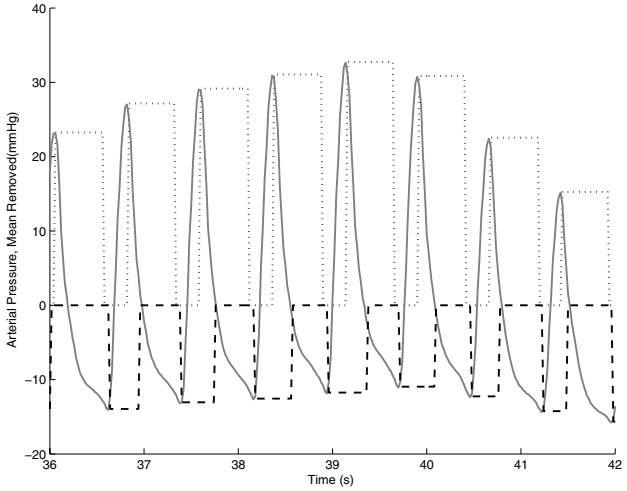

Fig. 1. Result of applying the non-linear operation $\zeta\{\cdot\}$ to extract the upper(dotted) and lower(dashed) envelopes.

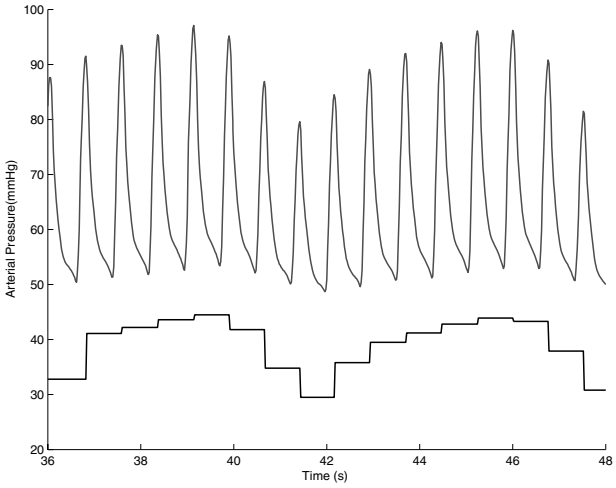

Fig. 2. Example of pulse pressure signal $p(n)$ during two respiratory cycles(bottom) and the corresponding ABP signal segment(top).

B. Step 2: Respiratory Rate Estimation

After the envelopes have been extracted, a pulse pressure signal $p_i(n)$ is constructed through an iterative process according to the following relationships:

$$p_i(n) = \begin{cases} e_u(n) - e_l(n) & \text{for } e_u(n) \neq e_l(n) \neq 0 \\ 0 & \text{otherwise} \end{cases} \tag{7}$$

and

$$p(n) = \begin{cases} p_i(n) & \text{if } p_i(n) \neq 0 \\ p_i(n-1) & \text{otherwise.} \end{cases} \tag{8}$$

An example of the pulse pressure signal $p(n)$ obtained after this operation is shown in Fig. 2 for illustration purposes. After obtaining the pulse pressure signal $p(n)$, the algorithm estimates the respiratory frequency according to:

$$fr = \frac{f_s}{N} \arg \max_{0 \leq k \leq \frac{N}{2}-1} \left| \sum_{n=0}^{L-1} \left(p(n) - \frac{\sum_{l=0}^{L-1} p(l)}{L} \right) e^{-j(\frac{2\pi}{N})kn} \right| \tag{9}$$

where L is the length of the user specified window, fs is the sampling frequency, and N corresponds to the number of points to evaluate the frequency (user-specified).

C. Step 3: Pulse Pressure Variation Estimation

A pulse pressure matrix $\mathbf{P} = (\mathbf{p}_1, \mathbf{p}_2, \ldots, \mathbf{p}_N)$ is created by segmenting the vector $\mathbf{p}$, where:

$$\mathbf{p}_i = (p(i), p(i+1), \ldots, p(i+m)) \tag{10}$$

with $m = \frac{f_s}{f_r} - 1$ and N equal to the number of segments in the signal. PPV is calculated from $\mathbf{P}$ according to:

$$\Delta p_j = 2 \frac{\max \mathbf{p}_j - \min \mathbf{p}_j}{\max \mathbf{p}_j + \min \mathbf{p}_j}. \tag{11}$$

Finally, the proposed algorithm uses a median filter of length w_l to smooth the Δp_j estimates and remove outliers due to artifacts.

III. Algorithm Development and Validation

The proposed algorithm was applied prospectively to 6 hours of ABP data obtained from subjects in the Animal Laboratory at the Oregon Health and Science University (Portland, OR, USA). These subjects experienced abrupt changes in hemodynamics due to severe blood loss. The algorithm was compared to the resulting PPV output from the PICCO® system (Pulsion Medical Systems, Germany) on the same subjects. Five "gold–standard" data points were manually annotated by trained experts. These gold–standard PPV calculations were annotated as follows: 1 immediately before the abrupt hemodynamic change corresponding to injury, 3 during the intermediate region between hemodynamic changes, and one following the hemodynamic change corresponding to subject recovery.

IV. Results and Discussion

Figs. 3 and 4 show the results of the proposed algorithm versus the commercially available PiCCO® system. Each figure shows the ABP signal from the subject (top), the PiCCO® system PPV output in light gray, the proposed algorithm results in dark gray, and the "golden-standard" points in black (bottom). As shown in both figures, during periods of relatively static hemodynamics, both the proposed algorithm and the PiCCO® system PPV estimates match. Note that the during the period of abrupt hemodynamic changes, the proposed algorithm tracks the PPV value much better than the commercially available device. In subject one (fig. 3), every "golden-standard" point was matched by the estimated value of the proposed algorithm. In subject two (fig. 4), four of the five "gold-standard points" were correctly estimated. Also, the single under-estimation was much closer than the PiCCO® system estimate. In addition, the artifacts present in the ABP signals in each subject tend to cause a failure in the commercially available system. Figs. 3 and 4, clearly show the proposed algorithm is more robust to

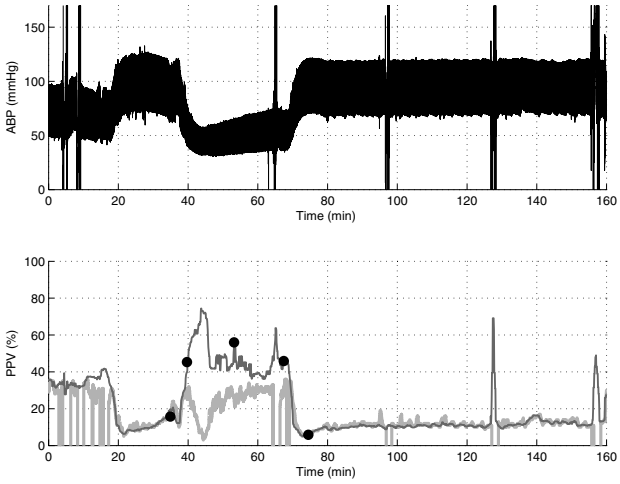

Fig. 3. Arterial Blood Pressure reading from subject 1 (top). PiCCO® PPV output in light gray, proposed algorithm PPV output in dark gray, and "golden-standard" points in black (bottom).

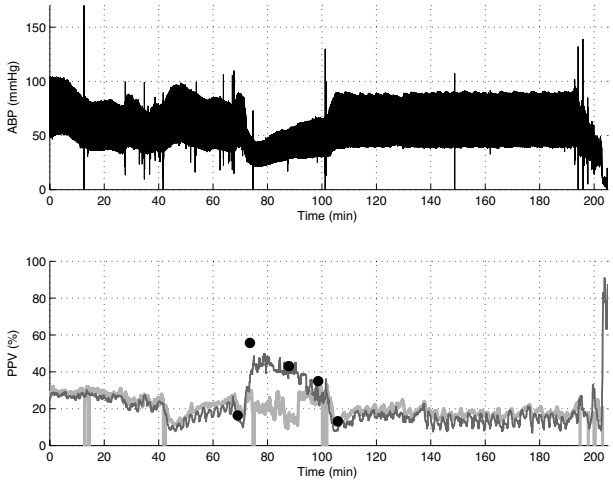

Fig. 4. Arterial Blood Pressure reading from subject 2 (top). PiCCO® PPV output in light gray, proposed algorithm PPV output in dark gray, and "golden-standard" points in black (bottom).

artifacts, and also provides a more accurate estimation of PPV. Finally, since the PiCCO® algorithm is proprietary, the proposed algorithm presents the advantage of being publicly available.

V. Conclusion

We provided a detailed description of a novel algorithm for PPV estimation. The algorithm was assessed on over six hours of ABP data, both against a commercially available system for hemodynamic monitoring and against "gold-standard" points annotated by trained experts. The results provided strong evidence to support the robustness and efficacy of the proposed algorithm. In addition, the ability of this algorithm to accurately estimate PPV during periods of drastic changes in hemodynamics was illustrated. This is in contrast to the inability of the commercially available system to track PPV during abrupt hemodynamic changes.

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