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# CENTRAL VENOUS PULSE PRESSURE ANALYSIS USING AN R-SYNCHRONIZED PRESSURE MEASUREMENT SYSTEM*

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**ABSTRACT. Objective.** The information derived from central venous catheters is underused. We developed an EKG-R synchronization and averaging system to obtained distinct CVP waveforms and analyzed components of these. **Methods.** Twenty-five paralyzed surgical patients undergoing CVP monitoring under mechanical ventilation were studied. CVP and EKG signals were analyzed employing our system, the mean CVP and CVP at end-diastole during expiration were compared, and CVP waveform components were measured using this system. **Results.** CVP waveforms were clearly visualized in all patients. They showed the *a* peak to be $1.8 \pm 0.7$ mmHg, which was the highest of three peaks, and the *x* trough to be lower than the *y* trough ($-1.6 \pm 0.7$ mmHg and $-0.9 \pm 0.5$ mmHg, respectively), with a mean pulse pressure of 3.4 mmHg. The difference between the mean CVP and CVP at end-diastole during expiration was $0.58 \pm 0.81$ mmHg. **Conclusions.** The mean CVP can be used as an index of right ventricular preload in patients under mechanical ventilation with regular sinus rhythm. Our newly developed system is useful for clinical monitoring and for education in circulatory physiology.

**KEY WORDS.** blood pressure, central venous pressure. Equipment, computer, monitor.

## INTRODUCTION

Central venous pressure (CVP) measurement is employed to assess right ventricular filling pressure and to determine the hydrostatic pressure in the venous system [1]. To assess ventricular filling pressure ($CVP_{preload}$), it must be determined at the end-diastolic point during expiration, necessitating a graphic view of the pressure waveform along with EKG [2]. The mean CVP ($CVP_{mean}$) is often substituted for them. $CVP_{mean}$, however, is a measure of the hydrostatic pressure, which plays an important role in the development of systemic edema or hepatic congestion. The difference between the two variables in patients under mechanical ventilation has not been exactly quantified.

In addition to the above mentioned variables, CVP waveforms provide important diagnostic information such as tricuspid function, myocardial condition, and arrhythmias [1, 2]. However, it is often difficult to identify the wave components of CVP on a clinical monitor, because the waveforms are very easily distorted by artifacts occurring through the water-filled, tubing transducer system and by respiration-induced cyclic changes. We have developed a system using EKG-R synchronization and signal averaging by which CVP waveforms can be closely observed and accurate measurements of their components (*a*, *c*, *v* peaks and *x*, *y* troughs) can be made.

*This study was performed in the Department of Anesthesiology & ICM, Kawasaki Medical School.

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The aim of this study was to compare the $CVP_{mean}$ and the $CVP_{preload}$ in patients under mechanical ventilation and to determine the relative value of CVP waveform components by using our newly developed system.

## PATIENTS AND METHODS

With institutional approval for this observational study, we performed sampling of hemodynamic data and processing through catheters that were already in place for clinical care using a hemodynamic monitor (INFINITY SC 9000XL, Siemens AG, Munich, Germany) in operating rooms or ICU between October 2004 and June 2005. Twenty-five patients who had regular sinus rhythm on EKG were studied. Patients with known moderate to severe tricuspid valvular diseases in the preoperative examination were excluded.

Arterial pressure was measured via a radial artery catheter. CVP was measured through a central venous catheter (AK-16702-J, Arrow, Japan) positioned in the superior vena cava via the right internal jugular vein or the right subclavian vein. The proper position of the catheter was verified on a chest radiogram in all patients. Zero reference was obtained at the mid-axial level. Measurements were performed, while the lungs were mechanically ventilated.

### *R-synchronization and the averaging CVP measuring system*

EKG signals and pressure signals were sampled for ten seconds at 1000 Hz from the hemodynamic monitor and transferred to a PC computer equipped with A-D interface (PCI-3156, Interface Co. Hiroshima, Japan) and custom-made software (R-Synch, Version 1). The code was written by one of the authors (Y.F.) in Visual $C^{++}$ 6.0 (Microsoft, USA). The program to view data can be downloaded from the site via the Internet from our URL (http://www.kawasaki-m.ac.jp/anesicu/english/indexenglish.html). The system provides an auto-gain-display of the mean EKG, arterial pressure, and CVP traces for 10 seconds or the R-synchronized average traces of the three signals as well as digital readouts of their values (Figure 1A and B).

### *Definition of CVP waveform components and measurements [2]*

The *a*, *c* and *v* peaks were defined as peaks of CVP after the EKG P wave but before the EKG QRS wave, as an interruption of CVP decline immediately after the EKG QRS wave and as the peak of CVP just after the EKG *T* wave, respectively. The minimum CVP between the *a* and *v* waves was labeled the *x* trough and that between the *c* peak and the following *a* peak the *y* trough. Their values of CVP waveform components were determined on the PC screen of the system using mouse clicks. The geometric mean of the CVP displayed digitally on the PC screen was regarded as the $CVP_{mean}$. $CVP_{preload}$ was determined at the end-diastolic point during expiration on a graphic view of the auto-gained pressure waveform.

### *Statistics and data analysis*

All data were presented as means ± S.D. The level of statistical significance was $P < 0.05$. Paired t tests were used to determine whether one peak differed from other peaks or *x* trough from *y* trough. Bonferroni analysis was applied to correct *P* values for comparisons of three peaks. The agreement of the $CVP_{mean}$ and the $CVP_{preload}$ was analyzed by the Bland-Altman method [3].

## RESULTS

This system permitted identification of all the CVP components in all patients. Figure 1A and B illustrate representative EKG and CVP tracings and their processed signals in a 55-year-old man through R-synchronization and averaging for 10 seconds. All components of the CVP waveform are clearly visualized on the R-synchronized auto-gain scale.

The hemodynamic data, including arterial pressure, heart rate and the mean CVP, are summarized in Table 1. The CVP amplitude of the averaged signals was $3.4 \pm 1.2$ mmHg. The CVP changed from a maximal value of $10.7 \pm 3.8$ mmHg to the minimum value of $4.5 \pm 3.8$ mmHg through respiratory cycles for ten seconds. Table 2 shows the mean values of each CVP waveform component. Each value is presented as a difference from the $CVP_{mean}$. These results revealed that the *a* peak was $1.8 \pm 0.7$ mmHg above the $CVP_{mean}$ and that it was the highest of the three peaks (*c* and *v* peaks, $0.6 \pm 0.6$ and $0.5 \pm 0.7$ mmHg, respectively), and that the *x* and *y* troughs were $-1.6 \pm 0.7$ and $-0.9 \pm 0.5$ mmHg, respectively. The *x* trough was greater than the *y* trough.

The differences between the $CVP_{mean}$ and the $CVP_{preload}$ are plotted against their means in Figure 2. The former was $0.58 \pm 0.81$ mmHg less than the latter.

*Table 1. Hemodynamic data*

| HR (bpm) | SysArt (mmHg) | DiaArt (mmHg) | $CVP_{mean}$ (mmHg) | $CVP_{preload}$ (mmHg) |
|---|---|---|---|---|
| 60.5 ± 9.6 | 112.9 ± 18.0 | 57.7 ± 10.1 | 7.3 ± 3.7 | 7.9 ± 4.1 |

Data of 75 measurements from 25 patients were expressed as the means ± S.D. Abbreviations: HR: heart rate, SysArt: systolic arterial pressure, DiaArt: diastolic arterial pressure, $CVP_{mean}$: the geometric mean of the CVP for 10 seconds, $CVP_{preload}$: the CVP at end-diastole during expiration.

*Table 2. The mean values of each CVP pulse wave component*

| | *a* | *c* | *v* | *x* | *y* |
|---|---|---|---|---|---|
| mmHg | 1.8 ± 0.7[a] | 0.6 ± 0.6 | 0.5 ± 0.7 | −1.6 ± 0.7[b] | −0.9 ± 0.5 |

Values are expressed as differences from the geometric mean of CVP. Data of 75 measurements from 25 patients were expressed as the means ± S.D.
[a]Significantly greater than *c* or *v* ($p < 0.05$).
[b]Significantly less than *y* ($p < 0.05$).

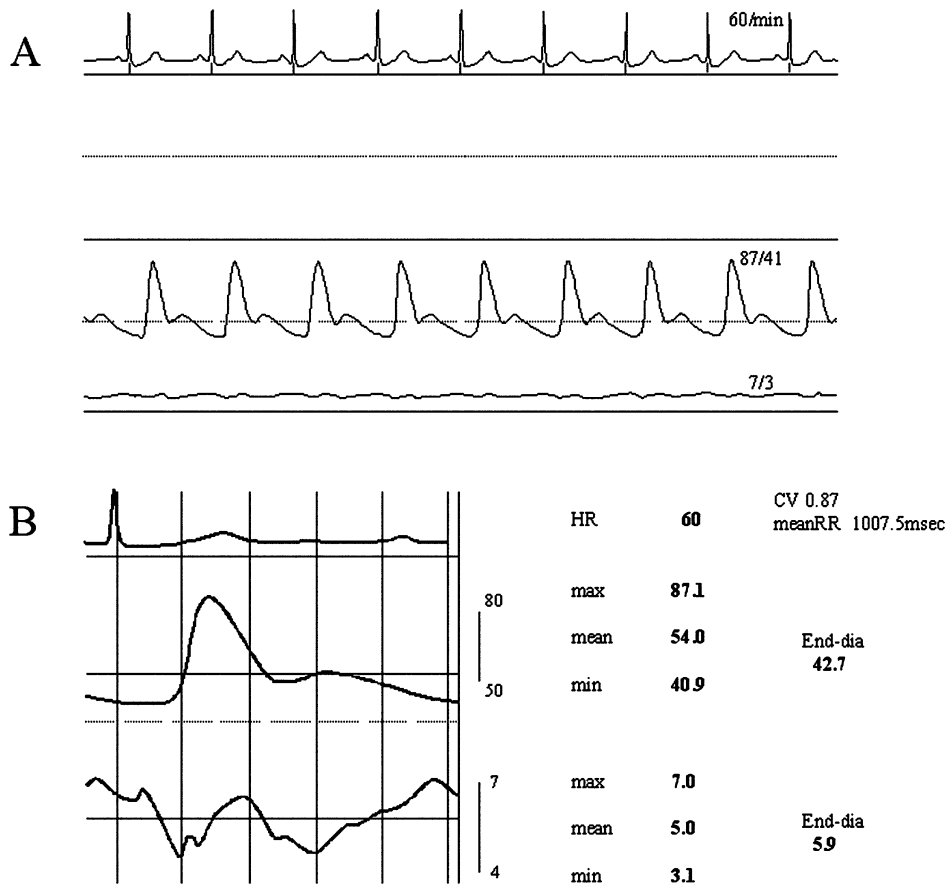


*Fig. 1. R-synchronization and averaging of CVP. EKG, arterial pressure and CVP signals are sampled from a hemodynamic monitor and transferred to a PC computer (A). The CVP signals are processed by R-synchronization and the averaged curves and digital readout appear on the display (B).*

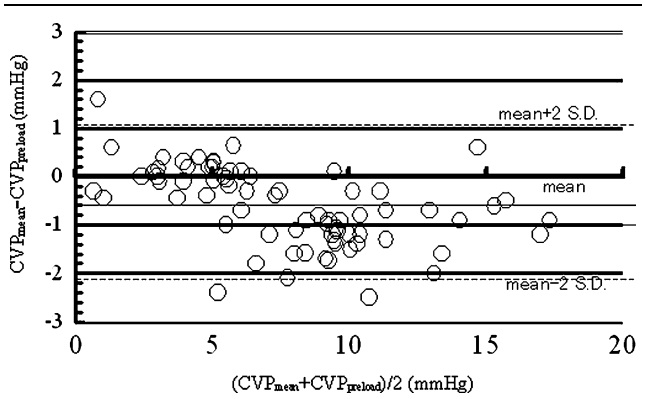


*Fig. 2. Bland-Altman plots showing the difference between the geometric mean of the CVP ($CVP_{mean}$) and the CVP at end-diastole during expiration ($CVP_{preload}$). The mean difference between the $CVP_{mean}$ and the $CVP_{preload}$ was $-0.6 \pm 0.8$ mmHg.*

## DISCUSSION

R-synchronization and averaging of CVP signals with our system clearly visualized CVP waveforms and permitted measurements of the waveform components in all patients. This study showed that in paralyzed patients with sinus rhythm under mechanical ventilation. $CVP_{preload}$ is slightly higher than $CVP_{mean}$, but the difference is clinically insignificant. Analysis of the averaged CVP waveforms revealed the *a* peak to be higher than *c* or *v* peaks, while there was no difference between the *c* and *v* peaks. The *x* trough was greater than the *y* trough.

CVP monitoring is routine during anesthesia for major surgery and in the critical care setting. A recent study demonstrated that physical examination of the jugular venous pulse has prognostic importance for heart failure [4]. However, information derived from invasive monitoring of CVP is underused, although it is continuously displayed on a clinical monitor [2]. To take the most advantage of it, identification of CVP waveform components and their analysis is a prerequisite. Our algorithm to average pressure signals with R-synchronization and to display signals with adequate amplitude by an auto-gain display permits preservation of the cardiac cycle-induced changes in CVP and reduction of respiratory influence. The system is useful for clinical monitoring and for education in circulatory physiology. Although we analyzed only CVP in this study, this system is applicable to the analysis of central vascular pressures such as arterial, pulmonary artery, and pulmonary artery wedge pressures.

$CVP_{preload}$ is determined at the end of expiration to minimize the effects of intrathoracic pressure. Because of the technical difficulty in determining it, $CVP_{mean}$, i.e., the readout of the clinical monitor, has been used for clinical purposes [1]. This is the main reason for the discrepancy between digital readouts and the graphic method. Comparative studies have indicated that digital readouts are unreliable and the graphic method is necessary for accurate measurements [5, 6]. This study suggested that the use of $CVP_{mean}$ as the index of right ventricular preload ($CVP_{preload}$) is acceptable, at least, in patients with normal CVP waveforms under mechanical ventilation. The discrepancy between our results and the previous studies may be explained by the difference in respiratory mode. In previous studies both patients with spontaneous breathing and ones under mechanical ventilation with and without spontaneous breathing are included [5, 6]. Respiratory fluctuation in the CVP can be removed by low frequency filtering [7], minimizing the difference between $CVP_{mean}$ and $CVP_{preload}$. This method has, however, not been well accepted world wide. We suggest that $CVP_{preload}$ should be measured manually by the graphic method when significant changes in intrathoracic pressure are suspected.

Although the importance of the jugular venous pulse for the cardiovascular physical examination was noted as early as 1902 [8], the normal waveform along with its components have not been quantitatively determined. The characteristic and amplitude of CVP waveforms are affected by arrhythmias and tricuspid valve pathology [1, 2]. While CVP changed between 10.7 and 3.4 mmHg during respiratory cycle, the amplitude of averaged CVP waveform was only $3.4 \pm 1.2$ mmHg. This difference is the main reason for the difficulty in obtaining distinct CVP waveforms and determining the level of each component. This study using our newly developed system delineated it. It revealed that the *a* peak is the highest of the three peaks. Accordingly, it can be inferred that if the *c* peak is greater than the *a* peak, the aright atrium is non-functioning or there is tricuspid regurgitation. A *v* peak greater than the *a* peak in patients with sinus rhythm may be compatible with tricuspid regurgitation.

### *Limitations of the system*

First, although CVP waveform analysis is diagnostically useful, it's sensitivity and specificity is not validated by the gold standard methods, such as echocardiography. For example, the *c-v* wave, i.e., an increase in CVP beginning early in systole and lasting to the end of the EKG T wave, is diagnostic for tricuspid incompetence. A typical *c-v* wave appears when there is a large regurgitant volume with a non-compliant atrium. Typical *c-v* wave may disappear in chronic tricuspid insufficiency due to right atrial enlargement, although regurgitation remains severe. Second, since we performed this study in paralyzed patients with normal CVP waveforms under mechanical ventilation, the results

cannot be simply extrapolated to patients with abnormal CVP waveforms. However, the data can be used as the basis for normal values to diagnose waveform distortion. Third, ability of our system to obtain clear waveform depends solely on accurate R wave detection. While irregular R–R intervals such as atrial fibrillation do not interfere with processing, baseline EKG shift during measurement for ten seconds may make it impossible to identify *R* wave, resulting in failure of processing of CVP signals.

In conclusion, R-synchronization and averaging of CVP signals permit clear visualization of the CVP waveform and accurate measurements of its components in patients under mechanical ventilation. It is useful for clinical monitoring and education in circulatory physiology. The results in this study revealed that the *a* peak was the highest of three peaks, and that the *x* trough was less than the *y* trough ($-1.6 \pm 0.7$ mmHg and $-0.9 \pm 0.5$ mmHg, respectively), resulting in a pulse pressure of 3.4 mmHg. It revealed that the $CVP_{mean}$ can be used as $CVP_{preload}$, at least, in paralyzed patients with regular sinus rhythm under mechanical ventilation.

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