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Accurate, Automated, Continuously Displayed Pulmonary Artery Pressure Measurement

Mark M. Mitchell, M.S., M.D., Edward A. Meathe, M.S.,† Brian R. Jones, M.D.,‡ Terese E. Donch, M.S.,§ William G. Ricks, B.A.,¶ Jonathan L. Benumof, M.D.,** Lawrence J. Saidman, M.D.***

A computerized signal processing technique that removes low-frequency respiratory variation from pulmonary artery pressure and other central vascular pressure measurements, and produces a waveform devoid of respiratory artifact, has been developed. This technique has been integrated into a portable bedside monitor. The authors tested the technique in critically ill patients, and found that, compared to physician readings of conventional strip charts, it proved to be a very convenient and accurate method of determining pulmonary artery pressures continuously, regardless of ventilation. (Key words: Algorithms. Equipment, computers: monitoring. Measurement techniques: computerized monitoring; pulmonary artery pressure. Monitoring: pulmonary artery pressure.)

PULMONARY ARTERY PRESSURE (PAP) measurements are often required to adequately assess the cardiopulmonary status of critically ill patients. However, PAP is affected by pleural pressure, and, consequently, by ventilation. Whenever pleural pressure changes are large and rapid, as with decreased pulmonary compliance or respiratory failure, reading accurate pulmonary artery systolic and diastolic pressures may be difficult.

Although there have been several previous attempts to facilitate the reading of PAP during large and rapid changes in pleural pressure, they each have major drawbacks, such as requiring a physician or technician with special training,[1–3] requiring non-real time processing with unacceptably long delays,[1–3]†† requiring additional sensors and equipment such as a detector for respiratory gas flow, temperature, or pressure,[1–3]†† yielding only intermittent digital information without a continuous waveform,[6–7] or failure to compensate completely for respiratory variations.[4–6]

We have developed an algorithm that has none of the limitations of the above methods, and has the added benefit of producing continuous on-line PAP waveforms that have ventilation-induced variations removed from the recording. The assumption is that respiratory events occur at a lower frequency than the heart rate. An adaptive automatic dynamic filtering process eliminates variations in PAP that occur less frequently than the heart rate. Consequently, a continuous PAP trace is produced that no longer has respiratory-induced fluctuations.

We have compared the pulmonary artery systolic and diastolic pressures determined by this automatic filtering process against readings by experienced clinicians. The study patients were critically ill, with a wide variety of respiratory patterns, heart rates and rhythms, and pulmonary vascular pressures. The method was found to be extremely accurate and precise. We also discuss and present examples of the effect of this automatic filtering process on pulmonary capillary wedge pressure (PCWP) and central venous pressure (CVP) waveforms.

Methods

The study protocol was evaluated and approved by the Human Subjects Investigation Committee. Only patients with indwelling systemic arterial and pulmonary artery catheters and endotracheal tubes, as part of their medical management, were studied. Arterial, airway, and PAP were measured with standard transducers which were zeroed by opening to atmospheric pressure at the level of the left atrium as determined by the mid-axillary line. All measurements were made with the patients in the supine position. Airway pressures were monitored *via* a T-connector interposed between the endotracheal tube and the ventilator hose. Vascular and airway transducers were calibrated with a standard mercury manometer.

The electrocardiogram, arterial pressure waveform, unprocessed PAP, and airway pressure were recorded on a multi-channel FM tape recorder. Examples of CVP and PCWP were also recorded for some patients. The data were processed by the algorithm either at the bed-

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* Associate Clinical Professor.
† Principal Development Engineer.
‡ Resident in Anesthesiology.
§ Hospital Laboratory Technician.
¶ Developmental Technician.
** Professor of Anesthesiology.

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Address reprint requests to Dr. Mitchell.

side as the data was collected, or the recorded raw data was processed by the algorithm at some later time.

The algorithm determines the heart rate on a beat-to-beat basis from the arterial pressure pulse, and the heart rate is used to dynamically select appropriate low-pass and high-pass digital filters. The raw PAP data is passed through the low-pass filter to remove high frequency (phasic cardiac) components. This results in a signal composed of only the low frequency airway components and the mean PAP. The mean PAP is determined from this signal. The unfiltered PAP signal is also passed through a high-pass filter to remove low frequency (respiratory) components, or, equivalently, the output of the low-pass filter is subtracted from the unfilter raw signal. This results in a signal containing only the high-frequency phasic cardiac components. The mean PAP is added to the high-pass filtered data. The reconstituted filtered PAP phasic waveform without the effects of respiration is plotted on standard strip chart paper along with the raw, unprocessed, PAP signal and the airway pressure.

To test the accuracy of the algorithm, two clinicians (authors MMM and JLB) each independently read the pulmonary artery systolic (PAS), and pulmonary diastolic (PAD) pressures from the unprocessed PAP trace at the end expiratory moment, just before the next breath starts. This moment was determined from the simultaneously recorded airway pressure trace.[1] The values chosen by the clinicians from the unprocessed PAP trace were compared to the values for the same beats from the processed PAP trace. The clinicians had no previous knowledge of the pressure determinations made by each other or the algorithm. Each physician's determinations, as well as their pooled determinations, were correlated with the computer's determinations. Processed and unprocessed examples of the CVP and PCWP were also compared for some of the patients.

Results

Figure 1 shows an example of an unprocessed PAP waveform (upper panel), and a processed (algorithm filtered) PAP waveform (lower panel). In this typical example, the unprocessed pulmonary artery systolic and diastolic pressure vary by more than 20 mmHg over the 20 heart beats displayed primarily due to changes in intrathoracic pressure with respiration. The processed pulmonary artery systolic and diastolic pressures have a variation of about 5 mmHg due to changes in pulse pressure.

In 18 patients, the processed PAP was found to correlate very well with the pressure determined by the physicians at the end-expiratory moment from standard

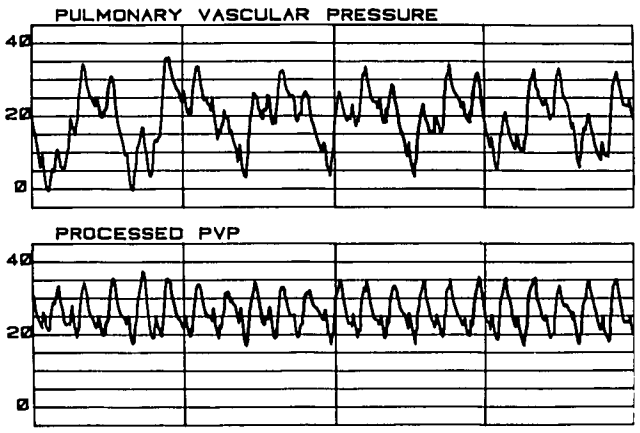

FIG. 1. Raw and processed pulmonary artery pressure. PVP = pulmonary vascular pressure.

strip chart tracings. These data are listed in table 1. For all patients, the difference between the values read by the physicians from the standard strip charts and the values produced by our automated algorithm was, at most, 3 mmHg. The linear regression line for PAS pressure between physician #1 (X) and our computerized algorithm (Y) is $Y = -1.81 + 1.07*X$ with an R value of 0.99, and, for PAD pressure, is $Y = 0.73 + 0.96*X$ with an R value of 0.98. The linear regression line for PAS pressure between physician #2 (X) and the computerized algorithm (Y) is $Y = -0.04 + 0.97*X$ with an R value of 0.99, and for PAD pressure is $Y = 0.45 + 0.97*X$ with an R value of 0.98. If the determinations of the two physicians are pooled, the linear regression for PAS pressure is $Y = -0.74 + 1.02*X$ with an R value of 0.99, and, for PAD pressure, is $Y = 0.58 + 0.97*X$ with an R value of 0.98. The 95% confidence limits (by t test) on the slope of the regression line are 0.93–1.11 for PAS pressure and 0.88–1.06 for PAD pressure. The 95% confidence limits for the zero crossing of the regression line are −1.68–1.60 mmHg for PAS pressure and −0.69–1.85 mmHg for PAD pressure.

Heart rates in the study patients varied from 60–150 beats/min, with one patient having a pacemaker-controlled heart rate of 70 beats/min. Data were collected from the patients with frequent premature ventricular contractions, premature atrial contractions, and other irregular heart rates, including atrial fibrillation. Ventilation rates were from 6–54 breaths/min. Ventilation was either completely spontaneous, aided by intermittent mandatory positive pressure ventilation, or entirely controlled. Ventilators were either volume or pressure cycled. Intervals between negative and positive pressure

TABLE 1. Comparison of Physician and Computer-picked Pulmonary Artery Systolic and Diastolic Pressures

Patient #	PAS Physician	PAS Computer	PAD Physician	PAD Computer
Physician #1				
1	24	25	14	13
2	30	27	12	13
3	14	14	9	9
4	30	30	21	20
5	30	29	28	26
6	15	14	12	11
7	39	27	25	24
8	33	33	20	20
9	33	34	17	17
10	17	16	11	12
11	19	19	9	9
12	39	41	16	18
13	38	38	23	24
14	28	28	11	11
15	32	33	16	15
16	30	30	12	14
17	44	46	26	28
18	30	32	15	15
Physician #2				
1	26	26	16	14
2	16	16	5	5
3	17	15	10	9
4	22	25	13	14
5	35	33	22	20
6	20	19	11	10
7	30	28	25	23
8	30	32	20	22
9	43	43	28	30
10	16	16	10	13
11	22	21	14	13
12	39	40	20	19
13	38	38	21	22
14	27	24	11	12
15	31	30	14	14
16	32	32	14	14
17	50	48	29	28
18	27	27	17	17

PAS = pulmonary artery systolic pressure; PAD = pulmonary artery diastolic pressure.

breaths, in the large majority of patients with a mixed ventilatory pattern, were irregular (fig. 1). PAS/PAD pressures ranged from 14/9 mmHg to 50/29 mmHg.

When examples of raw and processed CVP and PCWP were compared, it was evident that the cardiac-related features (*e.g.*, mean value, A, C, and V waves, etc.) of the waveforms are preserved, subject to the restrictions described in the discussion to follow. The effect of performing an electrical mean on these processed and unprocessed signals is complicated by the relationships of the heart rate, respiratory rate, and the time constant of the averaging process, and also by the details of the respiratory pattern, as described in the discussion.

Discussion

Respiratory variations can cause PAP signals to be difficult to read accurately.[1] Removal of respiratory variations can simplify interpretation of these pressures and help prevent mismanagement of patients based on erroneously interpreted PAP measurements. The algorithms previously developed to solve this problem suffer from major drawbacks.

Comparison with Other Algorithms

The algorithms used by presently available monitoring equipment to produce digital readouts may be misleading, especially for reporting PAP. Most digital systems were designed primarily for use with systemic blood pressure signals and display data selected randomly every 2–4 s from the pressure signal. They use electronic circuits that identify the highest, lowest, and mean values over predetermined interval (typically 1–4 s), regardless of the relationship of the predetermined interval to the respiratory cycle. In addition, since there is a delay in response, it is impossible for a clinician observer to correlate the digital readout with any particular ventilatory event or pattern. Therefore, if a digital system is being used, the only way a clinician can choose the end-expiratory moment to determine the appropriate PAP is to use simultaneous recordings of airway pressure (or some other indicator of the phase of the respiratory cycle) and PAP.

An automated method for reading PAP values during the end-expiratory moment, by sensing airway temperature changes due to inhalation and exhalation, has previously been described.[3] However, this method of discerning the end-expiratory period requires a special airway thermistor catheter placed in the endotracheal tube, or in nasal prongs for unintubated patients, and equipment to sense and process the respiratory cycle airway temperature changes. It yields accurate values, but only intermittently, and delayed by 16 s.

An algorithm using a Fast Fourier Transform has been described and shown to provide accurate PAP without respiratory variations.†† This method samples and processes discrete blocks of data (*i.e.*, 16-s epochs), and then reconstructs the final signal. Therefore, although accurate, this method, as for the one above, also has the drawback of providing intermittent digital data with an undesirably long delay.

Cengiz *et al.* compared manual to automated PAP readings using a computerized algorithm that measured PAP by digitizing the PAP signal in blocks, and then determining the PAS and PAD.[4] The algorithm did not incorporate any respiratory compensation, and they

found significant inaccuracies in the data obtained with the patients breathing spontaneously. However, they found that data from mechanically ventilated patients were fairly consistent with the pressures read manually from strip-chart recordings.

Ellis described an algorithm using variable weighted averaging, and illustrated its function in 15 patients.[5] This algorithm determines the difference in PAP in each successive beat, and the beats that are greatly different from previous beats are less heavily weighted. In theory, this algorithm should weight the "stable" beats occurring in the end-expiratory period the heaviest, and the final output should be closer to the real pressure occurring during the end-expiratory period than if simple averaging has been used. Yet, the weighted averaging method begins to fail when the respiratory rate approaches half the heart rate (*i.e.*, when every other beat is affected by respirations and there are not two successive beats to weight heavily). Unfortunately, the ability of the algorithm to handle rapid respiratory rates and other "difficult signals" was not reported. In addition, this algorithm does not produce a processed waveform. It displays only the unprocessed waveform on the monitor in the usual fashion, and presents the filtered PAS and PAD only in digital form. Consequently, the clinician has no easy way to check the accuracy and consistency of the digital values.

Another monitor and algorithm that calculates the beat-to-beat systolic, diastolic, and mean pressure was briefly described by Kari *et al.*[6] In their method, the mean pressure of each beat is calculated by dividing the area under each pressure wave by its duration. They state that the algorithm is able to detect end-expiratory period if the patient's respiratory pattern is simple spontaneous negative pressure ventilation or mechanical positive pressure ventilation, and the algorithm is told which pattern it is. The algorithm then provides a weighted average by only considering those values from the beats that occurred during the presumed end-expiratory periods. They reported with scant field testing that the algorithm slightly overestimates all pressures during spontaneous breathing and underestimates them during mechanical ventilation (measurement errors were reported under 4 mmHg). However, no filtered waveform is produced, and only digital values are available. Moreover, the logic of the algorithm is such that it will likely fail at higher respiratory rates due to errors in identifying the end-expiratory periods, even if the heart rate is not near the respiratory rate. The algorithm fails completely for complex intermittent manditory ventilation respiratory patterns which combine negative pressure spontaneous ventilation and positive pressure mechanical ventilation.

The algorithm we use has none of the deficiencies of the other methods. By utilizing digital filtering techniques, our algorithm is able to remove respiratory variations from a PAP waveform in real-time with minimal delays and provide a filtered waveform from which digital systolic and diastolic values can be determined by any accepted method. The present study demonstrates that the algorithm is accurate when used on patients that have extremes in their cardiopulmonary variables (respiratory rates, heart rates, vascular pressures). With our algorithm, occasional random events, such as bucking, coughing, hiccups, or catheter whip, are eliminated because they occur at a rate less than the heart rate.

After application of our algorithm to PAP raw data, there is often variability in the PAP pulse pressure evident, as in figure 1. This is due to changes in right ventricular stroke volume, and perhaps pulmonary vascular caliber, with respiration.[8] This variability in PAP pulse pressure related to the true hemodynamic effects of ventilation is frequently obscured by the effects of ventilation on pleural pressure in the raw PAP waveform.

Our system displays a filtered waveform, but the raw unprocessed waveform is also available for display. Systolic, diastolic, and mean pressures can be easily manually read from this resulting stable PAP signal (fig. 1). The algorithm also has the potential to easily incorporate an accurate digital readout for these values. This digital readout system can use standard analog or digital circuits designed for systemic blood pressure monitoring. Provision for display of both the processed and unprocessed waveforms allows the user to easily verify the accuracy of the digital values derived from the processed waveform. Our algorithm is presently integrated into a portable bedside monitor.

Our algorithm determines the mean PAP that would occur during the end-expiratory period by passing the raw PAP signal through a low-pass filter (removes high frequency cardiac cycle components). The algorithm then estimates the mean pressure during the end-expiratory period by determining the mode (most common), not the mean pressure. Then this most common amplitude of the low-pass filtered signal is added back to the high-pass filtered signal (cardiac phasic components) as the mean pressure. The reasoning is similar to that of Ellis' algorithm[5] and is validated by the results of this study since the mean pressure is a component of the systolic and diastolic pressures studied.

Limitations of the Algorithm

Our algorithm might be slightly inaccurate in two physiologically extreme conditions. First, this algorithm

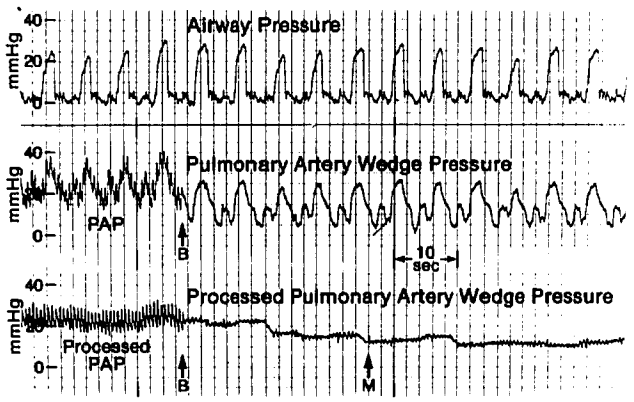

FIG. 2. Raw and processed pulmonary artery pressure (PAP), including transition to pulmonary capillary wedge pressure. The filter algorithm takes 30 s from catheter tip balloon inflation at ↑B to accurately reflect the change in mean pressure at ↑M. Airway pressure is included to allow confirmation of end-expiratory period pressures. Airway pressure and raw pulmonary artery and wedge pressure have been digitally delayed to be syncronized with the processed pulmonary artery and wedge pressure. This patient's heart rate was 106 beats/min, his spontaneous respiratory rate was about 30 breaths/min, and his mechanical ventilation rate was 10 breaths/min.

assumes that the mean pressure will be observable between breaths during the end-expiratory period when the pressure is as close as it will come to the actual transmural pressure.[1] Clearly, sustained inspiratory pressures (*e.g.*, an inspiratory "hold" in a mechanical ventilatory cycle) that are maintained for a longer period than the end-expiratory period could invalidate

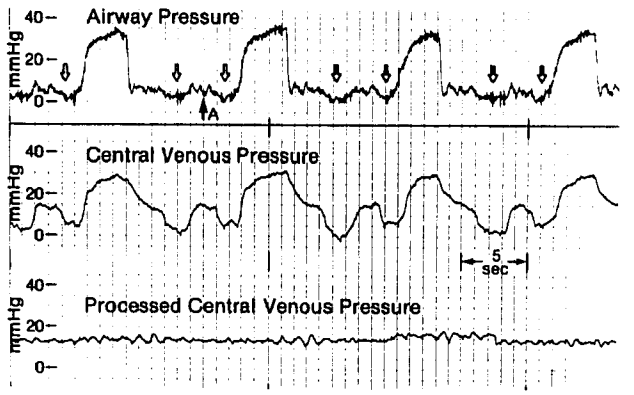

FIG. 3. Raw and processed central venous pressure (CVP). Airway pressure is included to allow confirmation of end-expiratory period pressures. Airway pressure and raw central venous pressure have been digitally delayed to be syncronized with the processed central venous pressure. Spontaneous breaths are marked ↓ on the airway pressure trace. Pneumocardiogram is discernible in the airway pressure trace at ↑A during the end-expiratory period. This patient's heart rate was 106 beats/min, his spontaneous respiratory rate was about 30 breaths/min, and his mechanical ventilation rate was 10 breaths/min.

this assumption. Nor is the contribution of positive end-expiratory pressure (PEEP) to the mean PAP automatically removed by our algorithm. The presence of PEEP needs to be considered by the clinician during hemodynamic analysis, just as it must be with all other techniques for measuring pulmonary vascular pressures.

Second, in studying individual patients with varying heart rates, our algorithm may display a slight, transient error in PAP for less than 1 s. The reason for this is that the heart rate determines what frequencies are filtered. Consequently, if the heart rate suddenly decreases, there can be a transient period where the selected filter removes part of the new slower cardiac waveform complex. This over-filtering is very transient, as the algorithm rapidly accomodates by determining a new heart rate and selecting a new appropriate filter for the slower heart rate. This phenomenon became apparent in analyzing PAP in the presence of atrial fibrillation, where the PAP was observed to be over-filtered. Slight modifications in the algorithm, by increasing the interval below the heart rate at which filtering begins and increasing the frequency at which the filter cut-off limit is updated, minimizes this particular problem (see Appendix).

With the present program, a 1-s delay is introduced due to the digital filtering. When viewed on a screen in real-time with the EKG only, observers did not find this objectionable. When comparison with other concurrent vascular pressures was desired, an equal delay of the other pressures was implemented. This was found to be quite acceptable.

In addition to this 1-s delay in the phasic components of the processed waveform, there is an additional delay in computing the mean value (by selecting the mode of the low-pass filtered signal component). This can become transiently important if there is a sudden change in the true mean of the raw waveform, such as occurs when the catheter tip balloon is inflated and a PAP becomes a PCWP. This is illustrated in figure 2. In this example, it takes 30 s for the algorithm to accurately represent the mean PCWP. This example was processed using a early version of our algorithm. Improvements have been made to reduce this period during which the mean pressure may be inaccurate to 7.5 s, and addition of a simple smoothing filter can eliminate the obvious steps in mean pressure estimation seen in figure 2. Since these improvements do not eliminate this possible source of transient error, we chose to use an example which clearly demonstrates it. Knowing about this limitation makes it easy to avoid reporting pressure values during this short period.

In figure 2, it is interesting to note that, because of

this delay in determination of the mean, immediately after inflation of the catheter tip balloon at 1B, the PAP mean pressure is easily seen.

Application to Other Central Vascular Pressures

When a PCWP or CVP waveform is processed with our algorithm, the cardiac-related features are preserved and the effects of respiration are reduced or eliminated (fig. 2, 3).

The most often reported feature of these signals is the mean, which is generally determined by passing the raw signal through a simple smoothing filter characterized by a single time constant. This is reasonable, since, in the absence of tricuspid or mitral valve incompetence or gross prolapse, arrhythmias such as atrioventricular dissociation, and other conditions which result in large "V" or "A" waves, the mean relates well to effective ventricular pre-load and cardiac performance, and the details of the waveforms are of little significance.

PEEP and/or positive pressure ventilation or spontaneous negative pressure ventilation also effects the mean value of the PCWP or CVP. Poor (low) pulmonary or thoracic compliance will exaggerate the effects of spontaneous negative pressure ventilation on the mean PCWP or CVP, and the effects of PEEP and/or positive pressure ventilation will be most pronounced when there is normal (high) compliance.

The time constant of the smoothing filter used to create a mean needs to be as long as or longer than several times the period of the heart rate. If the time constant is short relative to the respiratory rate, it may be possible to read an end-expiratory mean PCWP or CVP. If the heart rate and the respiratory rate are close together, or the time constant is long compared to the respiratory rate, this will not be possible. Our algorithm can identify these situations and alert the user.

If a PCWP or CVP signal is processed with our algorithm, the mean pressure is computed as the mode (most common value) of the output of the low pass filter. This value will be subject to the same effects discussed relative to PAS and PAD pressure determinations, such as PEEP and inspiratory hold, and the relationship of the heart rate to the respiratory rate. When the heart rate is not too close to the respiratory rate, and a stable end-expiratory period exists, then, by direct analogy with the analysis and clinical validation of this algorithm as applied to PAS and PAD pressures, the mean reported by the algorithm will be the end-expiratory mean PCWP or CVP, respectively. This may, of course, be a different value than the mean of the raw signal, determined with a time constant which is long compared to the respiratory rate, as the effects of ventilation are removed from the algorithm processed signal, but contribute to the mean value of the raw signal.

In summary, our method of signal processing removes low-frequency respiratory variations producing a continuously processed pulmonary arterial pressure signal devoid of respiratory artifact. The method improves the precision and consistency of measuring PAP, avoids the errors inherent in having personnel with various levels of experience and understanding reading the PAP from the presently available systems, and does not require airway pressure, temperature, or gas flow monitoring to detect the end-expiratory period. The method provides a continuous PAP waveform with an acceptable 1-s delay. It is also applicable to PCWP and CVP measurements. This algorithm is presently integrated into a portable bedside monitor, and has the potential to be installed in-line with present monitors, or incorporated into future monitoring systems.

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Appendix

The algorithm utilizes two channels of analog information. These are pulmonary artery pressure (PAP) and systemic arterial pressure. These signals are sampled at 50 Hz and converted to digital form by an analog-to-digital (A/D) converter. The digital data are processed in real-time, but can also be stored. Because of frequent artifacts in EKG signals, heart rate is determined from the arterial pressure signal. Sometimes the heart rate can also be determined from the PAP signal itself.

Any reliable monitoring modality that supplies an accurate determination of heart rate at least every 10 s can be used. Using the determined heart rate, a 60-pole, linear phase, finite impulse response digital low-pass filter (Butterworth Type) is selected that eliminates high-frequency components at and above the heart rate (suppression is 40 db or more outside the pass band). This results in a signal composed of the mean PAP, airway components, and other low-frequency variations. This resulting signal is divided into a histogram format (50 Hz samples cells by increments of ½ mmHg), and the most common (*i.e.*, the mode) amplitude is taken as the mean pressure. The raw PAP is also simultaneously passed through a high-pass filter with characteristics similar to the low-pass filter, producing a signal containing the phasic cardiac-waveform, without the low frequency respiratory components. The simultaneously derived mean PAP is then added back to this high frequency waveform signal, producing the final processed PAP signal. The information is then converted back to analogue form by a digital-to-analog (D/A) converter. The filtered PAP, raw PAP, arterial pressure, and EKG or airway pressure are displayed on both a video monitor and paper-strip-chart recorder. Because of the processing time required, the final PAP signal is delayed 1 s. To correlate the processed PAP on the scope and strip chart recorder, with the arterial pressure, airway pressure, and/or EKG, the latter three signals are also intentionally delayed 1 s to coincide with the processed PAP.

The heart rate determination is made very 10 s, and the selection of the high pass and low pass filters is updated. Performance is improved by actually selecting the cut-off frequency between high-pass and low-pass filters slightly below the heart rate (by about 5 beats/min) to accommodate slight variations in heart rate between updates and assure minimal loss of cardiac related information.

With some small increase, the processing delay, the heart rate determination, and cut-off frequency selection can be done more often, with an improvement in performance apparent on PAP waveforms with large beat-to-beat variability, such as atrial fibrillation. This is not necessary for most PAP waveforms with their usual much smaller beat-to-beat variability.