COMPUTERIZED ARTIFACT DETECTION FOR VENTILATORY INDUCTANCE PLETHYSMOGRAPHIC APNEA MONITORS

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ABSTRACT. Ventilatory inductive plethysmography allows noninvasive monitoring of patient ventilation. Patient movements unrelated to breathing introduce severe errors in ventilator inductive plethysmographic measurements and restrict its usefulness. The purpose of this research was to develop and test a microprocessor-based real-time digital signal processor that uses an adaptive filter to detect patient movements unrelated to breathing. The adaptive filter processor was tested for retrospective identification of artifacts in 20 male volunteers who performed the following specific movements between epochs of quiet, supine breathing: raising arms and legs (slowly, quickly, once, and several times), sitting up, breathing deeply and rapidly, and rolling from a supine to a lateral decubitis position. Flow was simultaneously measured directly with a pneumotachograph attached to a mouthpiece. A multilinear regression was used to continuously calculate the calibration constants that relate the pneumotachographic and ventilatory inductive plethysmographic signals. Ventilatory inductive plethysmographic data were then processed, and results scored. There were a total of 166 movements. The calibration coefficients changed dramatically in 146 (88%) of the 166 movements. These movements would have significant errors on ventilatory inductive plethysmographic flow calculation. The changes lasted for the duration of the movements and returned to baseline within two to three breaths. The changes in the coefficients were five or more times larger than the variability around baseline during quiet, supine breathing. All of the total body movements and changes in breathing patterns were detected accurately. The filter detected 46 of 53 upper body movements, 34 of 36 lower body movements, 38 of 38 total body movements, and 19 of 19 breathing pattern changes where the calibration changed. The filter was able to detect 94% of the total 146 movements. These results could help improve the effectiveness of ventilatory inductive plethysmography as an apnea monitor for use in patients receiving epidural narcotics. More accurate respiratory assessments could also be made during sleep studies, pulmonary evaluations, or exercise evaluations.

KEY WORDS. Measurement techniques: plethysmography. Monitoring: ventilation. Ventilation: spontaneous measurement.

It is important in a wide variety of situations to accurately measure a patient's ventilation. Commonly used techniques requiring a mouthpiece or face mask induce variable changes in tidal volume, minute ventilation, and respiratory rate [1–6]. These techniques are impractical for long-term monitoring. Respiratory or ventilatory inductance plethysmography (VIP) measures ventilation noninvasively with inductance transducers about the rib cage and abdomen. Movements unrelated to breathing are a major source of error when using VIP. Movements and postural changes alter the volume-motion coefficients that relate the rib cage and abdominal signals to volume and flow [7,8].

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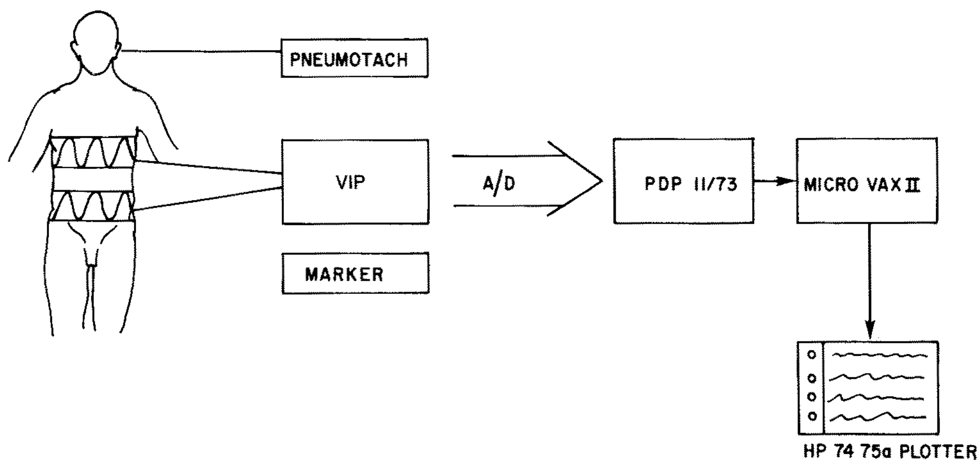

Fig 1. Ventilatory inductance plethysmography (VIP) *and data collection and analysis system.* A/D = *analog-to-digital;* HP = *Hewlett-Packard.*

The purpose of this study was to detect movements unrelated to breathing by implementing a real-time digital adaptive filter on a microprocessor system. The adaptive processor was tested in a group of volunteers to determine if the adaptive filter detected all incidents in which the calibration coefficients relating rib cage and abdomen to flow changed dramatically due to movements unrelated to breathing.

METHODS AND MATERIALS

The VIP system (Fig 1) consists of two inductance transducers placed about the chest and abdomen, an oscillator unit, and a demodulator/calibrator. The VIP transducers are zigzag coils of wire attached to an elastic bandage. One transducer is placed around the rib cage over the sternum just under the axilla. The second is placed around the abdomen below the rib cage and above the iliac crest. The coils are attached to a small oscillator module. The oscillators run at frequencies around a baseline of 300 kHz. The output is a frequency-modulated signal around a baseline of 20 kHz [8]. The frequencies of these oscillations are varied by changes in the inductance of the coils due to changes in their cross-sectional areas resulting from respiration. The oscillator module is connected to the demodulator/calibrator, which demodulates the signal to provide a direct-current output signal.

The relationship between tidal volume and the rib cage and abdominal signals was derived for a ventilation model of the chest and abdomen described by Sackner et al [9]. As the volumes of the rib cage and abdomen change during ventilation, the cross-sectional area of the inductance coils changes in proportion to the volume. In the model, the total volume change is proportional to the sum of the volume change of the rib cage (RC) and the abdomen (ABD). Therefore, the total (t) volume can be calculated from the following:

$$\mathrm{Volume}(t) = b \times \mathrm{RC}(t) + c \times \mathrm{ABD}(t), \qquad (1)$$

where b and c are calibration coefficients, or volume-motion coefficients. The coefficients represent the portion of the total volume that is contributed by the rib cage and by the abdominal excursion during breathing. Flow is then described by:

$$\mathrm{Flow}(t) = b \times d[\mathrm{RC}(t)]/dt + c \times d[\mathrm{ABD}(t)]/dt. \qquad (2)$$

In a typical monitoring session with the VIP system, these values (b and c) are calculated at the beginning of the session by using either a volume or flow technique; then the values are assumed to remain constant throughout the monitoring session.

With the volume technique, the subject rebreathes into a bag of known volume, breathes into a spirometer that measures inspired and expired volumes, or performs an isovolumetric maneuver shifting air between the abdomen and the rib cage with the airway blocked. The volume-motion coefficients are estimated by solving a set of simultaneous equations [10–14]. In the flow technique, the patient breathes through a pneumotachograph for approximately 2 minutes. Then, a multilinear [7,14] or orthogonal [15] regression is used to calculate the coefficients b and c. Calibration can be done by using one position (standing, sitting, or supine) [14–16] or a combination of positions [8,10–12].

Figure 1 shows the VIP signal processing system that was used. It consisted of a basic VIP system and a Fleisch pneumotachograph interfaced to a PDP-11/73 microcomputer (Digital Equipment Corp [DEC], Maynard, MA). The microcomputer included a 16-channel, 12-bit analog-to-digital converter, a 20-megabyte hard disk, two 5¼-inch (13.3-cm) floppy disks, 512 kilobytes of memory, and a DEC VT240 graphics terminal. A 5-V switch was used as an event marker by the observer to document movements. The flow, rib cage, abdomen, and event marker signal were sampled at 20 Hz and stored on hard disk in binary files for later processing with the adaptive filter. The data were transferred via an RS232 line to a MicroVax II for processing and plotting on a Hewlett-Packard 7475a plotter. It should be noted that the data can be processed on the PDP-11/73 microcomputer or a similar system. Our data were transferred to the MicroVax II for convenience of plotting.

We developed a method that uses an adaptive filter to detect patient movements from the VIP signals. The basic idea is that the relative volumes of the rib cage and the abdomen change during most patient movements. The adaptive filter is capable of tracking these relative changes and therefore will detect most patient movements. We chose a particularly efficient and popular adaptive filter known as the least-mean-square filter [17]. The first-order least-mean-square adaptive filter that we used estimates the rib cage volume as a scaled version of the abdominal volume, ABD(n), that is, the estimate of the rib cage volume $\widehat{RC}(n)$ is obtained as

$$\widehat{RC}(n) = h(n) \times ABD(n). \quad (3)$$

The scaling factor (adaptive filter coefficient), h(n), tracks the changes in the relative volume contributions. The least-mean-square filter minimizes the mean-squared estimation error by updating h(n) using a "gradient" approach as given by [17]:

$$h(n) = h(n - 1) + \mu \times ABD(n - 1) \times e(n - 1), \quad (4)$$

where e(n) is the error between the true and estimated values of the rib cage volume-motion coefficient, and is given by

$$e(n) = RC(n) - \widehat{RC}(n). \quad (5)$$

The speed at which the adaptive filter can track the changes is controlled by a small positive constant, μ. The selection of μ in our application is made with two objectives in mind: it must be large enough to be able to track fast changes in the relative volumes and at the same time small enough to be able to ignore the changes due to normal breathing. In our experiments we used a value of $\mu = 0.01$, which seems to provide a good compromise between the two conflicting requirements.

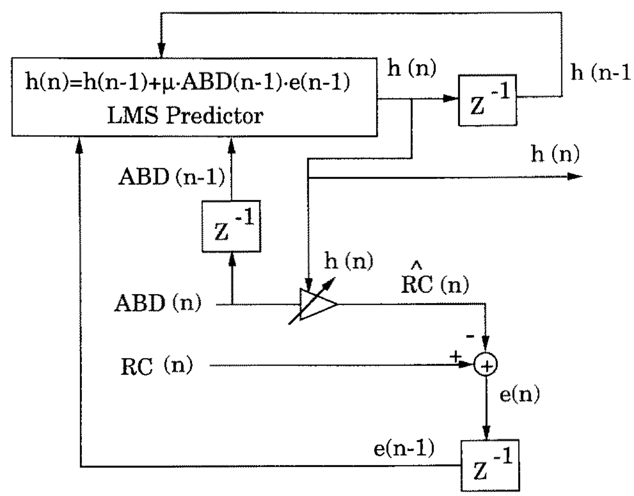

Fig 2. Adaptive filter processor diagram. LMS = *least-mean-squares;* ABD = *abdomen;* RC = *rib cage;* h(n) = *adaptive filter coefficient [*(n) = *0, 1, 2, . . .];* μ = *constant with value of 0.01;* e(n) = *error between true and estimated values of rib cage volume-motion coefficient;* Z^{-1} = *delay (i.e., taking the* $n - 1$ *value);* $\widehat{RC}(n)$ = *estimate of rib cage volume. The equations are* $h(n) = h(n - 1) + \mu \times ABD(n - 1) \times e(n - 1)$ *and* $e(n) = RC(n) - \widehat{RC}(n)$.

For each individual person, the relationship and relative contribution of each component, rib cage and abdomen, is unique. Movements, changes in posture, and changes in breathing patterns all affect the proportions the rib cage and abdomen contribute to the total tidal volume. The filter thus automatically "adapts" to these variations in individual patterns.

Figure 2 is a block diagram of the least-mean-square adaptive filter. The filter coefficient h(n) changes with movements, as described above. After a movement, the coefficient returns to the baseline value. These changes in the adaptive filter coefficient are used as indicators for patient movements.

After approval of our Institutional Review Board for human research, informed consent was obtained from 20 healthy male volunteers aged 21 to 30 years. The VIP bands were placed around the rib cage and abdomen. Airway flow was measured simultaneously throughout the session at the mouth with a Fleisch pneumotachograph. A standard mouthpiece and nose clip were used. The volunteers were asked to perform eight specific movements between 3-minute epochs of quiet supine breathing. The movements during the 20-minute session were (1) raising arms once slowly, (2) raising arms once quickly, (3) raising arms and moving them several times, (4) raising legs once slowly, (5) raising legs and moving them several times, (6) sitting up, (7) breathing deeply and rapidly, and (8) rolling from a supine to lateral decubitus position.

The data were processed in two different ways. First, the rib cage and abdominal signals were processed by using the adaptive filter, and the filter coefficient h(n) was calculated for all points in time. Next, flow from the pneumotachograph, rib cage, and abdomen signals were used to determine the instantaneous values of the calibration coefficients (b and c) by using the flow method (equation 2). A multilinear regression was done with the three signals to calculate b and c. This procedure was performed on all data collected during the study session. A window of 128 points was used for each regression. The window was shifted by one sample interval and the regression repeated until all the data were processed. This provided a continuous evaluation of the parameters b and c, which relate rib cage and abdominal signals to flow.

The first-order adaptive filter coefficient h(n), the coefficients b and c as functions of time, and the event marker were plotted. The results were scored to determine the number of events correctly detected and the number of events during which the calibration changed.

RESULTS

There were a total of 166 movements. The calibration coefficients changed dramatically in 146 (88%) of the 166 movements. The changes lasted for the duration of the movements and returned to baseline within two to three breaths. The changes in the coefficients were five or more times larger than the variability around baseline during quiet, supine breathing. The filter was able to detect 94% of these 146 movements.

The calibration coefficients were different for each individual. There were variations of 10 to 20% around the filter output baseline calibration coefficients during quiet, supine breathing. Over all 20 volunteers' sessions, there were 27 instances in which no movement occurred, but the calibration coefficients changed dramatically (more than the baseline variation for that individual). The duration of these instances was two breaths or less.

Figure 3 and the Table summarize the results. The calibration changed during 78% of the upper extremity movements, and the filter detected 87% of these. Ninety percent of the lower extremity movements caused a change in calibration, and the filter detected 94% of these movements. All of the total body movements and sudden changes in breathing pattern (breathing deeply and rapidly) caused a change in calibration

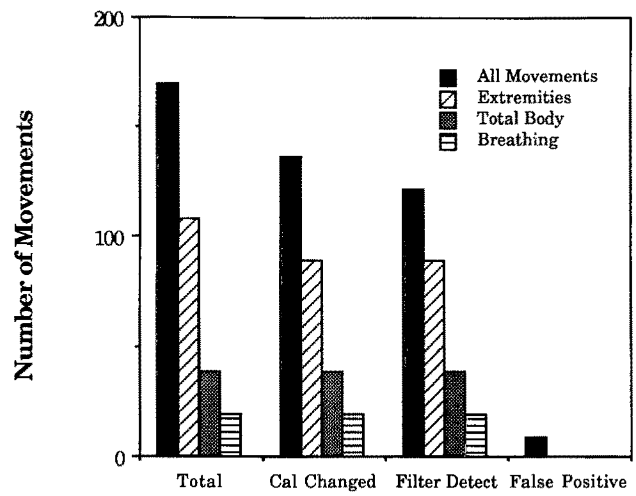

Fig 3. Performance of adaptive filter processor: the total number of movements, the number where the calibration (Cal) *changed, the number the filter accurately detected* (Detect), *and the number of false positives, for the different types of movements.*

Summary of Movements Performed and Adaptive Filter Results

	No. of Movements		
Activity	Total	Filter-detected	Where Calibration Changed
Raising/moving extremities			
Upper	69	46	53
Lower	40	34	36
Total body movements			
Rolling to side[a]	19	19	19
Sitting[a]	19	19	19
Breathing deeply and rapidly[a]	19	19	19

[a] One subject did not perform these movements.

coefficients. The filter accurately detected 100% of these movements and changes.

Figure 4 shows a representative interval containing movements unrelated to breathing. The movements are documented on each plot. The adaptive filter output h(n) is plotted with the calculated calibration coefficients; the calibration coefficients are the rib cage gain in liters per volt (c) and the abdomen gain in liters per volt (b) (see equations 1 and 2). The first movement is sitting up on the bed from a supine position, then lying down. The second is breathing deeply and rapidly. When the volunteer sat up, b decreased, c increased, and h(n) decreased. As the volunteer returned to a supine position, b once again decreased as did c, and h(n) increased. For this individual, breathing deeply and rap-

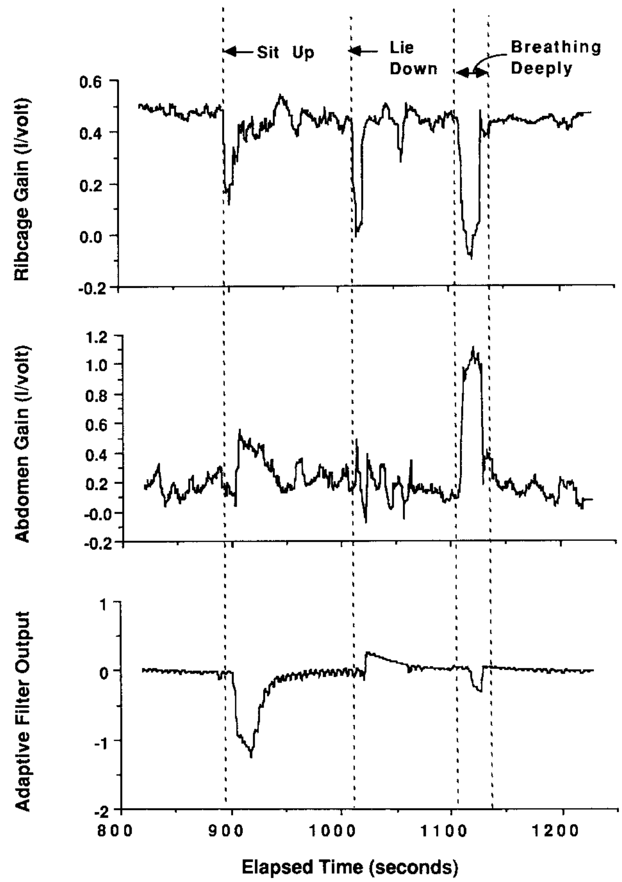

Fig 4. Example of the effects of different movements on the accuracy of ventilatory inductive plethysmography (VIP) and the detection of these events by the adaptive filter. The rib cage gain and abdomen gain are the calibration coefficients (see equations 1 and 2 in the text), which relate instantaneous flow measured at the mouth with the rib cage and abdominal signals. The assumption is that these are always constant. As can be seen, sitting up, lying down, and breathing deeply can dramatically alter these calibration values. These movements would introduce large errors in the VIP measurements. The direction and magnitude of the changes in the calibration values are dependent on the variable effects of nonrespiratory skeletal muscle activity on the VIP bands. The adaptive filter output changes dramatically from its baseline value whenever the calibration values change. The magnitude and direction of the changes are not important. The occurrence of any change from baseline indicates a potential source of error in the VIP data. By observing the adaptive filter output, one can reject all data that occurred during these deviations from baseline.

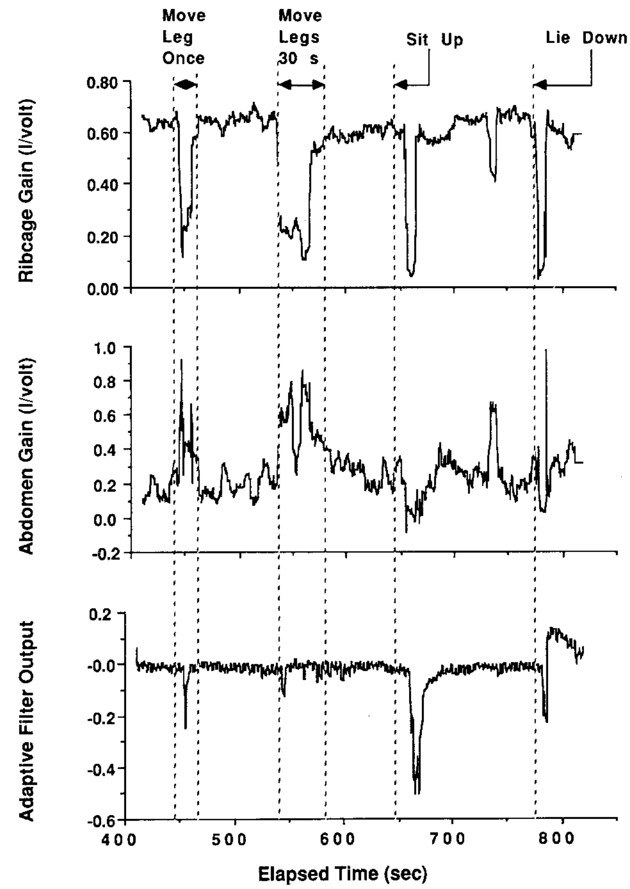

Fig 5. Another example of the effects of different movements on the accuracy of ventilatory inductive plethysmography (VIP) and the detection of these events by the adaptive filter. As can be seen in this example, sitting up, lying down, and leg movement can dramatically alter the calibration values. Leg movements for 30 seconds produced an initial change in the adaptive filter output; however, the adaptive portion of the filter rapidly changed to the new "quasi"–steady-state of leg movement. The speed at which it adapts can be adjusted and may be too rapid for this situation. See legend for Fig 4 for further explanation.

idly caused b to decrease and c to increase while h(n) decreased. The direction of change of b and c is determined by the change in the relative contributions of the rib cage and abdomen to breathing during each of these activities. For each of these, the rib cage increased and the abdomen decreased. A possible explanation is that during these activities, the abdominal muscles are contracted and used in the body movements. Thus, the contribution to breathing is less, or is at least perceived as less by the VIP monitor. The amount of change in h(n) is determined by the relative change in the abdomen and rib cage coefficients. For this initial evaluation, we were not concerned with the magnitude of the change other than whether it was detectable.

Figure 5 is a similar plot of lower extremity movements. The first movement is raising the legs once. The second is raising the legs and moving them for 30 seconds. The changes in the calibration coefficients or gain with movements are large enough to be detected easily by inspection. In some cases, the changes are more subtle, but are detectable. Moving the legs once caused b to decrease, c to increase, and h(n) to decrease for this individual. Moving the legs for 30 seconds caused the same changes, although the change in h(n) was less dramatic. When this individual sat up, the same changes occurred as with the individual in Figure 4, that is, b decreased, c increased, and h(n) decreased. When this subject laid down, b, c, and h(n) all decreased.

DISCUSSION

We have developed a functional adaptive processor capable of retrospective identification of movements unrelated to respiration. This will enable the user to discard the data collected during patient movements and use only the information that is known to be reliable when using VIP.

It has been well documented that breathing either through a mouthpiece with the nose occluded by a nose clip or through a face mask induces variable changes in tidal volume, minute ventilation, and respiratory rate [1–6]. The VIP system allows nonobtrusive measurement of tidal volume, minute ventilation, respiratory rate, mean inspiratory flow, and inspiratory and expiratory times [7,16,18–20]. VIP can be used in a variety of situations, which include documentation of ventilatory depression due to drugs [21], ventilatory pattern analysis [18,19], detection of central and obstructive apnea [16,20], and evaluation of exercise [9]. Reliable, noninvasive ventilatory monitoring is very important in patients who have had epidural narcotics for pain relief and for patients who are using patient-controlled analgesia (PCA) devices [22].

Movements that are unrelated to ventilation and that cause changes in the VIP signals from the chest or rib cage and/or abdomen comprise one of the biggest sources of error in VIP. The calibration coefficients for VIP are valid if the patient remains in the same position as during calibration. Both the volume and flow techniques for calibration determine the relative contributions of the rib cage and abdomen to the total volume. The calibration coefficients, or "volume-motion" coefficients, are calculated from these relative contributions [23]. However, positional or postural changes as well as any movements unrelated to breathing will alter the relative contributions of the rib cage and abdomen [7,23,24]. Because the relative contributions are changed, the calculated calibration coefficients are no longer valid. With any method of calibration there will be some errors associated with the tidal volume calculation. In normal subjects, Stradling et al [24] report errors as low as 3.5% when using a single-position flow technique, and Zimmerman et al [7] report errors as high as 23% when using a two-position isovolumetric technique. The error is increased from 3.5 to 9.5% in patients with chronic airway obstruction [7]. The error is increased to between 14.3 and 17.9% in normal subjects as their position changes, and becomes as high as 40% in patients with respiratory disease [23]. These increased errors and errors with the two-position calibration technique can be attributed to changes in the volume-motion coefficients, which are not accounted for during calculation of tidal volumes [7,8,23,24]. These problems have severely restricted the use of VIP in long-term monitoring or in subjects who are not at rest.

This adaptive processor represents a step toward an improvement over previously available technology in VIP. As seen in the Table, a high percentage of extremity movements caused changes in calibration coefficients and affected the accuracy of ventilatory variables. All of the total body movements caused a change in calibration coefficients.

There were some changes and variations in the calibration coefficients when no movements took place. The magnitude and duration of the changes were much smaller (two to three times) than during a movement. These changes are most likely due to variations in the breathing pattern (i.e., rib cage and abdomen contributions) that were not detected by the observer. The current filter did not detect these changes. The time constant of the filter can be adjusted to respond more quickly and perhaps detect these shorter changes. It may be important in some clinical situations to see subtle variations in breathing patterns, especially with respiratory depression due to narcotics, barbiturates, or other medications.

In the past, clinicians have avoided the use of VIP because of the inaccuracies in calibration and measurements as a subject moves about or changes position. It is cumbersome and impractical to have an observer always at the bedside to note movements unrelated to breathing or to recalibrate the VIP system when movements or position changes occur. The adaptive filter processor can automatically detect these movements and position changes, an important first step in making the VIP system a more usable monitor.

Implementing the filter will improve the accuracy of the monitor by detecting 83% of movements that are unrelated to breathing and that cause a change in calibration. This may improve the usefulness for long-term monitoring of patients who are receiving epidural narcotics for pain control or who are using a patient-controlled analgesia device. It also makes monitoring infants and children more practical, since these patients cannot always cooperate during a study session. More accurate ventilatory assessments could also be made during sleep studies, pulmonary evaluations, or exercise evaluations.

The adaptive filter processor has recently been implemented in real time at our institution. The processor will be used for studies on the effects of anesthetic agents on ventilation. Future goals for VIP signal processing include use of the adaptive filter signals along with the rib cage and abdominal signals to correct for changes in calibration coefficients. Then, the motion artifact data may still be useful in terms of calculating respiratory variables. We are currently working with advanced processing techniques to assess the possibilities of achieving this goal.

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