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Capnography and chest wall impedance algorithms for ventilation detection during cardiopulmonary resuscitation

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Abstract

Objective—Hyperventilation is both common and detrimental during cardiopulmonary resuscitation (CPR). Chest wall impedance algorithms have been developed to detect ventilations during CPR. However, impedance signals are challenged by noise artifact from multiple sources, including chest compressions. Capnography has been proposed as an alternate method to measure ventilations. We sought to assess and compare the adequacy of these two approaches.

Methods—Continuous chest wall impedance and capnography were recorded during consecutive in-hospital cardiac arrests. Algorithms utilizing each of these data sources were compared to a manually determined “gold standard” reference ventilation rate. In addition, a combination algorithm, which utilized the highest of the impedance or capnography values in any given minute, was similarly evaluated.

Results—Data were collected from 37 cardiac arrests, yielding 438 min of data with continuous chest compressions and concurrent recording of impedance and capnography. The manually calculated mean ventilation rate was 13.3 ± 4.3 /min. In comparison, the defibrillator’s impedance-based algorithm yielded an average rate of 11.3 ± 4.4 /min ($p=0.0001$) while the capnography rate was 11.7 ± 3.7 /min ($p=0.0009$). There was no significant difference in sensitivity and positive predictive value between the two methods. The combination algorithm rate was 12.4 ± 3.5 /min ($p=0.02$), which yielded the highest fraction of minutes with respiratory rates within 2/min of the reference. The

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Conflict of Interest

Dr. Eilevstjønn is employed by Laerdal Medical. Drs. Edelson and Abella have received research funding from Philips Healthcare and other research support from Laerdal Medical. Drs. Edelson and Abella have received speaking honoraria from Philips and Zoll Medical Corp, as well as research funding from the American Heart Association. Dr. Abella has additionally received research funding from the National Institutes of Health and Cardiac Science Corp.

impedance signal was uninterpretable 19.5% of the time, compared with 9.7% for capnography. However, the signals were only simultaneously non-interpretable 0.8% of the time.

Conclusions—Both the impedance and capnography-based algorithms underestimated the ventilation rate. Reliable ventilation rate determination may require a novel combination of multiple algorithms during resuscitation.

Keywords

cardiopulmonary resuscitation; cardiac arrest; capnography; artificial respiration; chest wall impedance

1. Introduction

Hyperventilation is common during cardiopulmonary resuscitation (CPR) for both in-hospital and out-of-hospital cardiac arrests, even among highly trained rescuers.^{1, 2} Despite resuscitation guideline recommendations to ventilate intubated patients at 8–10 per minute,³ clinical studies have often documented ventilation rates of over 30 per minute, which have been shown to impair hemodynamics and worsen outcomes in animal models of cardiac arrest.^{4, 5}

Devices have been developed to monitor ventilation rate and volume and deliver real-time feedback to rescuers providing ventilations,^{6, 7} but the efficacy of these devices depends upon the accuracy of ventilation detection. CPR-feedback devices routinely rely on changes in chest wall impedance (CWI) obtained via defibrillation pads to detect ventilations, but the accuracy of CWI signals is compromised by noise artifact from multiple sources, including chest compressions.^{6, 8} Capnography has been proposed as an alternate method of measuring ventilations during CPR, and has been used extensively in laboratory investigations of cardiac arrest.⁹ However, there is a paucity of data regarding the relative accuracy of CWI-based recording and capnography in measuring ventilation rates during actual resuscitation efforts. We hypothesized that capnography would be more accurate than CWI for measuring ventilation rate during in-hospital cardiac arrest and that both signals would be less accurate than an algorithm which utilized a combination of both signals.

2. Methods

2.1. Study setting and design

A prospective observational study of in-hospital cardiac arrests was conducted from July 2007 to January 2008. The study protocol was approved by the Institutional Review Board of the University of Chicago Medical Center, and all data collection was compliant with the Health Insurance Portability and Accountability Act of 1996 (HIPAA).

The study was performed in an academic, tertiary care facility with approximately 600 inpatient beds. Cardiac arrests occurring in the hospital are treated by a multidisciplinary resuscitation team, the members of which are Advanced Cardiac Life Support (ACLS) certified.

All consecutive adult patients suffering cardiac arrest were considered for analysis. Cardiac arrest cases were defined by the loss of a pulse, requiring the delivery of chest compressions. Cases were only included if resuscitation attempts had continuous capture of capnography and CWI data.

2.2. Data collection

We utilized a CPR-sensing monitor/defibrillator (MRx-QCPR, Philips Healthcare, Andover, MA) with continuous sidestream CO₂ recording capabilities (Microstream, Oridion Medical, Jerusalem, Israel) connected in line with the endotracheal tube. As such, all patients were intubated and manually bag ventilated. The device provided real-time audiovisual feedback to rescuers regarding CPR quality during each resuscitation, relying on CWI measurements via the defibrillator pads to detect ventilation rate. Additional patient information was collected from medical chart abstraction and review of hand-recorded cardiac arrest forms completed prospectively during resuscitation care.

2.3. Ventilation annotation

CPR performance data, including capnography and CWI measurements, were downloaded after resuscitation events and analyzed using custom software (QCPR Review, Laerdal Medical, Stavanger, Norway). Capnography recordings were preprocessed by time-shifting the data 3.5 sec to synchronize the capnograph with the impedance signal. Study personnel, blinded to the defibrillator's determination of ventilations, manually annotated ventilations within arrest transcripts using both the capnograph and CWI waveforms. Ventilations were only annotated during periods of active CPR with concurrent impedance and capnography waveforms. Annotators used the peaks of the impedance waveform and/or the troughs of the capnography waveform, as shown in Figure 1, to identify ventilations. Each arrest transcript was annotated independently by two investigators (EKW, ER), and the resulting annotations were compared for ventilation agreement. Annotation discrepancies were resolved by a third investigator (DPE) and then confirmed by a fourth investigator (JE). These consensus manual ventilation annotations were deemed our "gold standard" for comparing the relative performance of CWI and capnography-based ventilation detection.

2.4. Ventilation detection algorithms

The impedance-based algorithm used in this study was run offline with signal analysis software (Matlab, MathWorks, Inc., Natick, MA) using the impedance data recorded by the study defibrillator. This algorithm is very similar to what is implemented in the second generation QCPR sensor used with the HeartStart MRx (Philips Healthcare, Andover, MA). The impedance signal is first filtered using static and adaptive filters. The adaptive filter uses force and acceleration from the chest compression sensor to remove compression artifacts in a similar fashion to prior work with ECG filtering.^{10, 11} The state machine like detection algorithm itself, shown in Figure 2A, operates sample for sample in the time-domain with signal conditions determining stay or jump to another state. Basically, the algorithm looks for waves in the impedance signal with an amplitude of 0.25 to 4.5 ohm, an inflation time of 0.3 – 3 seconds, rise/fall rates of 0.2 – 5 ohm/s and a ventilation peak to next ventilation peak time of 1 s or more.

A novel capnography-based ventilation detection algorithm was developed and performed offline with the same signal analysis software (Matlab, MathWorks, Inc., Natick, MA) using the capnography data recorded by the defibrillator. This algorithm is similar to that for the impedance and shown in Figure 2B. Principally, the onset of a ventilation is determined by the abrupt decrease in the capnography signal to below a threshold value of 2 mm Hg. To detect a ventilation, the signal has to remain below this threshold value for >0.5 sec (with a ventilation inspiration time between 0.3 and 5 sec and time since last low period > 0.4 sec) before increasing to a higher CO₂ value and finally flattening out.

2.5. Data analysis

Both of the ventilation detection algorithms evaluated in this study provided time stamps for all ventilations detected. For the ventilation detection performance these time stamps were matched with the manual (“gold standard”) annotations if they were within 0.8 sec of one another.

Performance was evaluated in terms of sensitivity and positive predictive value (PPV). Sensitivity was defined as the proportion of manual annotations found by the automatic detection algorithm. PPV was defined as the proportion of the automatic detections that were also manually annotated. Specificity and negative predictive value determination for ventilation detection was not meaningful because there are not a definite number of non-ventilations. Hyperventilation was defined, in accordance with consensus guidelines for intubated patients, as greater than 10 ventilations/min.³ Sensitivity for hyperventilation detection was defined as the proportion of minutes with manually determined hyperventilation where the impedance or capnography algorithms also registered hyperventilation. Specificity for hyperventilation detection was defined as the proportion of minutes without hyperventilation, according to the manual method, which also registered ventilation rates ≤ 10 /min by the automatic detection strategies. In addition, we evaluated a simple algorithm combination strategy. The combination strategy chose the higher rate of the CWI or capnograph-based algorithm in every one minute period.

Analyses were performed using a statistics application (Stata version 10.0, Statacorp, College Station, TX). All ventilation calculations were performed per resuscitation. Normally distributed measures, such as ventilation rate, were reported as mean $\pm$ standard deviation (SD) and compared using paired two-sided Student’s t-tests. Skewed data, such as sensitivity, specificity, and PPV, were reported as median [q25, q75] and compared using Wilcoxon signed-rank tests. Significance was set at $p < 0.05$ for all analyses.

3. Results

Between July 2007 and January 2008, data were collected from 95 cardiac arrests in 75 hospitalized patients. Of these cases, 58 were excluded because compression, ventilation, and capnography waveforms were not captured concurrently. The 37 remaining cases from 33 patients yielded 438 min of data with continuous chest compressions and concurrent recording of both CWI and capnography data. The patient cohort represented in this sample was generally similar to our previously described in-hospital arrest study populations^{6, 12} (see Table 1). Return of spontaneous circulation (ROSC) was achieved in 15/37 (40.5%) cases, and 3/33 (9.1%) patients survived to hospital discharge.

Aggregating the 37 resuscitations, the impedance signal was uninterpretable 19.5% of the time, compared to 9.7% of the time for the capnography signal. However, the signals were only simultaneously non-interpretable 0.8% of the time. Per patient statistics are included in Table 2.

The mean ventilation rate, when ventilations were manually annotated, was 13.3 ± 4.3 /min. By comparison, the defibrillator’s impedance-based algorithm yielded a rate of 11.3 ± 4.4 /min ($p = 0.002$), which underestimated the consensus annotated ventilation rate by 1.9 ± 3.6 /min. Meanwhile, the capnography-based rate was 11.7 ± 3.7 /min ($p = 0.0009$), which underestimated the consensus annotated respiratory rate by 1.6 ± 2.6 /min. These two measured rates were not significantly different from one another ($p = 0.57$), as shown in Table 2. Likewise the two algorithms were similar with respect to sensitivity and positive predictive value for ventilation detection and sensitivity and specificity for hyperventilation detection. Hyperventilation was noted at some point in 35/37 resuscitation attempts.

When the higher of the impedance and capnography calculated rates was used in any given minute, the combination strategy yielded a mean ventilation rate of 12.4 ± 3.5 /min ($p=0.02$) per resuscitation. Hyperventilation detection sensitivity for the combination strategy was 1.00 [1.00, 1.00] and specificity 0.96 [0.61, 1.00] (all $p<0.02$ vs. the other algorithms). The histogram of rate deviations from the manually annotated ventilation rate for each of the 438 one-minute segments is shown in Figure 3 for all three methods of calculation. Under-reporting and over-reporting of the rate are present in each ventilation calculation method. The tendency of under-reporting the rate is most prominent with the CWI-based algorithm, while over-reporting was seen the most with the combination algorithm.

4. Discussion

In this study, we have developed an algorithm for measuring respiratory rate from a capnography signal during actual cardiac arrests and compared it to an established chest-wall impedance based algorithm. We have shown that both algorithms frequently underestimate ventilation rates but demonstrate that combining the algorithms yields a more robust signal for hyperventilation detection. This study represents the first to demonstrate the feasibility of a capnography-based ventilation detection algorithm as well as a comparison of different methods to quantify ventilation rate during actual in-hospital cardiac arrest.

Recent investigations have highlighted the need for increased attention to CPR quality during actual resuscitation care and the need to measure parameters of CPR performance. Specifically, ventilation rate has been found to be much higher in actual practice than recommended by resuscitation guidelines. One in-hospital investigation found widely variable ventilation rates during CPR, with a mean rate of 37 ± 13 /min.⁴ Paramedics in an out-of-hospital observational cohort performed ventilations at a rate of 30 ± 3.2 /min; when replicated in a swine model of cardiac arrest, this rate was highly detrimental to both hemodynamics and survival.⁵ Finally, in our previous work, the in-hospital ventilation rate during a cardiac arrest patient cohort was found to be 21 ± 12 /min,¹ which improved somewhat, although considerably less than other CPR quality parameters, with provision of audiovisual feedback and post-event debriefing, possibly due to missing feedback in the setting of unrecognized hyperventilation by the defibrillator.^{6, 12} In this current analysis we found that hyperventilation occurred at some point in 35/37 resuscitation attempts, despite the high level of training in our rescuers and quality improvement emphasis on preventing hyperventilation.

The recording of CPR quality parameters, such as ventilation rate, has become clinically available through a variety of devices including both CPR-sensing defibrillators and free-standing measurement tools.¹³ Measurement of chest compression parameters, such as rate and depth, can be measured via chest wall force transducers and/or accelerometers placed under the hands of a rescuer performing compressions; this method has been widely applied in a number of studies^{1, 14–17}. In contrast, measurement of ventilation rate has been less well characterized, in part due to limitations of clinically available methods. While capnography has been extensively used during laboratory studies of CPR,^{9, 18, 19} and has been used as a predictor of resuscitation success during human cardiac arrest,^{20–22} the use of capnography to monitor ventilation rate during actual CPR has not been established. Given the important impact of ventilation rate on hemodynamics and clinical outcomes, careful measurement of ventilation rate to allow for real-time feedback remains an important goal to improve resuscitation care.

CWI is an appealing method for automatically measuring ventilation rate, since it is theoretically available during any resuscitation in which adhesive defibrillation pads are used.⁵ However, several factors affect its sensitivity for ventilation detection. The two most common problems that we observed with chest wall impedance signals (Figure 1b) were noise

(presumably due to chest compressions and/or handling of the patient) and small impedance changes (presumably due to either low volume ventilations or patient body habitus). In addition, the reliability of CWI can be affected by pad positioning and/or poor adhesion of the defibrillator pads, which will attenuate impedance measurements.

Capnography, while a more direct measure of ventilation, is limited as well. This signal may also exhibit compression artifacts or noise (see Figure 1c) and can be attenuated by blood or secretions in the endotracheal tubing, a fairly common occurrence during resuscitation.²³ In addition, it is currently less widely available during resuscitations since it requires additional instrumentation beyond defibrillator pads. Furthermore, because capnography is dependent on perfusion and metabolism, it runs the risk of being too low to show changes that would indicate a ventilation. In our cases only one patient demonstrated occasional readings lower than 2 mmHg, which made detection using a 2 mmHg drop highly feasible.

Our results suggest that an algorithm combining the impedance and capnography signals improves ventilation detection, as there is less chance of both signals being compromised by artifact at the same time. We tested a simplistic combined method in this study, which utilized the highest of the two calculated rates in any minute, with the result of less under-reporting of ventilations, but at the cost of increased intermittent over-reporting. Given the known hazards of hyperventilation, erring on the side of increased ventilation detection is likely warranted but future work should seek to optimize a combination algorithm that concurrently takes into account both data streams.

A feasible alternative, or supplement, to both impedance and capnography based methods might be ventilation detection through direct measurement of airway pressure.¹³ Airway pressure detection is currently employed in respiratory profile monitors, and could also be incorporated into defibrillators or resuscitation ventilation equipment. We speculate that the rate of ventilations might be most reliably measured using airway pressure, but this remains to be investigated during the clinical delivery of CPR.

This study has several limitations. First, our “gold standard” ventilation detection, to which we compared CWI and capnography detection, was not independent of those measures. A better gold standard might have been airway pressure monitoring pressure, as described above, or visual recording of respirations during the resuscitation. However, neither of these were available in this study. We used two independent reviewers to manually record each ventilation, with two additional reviewers resolving any discrepancies in favor of detecting only “clear” ventilations, to minimize error. However, our gold standard likely missed some ventilations, which would result in overestimation of the sensitivities in this study. Second, we report only one algorithm for each data stream, both of which could likely be refined. It is important to note that defibrillator impedance measurements were originally designed for detection of defibrillator pad connections, with detection of ventilations only as a secondary development. Therefore, to obtain a more robust signal, the impedance measurement hardware, including electronics (excitation frequency and current) and defibrillation pads (size, type and position), could be optimized for detection of ventilations as a primary function. Finally, we did not account for resuscitation level characteristics such as defibrillator pad placement and adhesion, body habitus, and endotracheal tube fluid which could have affected our measures. Future work with a larger sample size will be required to understand the effects of these variables on CWI and capnography.

5. Conclusions

The important goal of minimizing hyperventilation during CPR requires robust ventilation detection methods, yet currently employed chest wall impedance methods have important

limitations which result in decreased ventilation detection. The use of capnography or pressure measurements, clinically available in a variety of monitor/defibrillators and stand alone devices, might allow for the development of improved algorithms for ventilation detection. Best performance is likely achieved using a combination of data streams.

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Signal clarity

Figure 1.

Examples of capnography (cap) and compression-filtered chest-wall-impedance (CWI) waveforms. Vertical grey lines represent 5 sec time markers. The vertical markers indicate manually annotated ventilations. (a) Good signal for both impedance and capnography. (b) Noisy impedance, clear capnography signal. (c) Clear impedance, noisy capnography signal.

A. **B.** **Figure 2.**

Flow chart of the impedance-based ventilation detection algorithm (A) and capnography-based ventilation detection algorithm (B).

Figure 3. Distribution of ventilation rate deviation from reference by calculation method [n=438 minutes].

Table 1

Baseline characteristics.

Cardiac arrests, n	37
Distinct patients, n	33
Total minutes of data, n	438
Manually identified ventilations, n	5316
Minutes of data per arrest, mean (range)	11.3 (0.9 – 29.3)
Reference ventilation rate, mean $\pm$ SD, /min	13.3 $\pm$ 4.3
Female sex	12/33 (36.4)
Patient age, mean (range), y	59 (19–87)
Initial rhythm:	
Ventricular fibrillation/tachycardia	6/37 (16.2)
Pulseless electrical activity	26/37 (70.3)
Asystole	5/37 (13.5)
Location:	
Intensive care unit	22/37 (59.5)
Ward	15/37 (40.5)
Return of spontaneous circulation	15/37 (40.5)
Survival to discharge	3/33 (9.1)

Results are shown as n/total (%) unless otherwise stated. Abbreviations: SD, standard deviation.

Table 2

Ventilation detection accuracy by method of calculation.

Parameter	Impedance-based algorithm	Capnography-based algorithm	p-value
Calculated ventilation rate, /min	11.3 ± 4.4	11.7 ± 3.7	0.57
Ventilation rate deviation from reference, /min	-1.9 ± 3.6	-1.6 ± 2.6	0.57
Fraction unreadable to human experts	0.05 [0.00 0.22]	0.00 [0.00 0.08]	0.08
Ventilation detection:			
Sensitivity	0.78 [0.67, 0.89]	0.82 [0.75, 0.93]	0.34
Positive predictive value	0.87 [0.77, 0.96]	0.91 [0.85, 0.95]	0.15
Hyperventilation detection:			
Sensitivity [n=35]	0.81 [0.51, 1.00]	0.86 [0.67, 1.00]	0.44
Specificity [n=32]	1.00 [0.98 1.00]	1.00 [0.77 1.00]	0.32

Rates are shown as mean ± standard deviation and performance indicators as median [q25, q75]. n=37 patients except in the case of hyperventilation detection where cases without identified hyperventilation by manual (sensitivity) or test algorithm (specificity) were excluded from analysis.