Variations in Respiratory Excretion of Carbon Dioxide Can Be Used to Calculate Pulmonary Blood Flow

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Variations in Respiratory Excretion of Carbon Dioxide Can Be Used to Calculate Pulmonary Blood Flow

David A. Preissa, Takaumi Azamib, Richard D. Urmanac, d

Abstract

Background: A non-invasive means of measuring pulmonary blood flow (PBF) would have numerous benefits in medicine. Traditionally, respiratory-based methods require breathing maneuvers, partial rebreathing, or foreign gas mixing because exhaled CO2 volume on a per-breath basis does not accurately represent alveolar exchange of CO2. We hypothesized that if the dilutional effect of the functional residual capacity was accounted for, the relationship between the calculated volume of CO2 removed per breath and the alveolar partial pressure of CO2 would be reversely linear.

Methods: A computer model was developed that uses variable tidal breathing to calculate CO2 removal per breath at the level of the alveoli. We iterated estimates for functional residual capacity to create the best linear fit of alveolar CO2 pressure and CO2 elimination for 10 minutes of breathing and incorporated the volume of CO2 elimination into the Fick equation to calculate PBF.

Results: The relationship between alveolar pressure of CO2 and CO2 elimination produced an R2 = 0.83. The optimal functional residual capacity differed from the “actual” capacity by 0.25 L (8.3%). The repeatability coefficient leveled at 0.09 at 10 breaths and the difference between the PBF calculated by the model and the preset blood flow was 0.62 ± 0.53 L/minute.

Conclusions: With variations in tidal breathing, a linear relationship exists between alveolar CO2 pressure and CO2 elimination. Existing technology may be used to calculate CO2 elimination during quiet breathing and might therefore be used to accurately calculate PBF in humans with healthy lungs.

Keywords: Pulmonary blood flow; Cardiac output; Carbon dioxide elimination; End-tidal carbon dioxide

Introduction

A clinician’s ability to determine a patient’s cardiac output is critical in the assessment of cardiopulmonary health. Clinically, this measurement can be required intermittently during surgery, as well as post-operatively. Cardiac output may also be measured in research settings, but doing so with invasive methods requires expensive critical care equipment, trained personnel, a sterile environment and readily available resuscitative equipment.

Cardiac output is arguably one of the most important parameters reflecting cardiovascular health and yet its measurement is currently limited to patients with Swan-Ganz catheters in the operating room, critical and intensive care units. To date, no non-invasive method has proven itself comparable to thermodilution in accuracy, repeatability, and in physiological soundness [1, 2]. Nevertheless, the thermodilution method is plagued with limitations in clinical practice, the most important ones being associated with its invasiveness [3].

Methods for measuring cardiac output non-invasively can be classified as respiratory-based or non-respiratory-based. The latter, which includes Doppler (external, transtracheal and transesophageal), bioimpedance, and pulse contour analysis, has not yet been accepted into clinical practice because of either theoretical or practical limitations [4]. Additionally, there are some respiratory-based methods that utilize foreign gas breathing, including argon and acetylene, but these methods require a source of external gas, are impractical in many settings, and will not be dealt with in detail here [5].

The respiratory-based methods are the oldest, the most physiologically sound, and traditionally the most accepted method of cardiac output measurement. Their principles and assumptions are well understood as are their limitations, the most important of which is that they more accurately measure pulmonary blood flow (PBF) rather than cardiac output. Originally described for oxygen, Fick equation, developed in the late 1800s, is based on the measurement of elements of the mass balance across the lungs [6]. The following is the
The need for measuring CvCO₂ is accurate. Six methods have been presented, but none has proven itself repeatable and applicable to the model: (Eq. 3a)

\[ \text{PBF} = \frac{\text{VCO}_2}{(\text{CvCO}_2 - \text{CaCO}_2)} \]

where PBF is the pulmonary blood flow, VCO₂ is the metabolic CO₂ production, and CvCO₂ and CaCO₂ are the concentrations of CO₂ in the blood entering (mixed venous, v) and leaving (arterial, a) the lungs respectively. To keep the method purely non-invasive, alveolar pressure is conventionally used as a surrogate for content. The relationship between PaCO₂ and CaCO₂ is usually assumed to be linear in the physiologic range, and through previous in-vitro, CO₂ content of oxygenated blood can be estimated using the following equation [7, 8]: (Eq. 3b)

\[ \text{Content} = 4 \times \text{Pressure} + 260 \]

Generally, PaCO₂ is approximated from alveolar CO₂ pressure (P̄_A CO₂), which in turn is approximated from end-tidal CO₂ pressure (P̄_ET CO₂). Mixed venous pressure of CO₂ (PvCO₂) however, is difficult to obtain non-invasively, and has classically posed the biggest challenge to measurement of PBF. Respiratory maneuvers such as breathholding and rapid equilibration with external reservoirs have been employed to estimate PvCO₂, but these have the disadvantage of requiring an external supply of CO₂, and may pose a challenge to patients with respiratory compromise [9]. Single-breath methods have been presented, but none has proven itself repeatable and accurate.

In 1980, a novel method of calculating PBF without the need for measuring CvCO₂ was described by Gedeon et al [9]. That method demonstrated that if a subject’s alveolar ventilation (VA) were acutely and transiently reduced, a step change in P̄_A CO₂ and VCO₂ would result (once steady state is reached, after a few breaths). At this point, assuming PBF and PvCO₂ had not changed in this short time (< 30 - 45 s), two Fick equations could be applied to the model: (Eq. 4a)

\[ \text{PBF} = \frac{\text{VCO}_2}{(\text{S} \times \text{PvCO}_2 - \text{S} \times \text{P}_A \text{CO}_2)} \]

\[ \text{PBF} = \frac{\text{VCO}_2}{(\text{S} \times \text{PvCO}_2 - \text{S} \times \text{P}_A \text{CO}_2)} \]

where VCO₂, and P̄_A CO₂ are the exhaled CO₂ volumes and alveolar CO₂ pressures after a new steady state has occurred, respectively, and S is a conversion constant to content. Since PvCO₂ and PBF remain unchanged with this maneuver, these equations together yield a single equation: (Eq. 4b)

\[ \text{PBF} = \frac{(\text{VCO}_2 - \text{VCO}_2')/(\text{S} \times \text{P}_A \text{CO}_2') - \text{S} \times \text{P}_A \text{CO}_2)}{\text{S} \times \text{PvCO}_2} \]

This way, a small, temporary change in VA can be used to calculate PBF without the need for invasive monitors. This method has been proven reliable in intubated patients by multiple studies [10-12].

Creative as this method is, however, there are several limitations to it. First, the time to reach the new steady state is a function of the subject’s VA and functional residual capacity (FRC). On average, this time is approximately 30 - 45 s, or about five breaths, which means that only two data points are used to calculate PBF before recirculation of arterial blood occurs, limiting the method’s accuracy. Second, because VA is required to be transiently reduced, and because the normal variability of breathing creates noise in the measurement signal, this method is limited only to intubated patients. Finally, P̄_A CO₂ must be allowed to re-equilibrate to steady state levels before a subsequent test can be performed, which may require an additional 60 - 120 s.

**Materials and Methods**

**Flux of CO₂ at the alveoli**

The method proposed here redefines the term VCO₂ as described in Fick’s equation. According to conventional understanding, VCO₂ is defined as the excretion of CO₂ past the lips, measured with a pneumotachometer and CO₂ monitor. However, exhaled CO₂ on a breath-by-breath basis seldom reflects actual metabolic CO₂ production or pulmonary capillary excretion into the lungs [13]. Instead, an average over several breaths is required for this estimation, as the FRC acts as a reservoir for CO₂. For example, a large breath that expunges a large volume of CO₂ past the lips would incorrectly reflect a large “production” of CO₂. The Fick equation, therefore,
would be more physiologically accurate if the term VCO\textsubscript{2M} represented the continuous flux of CO\textsubscript{2} from the blood into the alveoli (VCO\textsubscript{2A}), rather than the discrete, tidal excretion of CO\textsubscript{2} past the lips (VCO\textsubscript{2M}). Averaged over many breaths, VCO\textsubscript{2M} will accurately reflect VCO\textsubscript{2A}.

**Relationship between VCO\textsubscript{2A} and P\textsubscript{A}CO\textsubscript{2}**

After redefining VCO\textsubscript{2} in this manner, it becomes clear that there exists a linear relationship between the flux of CO\textsubscript{2} out of the blood (VCO\textsubscript{2A}) and P\textsubscript{A}CO\textsubscript{2} (Fig. 1).

At steady state VCO\textsubscript{2A} and P\textsubscript{A}CO\textsubscript{2} produce a single point on this line. The slope of this line, PBF, is proportional to the PBF. A second theoretical line exists on this diagram, connecting the origin with the steady state point. This second line is described by the basic physiologic equation:

Eq. 5: \[ VCO_{2A} = V_A \times P_{A}CO_{2}/713 \]

where \( V_A \) is the alveolar ventilation - the slope of the line. It becomes clear that VCO\textsubscript{2A} - not VCO\textsubscript{2M} - must be applied when describing these relationships, as washout of the FRC can confound the changes that relate these variables.

Gedeon’s method aims to use two steady state VCO\textsubscript{2M} points to determine the slope of the PBF line: one attained from the subject breathing at rest (where VCO\textsubscript{2M} only equals VCO\textsubscript{2A} if averaged over many breaths), and the second after a small change in \( V_A \) (measured after 4 - 5 breaths, when, after a transient period, VCO\textsubscript{2M} is assumed to be equal to VCO\textsubscript{2A}).

In an extreme case, if the breath were held for 30 s and a new steady state were reached, P\textsubscript{A}CO\textsubscript{2} would be equal to P\textsubscript{v}CO\textsubscript{2}. However, if VCO\textsubscript{2A} can be measured with each breath, one data point along this line could be produced with each exhalation, and the slope, PBF, could be calculated with sequential breaths. Variations in VCO\textsubscript{2A} are needed to explore this line, and the slope of the second line, \( V_A \), would vary with each breath due to normal variations in tidal volume and respiratory frequency. Since VCO\textsubscript{2A} likely fluctuates little during quiet breathing (as opposed to VCO\textsubscript{2M}), variability of breathing, which was once the system “noise”, therefore becomes important for measurement of PBF.

Once this line is observed over multiple breaths, PBF can be calculated by extrapolating to the x-intercept, P\textsubscript{v}CO\textsubscript{2}, and calculating PBF using equation 1.

**Method for calculating VCO\textsubscript{2A}**

The method described above requires calculation of the VCO\textsubscript{2A}, the flux of CO\textsubscript{2} across the alveolar membrane. Calculation for VCO\textsubscript{2A} has been described thoroughly elsewhere [13]. Briefly, a simple mass balance dictates that the flux of CO\textsubscript{2} into the alveoli from the blood is related to the flux of CO\textsubscript{2} at the mouth and the change of CO\textsubscript{2} stores in the lung:

Eq. 6: \[ VCO_{2A} = VCO_{2M} - \Delta VCO_{2S} \]

where VCO\textsubscript{2A} is the alveolar flux of CO\textsubscript{2}, VCO\textsubscript{2M} is the flux of CO\textsubscript{2} at the mouth, and \( \Delta VCO_{2S} \) is the change in lung stores of CO\textsubscript{2}. VCO\textsubscript{2M} is easily measured using a metabolic cart, which integrates exhaled CO\textsubscript{2} concentration and expiratory flow.

The change in lung stores of CO\textsubscript{2} from breath 1 to breath 2, \( \Delta VCO_{2S} \), can be described as the sum of two terms describing the change in CO\textsubscript{2} concentration at a constant V\textsubscript{A}, and the change in V\textsubscript{A} at a constant CO\textsubscript{2} concentration:

Eq. 7: \[ \Delta VCO_{2S} = V_A (P_{A}CO_{2}' - P_{A}CO_{2})/713 + P_{A}CO_{2}' \Delta V_A \]

where \( \Delta VCO_{2S} \) is the change in alveolar stores of CO\textsubscript{2}, \( V_A \) is the alveolar volume at the end of breath 1, P\textsubscript{A}CO\textsubscript{2} and P\textsubscript{A}CO\textsubscript{2}' are the fractions of CO\textsubscript{2} in the alveoli at the end of breaths 1 and 2, respectively and \( \Delta V_A \) is the change in alveolar volume from breath 1 to breath 2.

**Computer simulation**

For this study, a computer simulation of tidal breathing was designed to test the theory under ideal conditions using Microsoft Excel 2003 (Redmont, Washington, USA). Incremental calculations of lung CO\textsubscript{2} volume were made for each 0.001 min. Tidal breathing was simulated using variable inhaled and exhaled tidal volumes (±30%), allowing lung volume to return to a different FRC with each breath. Complete alveolar mixing was assumed, inspiratory and expiratory times were equal, inhalation and exhalation flows were linear, and alveolar dead-space and shunt were assumed to be minimal. P\textsubscript{A}CO\textsubscript{2} was recorded once per breath (the final P\textsubscript{A}CO\textsubscript{2} value at the end of exhalation) and VCO\textsubscript{2M} was calculated at the mouth during exhalation. The model was run over a period of 10 min (100 breaths), and P\textsubscript{A}CO\textsubscript{2}, VCO\textsubscript{2M}, Vt-in and Vt-out were recorded with each breath. V\textsubscript{A} for sequential breaths was calculated as \( V_{A} = V_{A}' + V_{t-out} - V_{t-out} \), where Vt-in and Vt-out are inhaled tidal volume and exhaled tidal volume, respectively. Detailed parameters are outlined in Table 1.

**P\textsubscript{A}CO\textsubscript{2}-ave versus P\textsubscript{ET}CO\textsubscript{2}**

As stated above, the diffusion of CO\textsubscript{2} across the alveolar membrane should form a linear relationship with the pressure of CO\textsubscript{2} in the alveoli. Therefore, measurements of P\textsubscript{ET}CO\textsubscript{2}, a single sample taken at the end of exhalation, would be inappropriate to use in equation 3 since it reflects an end-expiration value rather than an average (P\textsubscript{A}CO\textsubscript{2}-ave). In the computer model, an average P\textsubscript{A}CO\textsubscript{2} is easy to calculate, but this is not the case clinically, when P\textsubscript{ET}CO\textsubscript{2} samples are the only measurements

**Table 1. The Parameters Used in the Mathematical Model**

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<th>Parameter</th>
<th>Error</th>
<th>Value</th>
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<tr>
<td>( P_{v}CO_{2} ) (mm Hg)</td>
<td>None</td>
<td>50</td>
</tr>
<tr>
<td>Pulmonary blood flow (L/min)</td>
<td>None</td>
<td>6</td>
</tr>
<tr>
<td>Tidal volume (mL)</td>
<td>±30%</td>
<td>500</td>
</tr>
<tr>
<td>Respiratory frequency (/min)</td>
<td>None</td>
<td>12</td>
</tr>
<tr>
<td>FRC (L)</td>
<td>None</td>
<td>3</td>
</tr>
<tr>
<td>PBF-preset (L/min)</td>
<td>None</td>
<td>6</td>
</tr>
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</table>
readily available non-invasively. To estimate $P_A^{CO_2}$-ave for a single breath using only non-invasive data, we back-calculated to the $P_A^{CO_2}$ that might exist at peak inhalation and averaged this with $PETCO_2$. The $P_A^{CO_2}$ at peak inhalation might be estimated as:

\[
\text{Eq. 8: } (\text{Volume of CO}_2 \text{ in Lung at End-Exhalation} - \text{Volume of CO}_2 \text{ Diffused during Exhalation} + \text{Exhaled Volume of CO}_2) \times \frac{713}{\text{Peak Alveolar Volume}}
\]

Averaging this value with $PETCO_2$ may produce a reasonable estimate of $P_A^{CO_2}$-ave which can be applied in equation 1.

**Calculation of $VCO_{2A}$**

$VCO_{2S}$ was calculated using equation 6 using sequential breaths, $\Delta V_A$ was assumed to be measurable without the need for nitrogen monitors ($Vt-in$ and $Vt-out$ were measurable), and $VA$ was initially assumed to be 3 L. $VCO_{2A}$ was then calculated using equation 5. Using 10 min of breathing data, when PBF and $PvCO_2$ were assumed to be in steady state, a plot of $VCO_{2A}$ versus $P_A^{CO_2}$ was created and $R^2$ was calculated for

![Figure 3](image1.png)

**Figure 3.** Alveolar partial pressure of $CO_2$ over a period of 2 min as predicted by the computer model.

![Figure 4](image2.png)

**Figure 4.** $VCO_2$ over time as measured at the mouth ($VCO_{2m}$, open circles), as pre-set by the model ($VCO_{2A}$-model, open squares) and as calculated using the proposed method ($VCO_{2A}$-calculated, closed circles).
the line of best fit. FRC was then iterated in increments of 0.25 L from 2 L to 4 L to achieve the most optimal linear relationship between VCO$_2$A and PA$_{CO_2}$.

**Real-world adjustments**

To simulate real-world measurements, error was introduced into each measurement (P$_A$CO$_2$, Vt-in, Vt-out, VCO$_2$M, Table 1), consistent with manufacturer specifications [14] and calculation of PBF was repeated.

**PBF comparisons**

PBF was calculated from equation 3 using two techniques: first, using known VCO$_2$A measurements directly from the model (PBF-preset), calculated using actual diffusion of CO$_2$ across the alveolar membrane, and second, using VCO$_2$M to calculate VCO$_2$A using equation 5 (PBF-calc), which represents the strategy that might be used practically in subjects. In the either case, PBF would be calculated using equation 1 where PvCO$_2$ was attained by extrapolating to the x-intercept of a plot relating VCO$_2$A and P$_A$CO$_2$-ave. We determined the appropriate number of breaths required for accurate measurement of PBF-calc, by increasing the number of breaths used to calculate the average PBF-calc until the repeatability was similar to that of thermodilution [15].

PBF-calc was assessed and evaluated using Bland-Altman analysis where the acceptable error was taken from prior established theory [16].

**Results**

The model for tidal ventilation produces reasonable values for P$_A$CO$_2$ over 10 min of breathing. Sample oscillations in P$_A$CO$_2$ and lung volume can be seen in Figure 3.

VCO$_2$A tended to vary less over the course of 10 min of breathing than VCO$_2$M, as can be seen in Figure 4. When VCO$_2$A was plotted against P$_A$CO$_2$-ave a linear relationship was revealed which was stronger than with P$_{ET}$CO$_2$ (R = 0.83 versus 0.74, Fig. 5).

Iterating FRC demonstrated that small deviations from the model value produced poorer relationships between VCO$_2$A and P$_A$CO$_2$ (Fig. 6). However, the optimal FRC achieved after iterating FRC was 3.25 L, which was 0.25 L (8.3%) greater than the actual FRC used in the model. Non-optimal FRCs did not significantly reduce the accuracy of the calculated PBF, but did increase its variability.

When P$_A$CO$_2$-ave was used instead of P$_{ET}$CO$_2$, the relationship was improved (R$^2$ = 0.83 versus 0.74). PBF-calc was calculated using 10 breaths at a time (see repeatability below) by extrapolating the line in Figure 3 to the x-intercept (PvCO$_2$) and applied into equation 1. Using this method, the difference between PBF-preset and PBF-calc was 0.62 ± 0.53 L/min.

Figure 7 shows that the repeatability coefficient fell from 1.13 to 0.09 L/min as the number of breaths used to calculate PBF-calc increased from 2 to 30. The repeatability coefficient decreased and leveled at 0.09 when approximately 10 breaths were used to average calculation for PBF-calc. The repeatability coefficient for thermodilution is likely between 0.4 and 0.6 L/min [4, 15], which is the equivalent of using 5 - 6 breaths to average PBF-calc measurements in the proposed method.
Discussion

The most common method for cardiac output measurement involves indicator dilution, normally dyes and temperature. A dye, or cold saline, is typically injected via an invasive catheter into the pulmonary artery and the temporal profile of the concentration of the dye or the temperature of the blood is measured downstream. The profile of the indicator change over time is used to calculate the cardiac output. Over the last three decades, there has been a steady improvement in the technology required to manufacture the catheters and thermistors, and to analyze the indicator curves and calculate the cardiac output [17]. In addition, the expertise to place the catheters and look after catheterized patients has become widespread.

Nevertheless, there are at least three drawbacks necessarily associated with these methods. First, pulmonary artery catheters are inherently invasive and have associated complications including damage to the carotid artery, subclavian artery and lung, air emboli, pneumothorax, malignant arrhythmias and heart block, rupture of right atrium, right ventricle and/or pulmonary capillary, local infection and sepsicaemia, and more [3]. Second, they have many associated costs as a result of the requirement for critical care areas, equipment and personnel. Third, their accuracy is questionable, and can be less reliable and helpful as required for management of critically ill patients [18]. Despite these drawbacks, thermodilution methods continue to be widely used as less invasive alternatives are not sufficiently accurate, not sufficiently robust, or too cumbersome to perform in a large variety of clinical settings [19].

In this study, a new non-invasive method of measuring PBF is introduced based on principles that are already accepted in medicine. The Fick method for measuring PBF is well established and is considered one of the most accurate techniques available. Until recently, however, the Fick technique could only be performed using blood samples despite numerous attempts to measure \( P_{V CO_2} \) non-invasively. The method of creating a step-change in \( V_A \) is the only validated non-invasive Fick method of cardiac output measurement without a special maneuver required by the patient, but can only be used on patients ventilated by a mechanical ventilator, uses only two breaths for measurement ETC. If a spontaneously breathing patient is made to rebreathe previously exhaled gas, the minute ventilation will tend to increase in order to increase the volume of air entering the lungs for gas exchange. The method presented here would require no maneuver on the part of the subject, and no foreign or compressed gases.

The method outlined here describes an original relationship between \( VCO_2 \) and \( PA_{CO_2} \). Its principles are grounded in basic physiology, and its application may extend beyond that of PBF measurement. Since \( PA_{CO_2} \) relates directly to \( VCO_2 \) given a specific set of conditions, other variables that may influence these parameters such as alveolar dead space, may be measurable as well.

There are several practical limitations of this method. First, this strategy for measuring PBF could not be achieved without a perfect, air-tight seal around the mouth, nose, or face. Any air lost would reduce the accuracy of \( V_{t-in}, V_{t-out}, \text{ or } VCO_2_{M} \). This may be inconvenient or impossible for some patients, depending on the presence of facial hair, anomalous anatomy, or trauma.
As few as two breaths can be used to calculate PBF but its reliability is increased as more breaths are incorporated into the calculation. Precision reaches a plateau at about 8 - 10 breaths.

This model assumes that all exhaled gas had participated in exchange of CO₂ with the blood. P_{A}CO_2 may vary depending on differences in ventilation-to-perfusion matching throughout the lung [2, 20]. If some exhaled gas originated from alveolar dead space, P_{ET}CO_2 would underestimate P_{A}CO_2 and as a result, PBF-calc would overestimate PBF. The significance of this was not quantified in this study, but theoretically, it is possible that iterations of alveolar dead space estimates can be coupled with the estimates of FRC to provide an optimal relationship between P_{A}CO_2 and VCO_2A. Still, deviations of P_{ET}CO_2 from P_{A}CO_2 may also be due to incomplete breaths during exhalation and not alveolar dead space per se. This also impacts the calculation for P_{A}CO_2-ave, which will also be affected by all of the above.

In this study, 10 min of breathing were permitted to achieve initial FRC estimate. At this point, it is unclear why the ideal FRC produced from the iterative process differed from that used in the model. Nevertheless, its impact on PBF-calc was marginal. Furthermore, the purpose of this extended period of breathing was to demonstrate the principle rather than practicality. Follow-up studies in humans will be required to further explore this strategy.

The real-world error in measuring Vt, P_{ET}CO_2, VCO_2 was based on products currently available for purchase. These will vary depending on the manufacturer and may improve in the future, making this method more practical.

The accuracy of this method for measuring PBF was similar to others that have been proposed [15]. The repeatability demonstrated that 10 breaths were needed for optimal accuracy, a time slightly greater than would be required for a complete test of thermodilution (about 30 s) depending on respiratory frequency. However, one advantage of the proposed method in this respect is that measurements for PBF are continuous, and no time is needed for “reset”, as may be needed to washout indicator from the pulmonary artery for thermodilution.

Finally, whereas the accuracy of other respiratory-based non-invasive methods may be diminished by respiratory fluctuations, the proposed method is enhanced by large changes in tidal volume and respiratory frequency. It may even be suggested that a subject ought to be encouraged to take deep breaths, or sigh to exaggerate the variability of quiet tidal breathing.

**Conclusion**

The study we describe here is safe, theoretically sound, and demonstrates acceptable accuracy and repeatability. It represents a potential advancement towards measurement of important cardiopulmonary parameters that may be clinically important in the management of both outpatients and those in-hospital. Further studies in humans are required to quantify and evaluate its strengths and limitations.

**Acknowledgement**

None.

**Conflict of Interest**

The authors report no conflicts of interest.

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