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# Exposure assessment by physiologic sampling pump—prediction of minute ventilation using a portable respiratory inductive plethysmograph system

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A study was conducted to evaluate a portable respiratory inductive plethysmograph (RIP) as a means to estimate minute ventilation ($\dot{V}_E$) for use in controlling the flow rate of a physiologic sampling pump (PSP). Specific aims were to: (1) evaluate the ability of the portable RIP system to measure $\dot{V}_E$ using a direct (individual) fixed-volume calibration method (Direct RIP model), (2) develop and evaluate the performance of indirect (group) regression models for $\dot{V}_E$ prediction using output data from the portable RIP and subject demographic characteristics (Indirect RIP model), and (3) compare $\dot{V}_E$ estimates from indirect and direct portable RIP calibration with indirect estimation models published previously. Nine subjects (19–44 years) were divided into calibration ($n = 6$) and test ($n = 3$) datasets and performed step-tests on three different days while wearing the portable RIP and breathing through a pneumotachometer (reference). Minute ventilation and portable RIP output including heart rate, breathing rate, and a motion index were recorded simultaneously during the 80 minute sessions. Calibration data were used to develop a regression model for $\dot{V}_E$ prediction that was subsequently applied to the test dataset. Direct calibration of the portable RIP system produced highly variable estimates of $\dot{V}_E$ ($R^2 = 0.62$, average % error $= 15 \pm 50$) while Indirect RIP model results were highly correlated with the reference ($R^2 = 0.80$–$0.88$) and estimates of total volume were within 10% of reference values on average. Although developed from a limited dataset, the Indirect RIP model provided an alternative approach to estimation of $\dot{V}_E$ and total volume with accuracy comparable to previously published models.

## Introduction

Workers inhale gas-phase chemical mixtures from solvents, cleaners, building materials, process chemicals, and manufacturing operations in the workplace atmosphere. The most common sampling technique to assess workers' exposures to these gases and vapors is personal monitoring, which entails collection of an integrated air sample using a traditional sampling pump (TSP) running at a fixed flow rate for the entire sampling period to estimate a time-weighted average (TWA) concentration. However, concerns have been raised that operating a sampling pump at a constant flow rate may reduce the accuracy of exposure assessment for scenarios in which workload and ventilation rates vary significantly.[1–5] Variation in ventilation levels, breathing patterns, and activity burden may directly influence the extent to which an inhaled pollutant is absorbed and distributed *via* the bloodstream.[6] As a result, the dose of inhaled gases or vapors may potentially be increased, thereby causing more damage in the airways, lungs and other target organs.[7,8] Further, if contaminant concentrations and ventilation rate are correlated, dose can be dramatically affected resulting in a corresponding underestimation of exposure when traditional sampling methods are used.[1,5]

Several investigators have pursued the development of devices generally referred to as physiologic sampling pumps (PSP) in order to account for changing ventilation rates, thereby improving estimates of exposure and dose.[1–5] These PSPs typically employ electronic control features that enable the pump sampling flow rate to be adjusted to correlate with the worker's ventilation rate. Kucharski[2] developed a dust sampler wherein pump rate varied in proportion to the pulmonary ventilation rate, using measured heart rate (HR) as a surrogate for minute ventilation ($\dot{V}_E$). Although the results showed that exposure data collected by the sampler displayed no difference from reference data collected by a respirator filter, this finding may have been influenced by size-selective sampling bias resulting from the fluctuating flow rate of the PSP device and its application to sampling dust. Other attempts have been made to construct $\dot{V}_E$ prediction models for use in regulating a PSP. Satoh *et al.*[4] developed a $\dot{V}_E$ prediction model using HR, weight, height, and age as variables, yielding predictions within 30% of the true values. The model was to be used to control a PSP; however, the performance of the resulting device was never presented. Levine[3] described PSP control circuitry based on thoracic impedance which employs low voltage, high frequency (30–600 kHz) alternating current between two surface electrodes to record thoracic movements or volume changes at the rib cage during

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a respiratory cycle.[9] It was noted that individual variability in ventilation rates may be better modeled using additional parameters such as body fat, thoracic diameter, age, weight, and gender, and further, the need for a lightweight portable pneumotachometer to establish individual calibration equations for relating $\dot{V}_E$ to subject characteristics was identified.

More recently, Hart[1] and Yost[5] described development of a PSP that controls flow rate using measured HR with two different approaches to calibration: (1) an "Indirect HR Method" (group calibration), which was developed using regression on ventilation data from 110 subjects, and was then subsequently applied to new subjects wearing the PSP and (2) a "Direct HR Method" (individual calibration) in which each subject that wears the PSP would undergo a calibration process to establish the relationship between HR and $\dot{V}_E$ for varying levels of exertion. Hart also explored the use of other methods for deriving a real time estimate of a subject's ventilation rate to be used for pump control including respiratory inductive plethysmography (RIP) which is a noninvasive technique for characterizing ventilation. In comparing the different methods for prediction of $\dot{V}_E$, Hart concluded that RIP performed best, but was not appropriate for field use based on the complicated calibration required for each individual and the inadequate packaging of system components. Neither the indirect or direct calibration methods yielded satisfactory results for $\dot{V}_E$ estimation; however, it was noted that the inherent simplicity of the indirect method warranted further investigation.

The relative ease of measuring HR in the field, and its correlation with ventilation parameters has made it an attractive surrogate for $\dot{V}_E$ in many studies including those related to development of PSPs.[8,10] Several investigators have examined relationships between HR, $\dot{V}_E$, and other ventilation parameters at various workloads.[8,11–17] Typical $\dot{V}_E$ and HR responses at various work loads are summarized in Table 1.[18,19] Results generally show a strong correlation between HR and $\dot{V}_E$; however, linear, curvilinear, or more complex relationships with two distinct linear regions have been proposed in the literature.[8,17,20] Further, some experimental results have indicated that HR responds faster to changes in work load than does $\dot{V}_E$.[11,18] These results suggest that additional predictive parameters may be needed to improve the accuracy of equations used to relate HR to $\dot{V}_E$.

In a well-controlled environment, accurate characterization of an individual's respiratory patterns can be accomplished using a spirometer or pneumotachograph. This usually requires that a subject wear a nose clip and then breathe *via* a mouthpiece or face mask connected to the spirometer or pneumotachograph. Minute ventilation can be directly obtained as well as the cumulative volume of air exchanged over the duration of an experiment. However, these methods are difficult to apply in field studies due to the size and complexity of the instrumentation and the invasive nature of the masks or mouthpieces required. As a result, alternative (noninvasive) methods for measuring ventilation patterns have been proposed and evaluated. In particular, RIP has gained acceptance since its introduction more than 20 years ago.[21] This method utilizes two elastic bands circumscribing an individual's rib cage and abdomen to provide electrical signals that vary with the change of the cross-sectional area during breathing. The accuracy of RIP depends on lung volume calibration at the time of study, and numerous calibration methods have been proposed.[22] All require a determination of a volume/motion coefficient for the rib cage and abdomen band signals and several also require simultaneously recording changes in lung volume during the period of calibration *via* a spirometer or pneumotachograph.[23,24] It was the complex nature of the RIP calibration process and shortcomings of the packaging of system components that led Hart[1] to conclude that the approach was not feasible for use with a PSP. Further, in several studies comparing RIP results to reference measurement methods for minute ventilation and cumulative volume during cycling or treadmill exercise, agreement between methods was highly variable.[25,26] Potential problems such as slippage of the sensors on the elastic bands, and complicated and troublesome calibration procedures for the plethysmograph were cited as possible sources of error.

Recently, a portable RIP system known as the LifeShirt® (VivoMetrics, Inc., Ventura, CA) became commercially available.[27] The system consists of an elastic sleeveless garment (Fig. 1a) with embedded RIP sensors and a miniaturized, battery-powered recording unit that can simultaneously monitor several physiological signals in real-time. In addition to recording respiratory and ECG waveforms, the recording unit is capable of transmitting six physiological parameters at one second intervals *via* a serial port: instantaneous and average breathing rate, instantaneous and average heart rate, and motion and position indices. The garment is intended to minimize the

**Table 1** Typical heart rate, minute ventilation, and work load for various activities

| Activity | Heart rate[a]/beats min⁻¹ | Minute volume[a]/l min⁻¹ | Work load[b]/W |
|---|---|---|---|
| Resting | 60–70 | 6–7 | |
| Light work | 75–100 | 11–20 | ≤50 |
| Moderate work | 100–125 | 20–31 | 50–80 |
| Heavy work | 125–150 | 31–43 | 80–120 |
| Very heavy work | 150–175 | 43–56 | 120–170 |

[a] Wallaart, 1997.[19] [b] Work load obtained from Fig. 11-4 (Åstrand and Rodahl, 1986).[18]

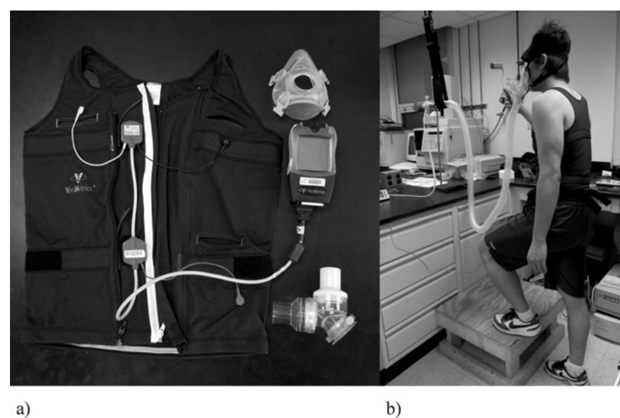

**Fig. 1** Calibration protocol showing (a) portable RIP system components (garment, ECG leads, recorder), pneumotachograph mask, and valve; and (b) step-test session with pneumotachograph configuration.

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potential for sensor displacement during exercise[27] and the stored raw waveforms can be post-processed using software that performs calibration, parameter extraction including tidal volume and minute ventilation, and trend analysis. The system offers different calibration options including a routine for semi-automated fixed-volume calibration which facilitates direct (individual) calibration using an 800 ml disposable bag. The protocol requires that each subject breathe into the fixed-volume bag several times in both sitting and standing postures. Software then determines the multiplication factors for the outputs of rib cage and abdominal bands of the plethysmograph using least-squares regression analysis.[27,28] In previous studies in which the plethysmograph was first calibrated against a pneumotachograph, the system provided accurate estimates of minute ventilation during treadmill exercise.[29,30]

The purpose of this project was to evaluate a portable RIP system (LifeShirt system) as a means to estimate $\dot{V}_E$ for use in controlling the flow rate of a PSP, and to compare results to those of alternative approaches. The shortcomings of previous RIP systems included possible band slippage during exercise, complicated calibration protocols, and packaging that was not conducive to field use, each of which may have been addressed to some extent with the new system. Earlier approaches to controlling PSP flow rate have relied on indirect (group) and direct (individual) modeling of $\dot{V}_E$ based on HR and other descriptive parameters, or direct (individual) estimation of $\dot{V}_E$ using a calibrated plethysmograph. Therefore, the specific aims of this project were to (1) evaluate the ability of the portable RIP system to predict minute ventilation when using a direct calibration method—the fixed volume least-squares approach employed in the instrument protocol (Direct RIP model); (2) develop and evaluate the performance of indirect (group) regression models for predicting $\dot{V}_E$ based on output data available from the portable plethysmograph and subject demographic characteristics (Indirect RIP model); and (3) compare the minute ventilation estimates of indirect and direct models derived from the portable plethysmograph with the indirect models developed previously by Satoh *et al.*[4] and Hart:[1]

Satoh model

$$\log_{10} \dot{V}_E = (9.38(\mathrm{HR} - \mathrm{HR}_{rest}) + 4.22H + 1.19W + 2.22A + \mathrm{HR}_{rest}) \times 10^{-3} - 0.0439$$

Hart model

$$\ln \dot{V}_E = 0.01894(\mathrm{HR} - \mathrm{HR}_{rest}) + 0.01052W + 1.9008$$

where, $\dot{V}_E$ = minute ventilation (l min$^{-1}$), HR = heart rate (beats min$^{-1}$), $\mathrm{HR}_{rest}$ = resting heart rate (beats min$^{-1}$), $H$ = height (cm), $W$ = weight (kg), and $A$ = age (years).

## Methods

### Subjects

Research was conducted according to the procedures of the Pennsylvania State University Office for Research Protections (ORP). Nine healthy subjects (three female, six male) ranging in age from 19 to 44 years were recruited to participate in the study. All potential participants were screened using a medical questionnaire to ensure that they were capable of performing a stepping exercise at moderate intensity. Those with a medical history of cardiovascular or respiratory disease, recent surgeries limiting physical activities, back, knee, or ankle pain, limb-related injuries over the past year, or sensitivity to spandex materials were excluded. According to the chronological order of enrollment, subjects were divided into a model calibration dataset consisting of six subjects, and a model test dataset consisting of three subjects. Data obtained from the calibration dataset were used to develop $\dot{V}_E$ prediction models, which were then evaluated using the test dataset. Following the development and evaluation of $\dot{V}_E$ prediction models, the datasets were combined and used as the basis for a revised regression model for $\dot{V}_E$. A summary of subject characteristics for the calibration, test, and combined datasets is presented in Table 2.

### Experimental protocol

Subjects performed step tests (Fig. 1b) on three different days while wearing the portable plethysmograph and breathing through a pneumotachometer. Subjects were fitted with an appropriately sized Lifeshirt® garment (Vivometrics, Inc., Ventura, CA) and a face mask equipped with a two-way non re-breathing valve (7900 Series, Hans Rudolph, Inc., Shawnee, KS). Prior to beginning a test, subjects sat quietly for approximately 5 min and resting heart rate was recorded. Calibration dataset subjects then underwent the fixed-volume calibration procedure as specified by the portable RIP manufacturer. This process entails breathing in and out of an 800 ml calibration bag (P/N 910-0185-000, Vivometrics, Ventura, CA) eight times, completely filling and emptying the bag with each breath. The process consists of six sets of these maneuvers in both sitting and standing postures. The resulting RIP waveforms are recorded as part of the experimental run, at the completion of which data from the LifeShirt® recorder are uploaded into a software application (Vivologic Version 2.9, Vivometrics, Inc., Ventura, CA) for post-processing. The software includes a fixed-volume least squares calibration subroutine which assists the user in identifying the location of the 800 ml calibration breaths within the experimental run, and then automatically adjusts the plethysmograph rib cage and abdominal band coefficients for the individual using a least-squares algorithm and the known tidal volume of the breaths (800 ml). The resulting relationship can then be applied to the RIP waveform for the entire run, allowing tidal volumes and minute ventilation to be estimated.

**Table 2** Subject demographic characteristics[a]

| Variable | Calibration set | Test set | Combined set |
|---|---|---|---|
| $N$ | 6 | 3 | 9 |
| Gender (female/male) | 2/4 | 1/2 | 3/6 |
| Age/years | 29.7 ± 10.8 | 22.6 ± 2.1 | 27.3 ± 9.3 |
| Height/in | 67.8 ± 5.0 | 68 ± 2.6 | 67.9 ± 4.2 |
| Weight/lb | 165.7 ± 39.8 | 144 ± 28.8 | 158.4 ± 36.2 |
| BMI/kg m$^{-2}$ | 25.0 ± 3.3 | 21.8 ± 3.2 | 23.9 ± 3.5 |
| Resting HR/beats min$^{-1}$ | 75.8 ± 7.1 | 66.8 ± 5.4 | 72.7 ± 7.8 |

[a] Values are means ± SD, BMI = body mass index.

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Following RIP calibration, subjects donned the pneumotachometer mask and the experimental run began. A step-test session consisted of four 10 min exercise stages with stepping frequencies increasing incrementally from 5 steps $\text{min}^{-1}$ to 20 steps $\text{min}^{-1}$ using a platform 32.5 cm in height (Fig. 1b). Subjects were asked to simulate a painting task by holding a paint roller in one hand and raising it above shoulder height with each step to introduce typical work-related upper body movement into the protocol. The resulting physical workload varied according to the weight of the subject and the stepping rate, ranging on average from approximately 20 W (5 steps $\text{min}^{-1}$) to 80 W (20 steps $\text{min}^{-1}$). A 10 min resting period was provided between each exertion. Minute ventilation was recorded using a calibrated pneumotachograph, and portable RIP system output was recorded simultaneously over the course of the 80 minute session. The portable RIP recorder unit is equipped with a serial port which transmits physiological parameters including heart rate, breathing rate, motion and position indices at one second intervals. At the same time, the recorder logs the output from the ECG electrodes, the plethysmograph, and an embedded accelerometer, at frequencies of 200 Hz, 50 Hz, and 10 Hz, respectively. It is these more detailed waveforms that are stored on a compact flash memory card within the recorder, and later transferred to a PC for post-processing, while the less detailed summary parameters are produced at one second intervals in real time and transmitted over the serial port.

The pneumotachograph (PT) consisted of a respiratory flow head (MLT300L, ADInstruments, Inc., Colorado Springs, CO) connected by a 182 cm length of 35 mm id clean bore tubing (Vacumed, Ventura, CA) to a two-way non re-breathing valve and face mask. The pressure ports of the flow head were connected to a calibrated differential pressure transducer (PX163-2.5BDV5V, Omega Engineering, Inc., Stamford, CT), the output of which was recorded using a USB data acquisition module (DT9802, Data Translation, Inc., Marlboro, MA) controlled by custom software (Agilent VEE One Lab, Agilent Technologies, Santa Clara, CA) for the storage and display of inhalation parameters including instantaneous flow rates and integrated cumulative volumes. The PT was initially calibrated by drawing dry room temperature air (~24 °C) through the flow head and recording the resulting pressure drop for flow rates ranging from 0.1 l $\text{min}^{-1}$–188 l $\text{min}^{-1}$ (determined by reference mass flow meter, Model 4043, TSI, Inc., Shoreview, MN). The resulting linear pressure-flow response curve ($R^2 = 0.998$) was subsequently used to yield instantaneous flow rates and integrated inhalation volume. PT performance was confirmed to be within ±5% of expected values before and after every run using a 3 l spirometer calibration syringe (Series 5530, Hans Rudolph, Inc., Shawnee, KS). Output from the portable plethysmograph was acquired *via* a serial connection and was also displayed and recorded using the data acquisition software package.

At the conclusion of each run, a single data file including the outputs from the pneumotachometer and portable plethysmograph was produced by the data acquisition program for use in subsequent analyses. The comma-delimited text file included instantaneous flow rate (l $\text{min}^{-1}$) and cumulative volume (l) from the pneumotachometer, instantaneous and average heart rate (beats $\text{min}^{-1}$), instantaneous and average breathing rates (breaths $\text{min}^{-1}$), a motion index (arbitrary scale of 0–255), and an encoded position parameter (0–146 = down; >146 = up) from the portable RIP serial output, all recorded at one second intervals. The resulting data combined with the demographic characteristics of the subjects then formed the basis of regression models developed to estimate $\dot{V}_E$.

### Data analysis

Date were analysed using Minitab (Release 14.2, Minitab, Inc., State College, PA). Best subsets regression was first used to identify combinations of predictor variables likely to be most useful in modeling $\dot{V}_E$ based on the available demographic characteristics of subjects and portable plethysmograph output parameters from the calibration dataset. The resulting preliminary regression model was then applied to the test dataset and performance was evaluated by examining the agreement of predicted values for $\dot{V}_E$ *versus* actual values. Summary statistics including the coefficient of determination ($R^2$) and average percent errors were used to assess overall performance, and Bland–Altman plots were prepared to more closely examine the performance of $\dot{V}_E$ models for individual subjects.[31] Regression models for $\dot{V}_E$ prediction were further evaluated by comparing the resulting estimates of cumulative volume to the actual values recorded during runs—a prediction model for $\dot{V}_E$ can be highly correlated with actual rates, but still yield significant error in estimates of total cumulative volume due to biases in the estimate of flow rate.

A mixed model ANOVA was used to determine whether significant differences existed between cumulative volumes estimated using the various $\dot{V}_E$ prediction methods examined in the study: Direct RIP model, Indirect RIP model, and indirect equations described by Satoh *et al.*[4] and Hart.[1] Two factors were used in the ANOVA model: the model used to estimate cumulative volume was a fixed factor and subject was a random factor. Tukey's method (95% simultaneous confidence intervals) was then used to perform multiple comparisons of the cumulative volume estimates, and to identify groupings that differed significantly from the reference standard (pneumotachometer).

## Results and discussion

### Application of models to calibration dataset

Regression models based upon the calibration dataset were developed first. In selecting candidate models, it was desired to minimize the number of parameters required since the processing power and storage space of the planned PSP was likely to be limited. Further, the generalizability and physiological plausibility of the resulting regression models was a primary concern. While stepwise regression modeling of $\dot{V}_E$ can yield higher values of $R^2$, the complexity of the resulting models and the lack of plausible explanations for many coefficients likely limit the generalizability of such expressions to new populations. In addition, since many of the parameters considered for inclusion are highly collinear (*e.g.*, breathing rate and heart rate, height and weight), it was necessary to consider variance inflation factors in order to eliminate highly correlated prediction parameters from the model. The resulting Indirect RIP model is given as:

$$\dot{V}_E = 7.90 + 0.392IH + 19.9IG + 0.000751\text{MI}W$$

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where, $I$ = intensity = [(HR − $HR_{rest}$) / ($HR_{max}$ − $HR_{rest}$)] = (ΔHR / HRR), HR = heart rate from the RIP system, $HR_{rest}$ = resting heart rate, $HR_{max}$ = 205.8 – (0.685 age), HRR = heart rate reserve, $H$ = height (inches), $G$ = gender (male = 1, female = 0), MI = motion index which is a composite measure of movement provided by the portable RIP accelerometer output and ranging in value from 1–255, and W = weight (pounds).

A measure of exercise intensity ($I$) is commonly employed in calculating target heart rates for exercise regimens and is usually derived from the elevation of heart rate (HR) above resting heart rate ($HR_{rest}$), relative to the estimated heart rate reserve (HRR).[32] $HR_{max}$ can be measured in a laboratory setting or estimated using an appropriate correlation. For this model, $HR_{max}$ was estimated as $HR_{max}$ = 205.8 – (0.685 age).[33] Height and weight were also found to be significant predictors as expected based on the previous models of Hart[1] and Satoh *et al.*[4]

The resulting regression equation is referred to as an Indirect RIP model since the approach to estimating $\dot{V}_E$ for a subsequent population relies on a group or indirect calibration. The model employs demographic information from the subjects and summary physiologic parameters output by the portable RIP at one second intervals. The performance of the Indirect RIP model when applied to the calibration dataset is summarized in Fig. 2a and Table 3. There is strong correlation between predicted and reference values of $\dot{V}_E$ with an $R^2$ value of 0.88 (Fig. 2a). Representative time-series plots of predicted and reference $\dot{V}_E$ are presented in Fig. 3. The plots generally show good agreement between modeled and actual results, but also demonstrate the type of deviations that can produce errors in the estimate of cumulative volume despite a strong correlation for instantaneous flow rate. However, as shown in Table 3, agreement between Indirect RIP model total volume and reference total volume is very good with an average error of 1.2%. It should be noted that this performance is based on application of the model to the same dataset used to calibrate—in this comparison the model is not being applied indirectly.

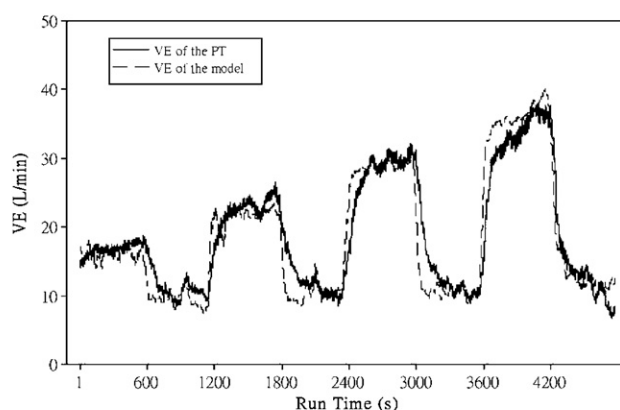


**Fig. 3** Representative time-series plot showing predicted (dashed line) *versus* reference (solid line) $\dot{V}_E$ (subject 4, run 2).

Comparison of $\dot{V}_E$ estimates resulting from Direct RIP calibration and actual values is depicted graphically in Fig. 2b, and numerically in Table 3. Results were highly variable—the value of $R^2$ was 0.62, suggesting considerable scatter in the estimates relative to the reference standard, which was further confirmed by a plot (Fig. 2b). It was anticipated that portable RIP estimates of $\dot{V}_E$ resulting from direct calibration for each subject would yield the best results based on previous reports[29,30] and given the characteristics of the current portable RIP system, which were

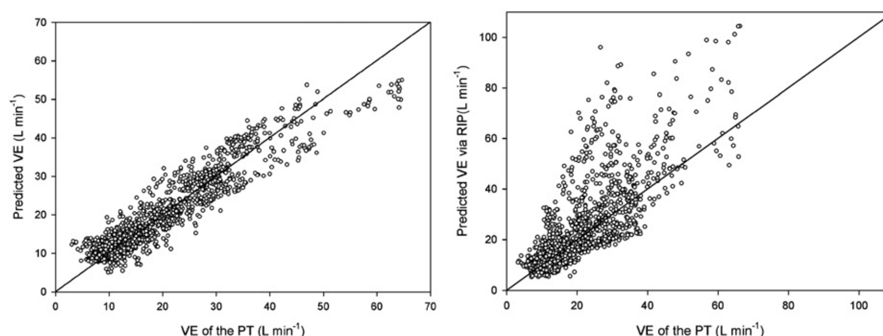


**Fig. 2** Comparison of predicted and actual $\dot{V}_E$ for the calibration dataset using (a) Indirect RIP model, and (b) Direct RIP method. (Note: every 60th data point plotted for clarity, solid line—line of identity.)

**Table 3** Comparison of results for different $\dot{V}_E$ prediction models applied to the calibration dataset (6 subjects × 3 trials = 18 runs)[a]

| | Minute ventilation/l min$^{-1}$ | | | Total volume/l | | |
|---|---|---|---|---|---|---|
| Model | Mean | $R^2$ | Average % error | Mean | Average % error | Tukey grouping |
| Reference (PT) | 19.7 ±11.0 | — | — | 1570 ± 309 | — | A |
| Indirect RIP | 19.7 ± 10.3 | 0.88 | 4.2 ± 27 | 1570 ± 256 | 1.2 ± 12 | A |
| Direct RIP | 23.2 ± 17.1 | 0.62 | 15 ± 50 | 1820 ± 638 | 22 ± 50 | A |
| Satoh | 16.1 ± 9.05 | 0.81 | −13 ± 29 | 1280 ± 327 | −18 ± 17 | B |
| Hart | 26.2 ± 12.7 | 0.81 | 46 ± 51 | 2090 ± 467 | 35 ± 27 | C |

[a] Average % error = 100[(predicted value − reference value)/(reference value)].

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designed specifically to address noted problems such as complex calibration and band slippage. However, it was found that Direct RIP results could be relatively accurate for a given subject on one day, and then significantly less accurate on another day, despite the fact that the fixed-volume calibration protocol was performed in a similar systematic manner as specified by the manufacturer. The problem may be related to the fixed-volume calibration routine, since in previous studies which showed excellent agreement between the LifeShirt® system and reference volume measurements (pneumotachometer), the plethysmograph was first calibrated against the reference.[29,30] However, the current results are consistent with those described previously by Stark *et al.*[26] for other RIP systems and may reflect inherent variability in the performance of RIP-based systems. While it may be possible to achieve better results with the portable plethysmograph calibrated against the pneumotachometer, this defeats the desired goal of providing a simplified calibration process that is conducive to direct calibration in the field for use with a PSP. For this reason, Direct RIP calibration was not pursued and further efforts were focused on developing Indirect RIP models employing the physiological summary data provided in real-time by the portable RIP recording unit. Although it might be possible to obtain similar information from commercially available heart rate monitors and three-axis accelerometers, it was decided that since the LifeShirt system was already available and packaged in a manner convenient for field use, subsequent evaluation of the PSP would continue to rely on the use of the portable RIP system for generation of heart rate and motion data.

Performance of the Hart and Satoh indirect $\dot{V}_E$ prediction models was examined for the calibration dataset and results are presented in Table 3. Values of $R^2$ were similar for both (0.81) and lower than that of the Indirect RIP model (0.88); however, it is again important to note that the Hart and Satoh models are being applied indirectly (to new subjects), while the Indirect RIP model was applied to its own calibration dataset. Total volumes were also compared to the reference and it was found that the Satoh model had a negative bias of approximately 20% while the Hart model was positively biased by 35%. While the source of bias cannot be determined, it was noted that the relationship between the two models is consistent with the findings of Hart—the Satoh model had a significant negative bias relative to the Hart measures of volume and the resulting indirect model.[1,5] The results of a group comparison of the various methods for estimating total volume are also presented in Table 3. The Tukey test shows that significant differences were seen for the total volume estimates resulting from the methods: the Hart model yielded estimates significantly higher than the other approaches, while the Satoh method was significantly lower. Total volumes for the Direct and Indirect RIP models were not significantly different from the reference method, but were different from the Satoh and Hart models.

### Application of models to test dataset

In order to evaluate the performance of the Indirect RIP model, the equation was used to estimate $\dot{V}_E$ for a test dataset of three new subjects. The results of this comparison are presented in Table 4 along with similar results for the Satoh and Hart models. All three models showed similar correlation with reference measures of $\dot{V}_E$—values of $R^2$ were 0.80, 0.77, and 0.79 for the Indirect RIP, Satoh, and Hart models, respectively. The average percent error for total volume was similar for the Indirect RIP (9.5%) and Satoh (−11%) models, while the Hart model was significantly higher (41%). The relative trend for bias in volume estimates was consistent with that seen for the calibration dataset—the Hart model had a consistently larger positive bias whereas the Satoh model bias was smaller but consistently negative. The Tukey grouping of total volume estimates indicated that although the Indirect RIP and Satoh models were not significantly different from the reference method, all models were significantly different from each other.

A plot of Indirect RIP model predicted $\dot{V}_E$ *versus* reference values is presented in Fig. 4a along with a Bland–Altman diagram (Fig. 4b). Results show relatively good correlation between predicted and actual flow rates, consistent with an $R^2$ of 0.80. The accompanying Bland-Altman diagram shows that model performance varies—the best agreement appears to be for subject 9 (filled triangle) while the largest differences are exhibited for subject 8 (open circle). Referring to the subject demographic characteristics (Table 2) it is seen that the test dataset was not well-matched to the calibration dataset—the test set subjects are generally younger, weigh less, have lower resting HR, and lower BMI. While it would be possible to rearrange the subjects to create more similar calibration and test datasets, it could be argued that this manipulation would invalidate the purpose of the test set since the original design called for the formation of the datasets as subjects were enrolled. In any case, the performance of the Indirect RIP model for the test dataset was relatively good with average error less than 10% overall, and individual errors of −8.6% (17.8), 28% (29.5), and 4.2% (16.5) for subjects 7, 8, and 9, respectively (SD in parentheses). It was noted that the Satoh model performed similarly well for the test set—the Indirect RIP model had slightly better correlation and average percent error. This may be an indication that the

**Table 4** Comparison of results for different $\dot{V}_E$ prediction models applied to the test dataset (3 subjects × 3 trials = 9 runs)[a]

| | Minute ventilation/l min$^{-1}$ | | | Total volume/l | | |
|---|---|---|---|---|---|---|
| Model | Mean | $R^2$ | Average % error | Mean | Average % error | Tukey grouping |
| Reference (PT) | 17.6 ± 9.34 | — | — | 1380 ± 318 | — | A/B |
| Indirect RIP | 19.0 ± 11.3 | 0.80 | 8.0 ± 27 | 1490 ± 345 | 9.5 ± 20 | A |
| Satoh | 15.3 ± 9.37 | 0.77 | −12 ± 20 | 1200 ± 208 | −11 ± 14 | B |
| Hart | 24.4 ± 13.2 | 0.79 | 43 ± 29 | 1910 ± 365 | 41 ± 21 | C |

[a] Average % error = 100[(predicted value − reference value)/(reference value)].

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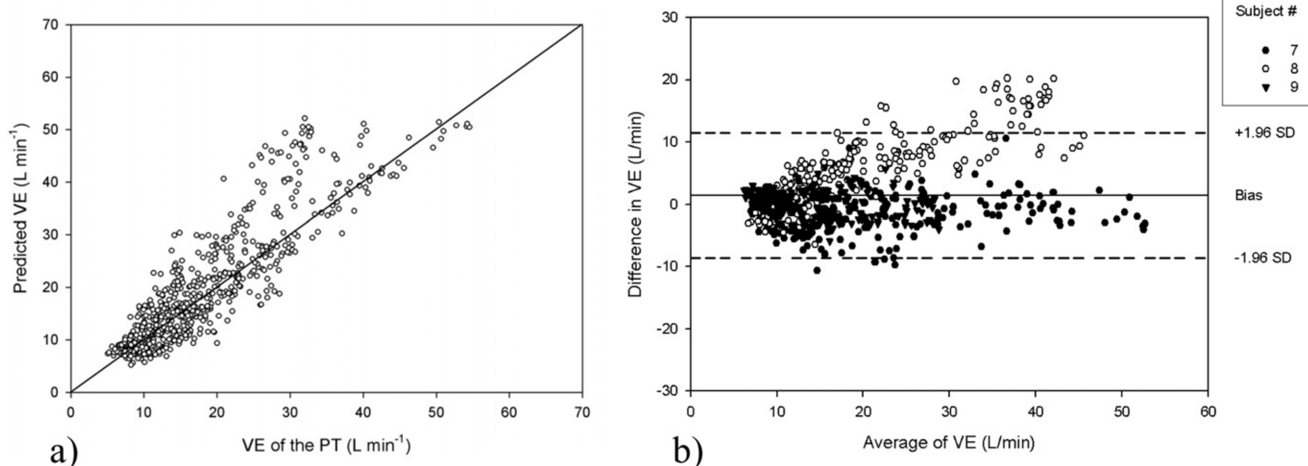


**Fig. 4** plethysmograph model results for the test dataset: (a) comparison of predicted and actual $\dot{V}_E$, and (b) Bland–Altman diagram summarizing results for test-dataset by subject. (Note: every 60th data point plotted for clarity, solid line—line of identity.)

**Table 5** Comparison of results for different $\dot{V}_E$ prediction models applied to the combined dataset (9 subjects × 3 trials = 27 runs)[a]

| | Minute ventilation/l min$^{-1}$ | | | Total volume/l | | |
|---|---|---|---|---|---|---|
| Model | Mean | $R^2$ | Average % error | Mean | Average % error | Tukey grouping |
| Reference (PT) | 19.0 ± 10.5 | — | — | 1510 ± 320 | — | A |
| Indirect RIP | 19.0 ± 9.68 | 0.84 | 4.7 ± 26 | 1510 ± 228 | 2.0 ± 15 | A |
| Satoh | 15.8 ± 9.17 | 0.79 | −13 ± 26 | 1260 ± 291 | −16 ± 16 | B |
| Hart | 25.6 ± 12.9 | 0.80 | 45 ± 45 | 2030 ± 437 | 37 ± 25 | C |

[a] Average % error = 100[(predicted value − reference value)/(reference value)].

demographic characteristics of the test set subjects align well with the Satoh dataset.

### Application of models to combined dataset

In a final examination of model performance, the calibration and test datasets were combined and a revised Indirect RIP model was developed using regression of the pooled dataset with the same prediction variables. The resulting model follows:

$$\dot{V}_E = 8.60 + 0.331IH + 12.9IG + 0.000975\text{MI}W$$

The revised equation described 84% of the variability in $\dot{V}_E$ with an average percent error of 2 ± 15 for the estimated total volume (Table 5). Performance of the Satoh and Hart models was also examined for the combined dataset yielding results similar to those of the calibration and test sets. The Satoh model described 79% of the variability and produced estimates of total volume that were on average 16 percent lower than the reference method (−16 % ± 16) while the Hart model displayed a similar $R^2$ (0.80) and relatively large positive bias (37% ± 25). The Tukey comparison of total volume estimates indicated that there was not a significant difference between the Indirect RIP model and the reference method, whereas both the Satoh and Hart models were significantly different from the reference. In addition, all three models were significantly different from each other. It is again important to note that the revised Indirect RIP model was developed and applied to the same combined dataset for this comparison in order to characterize potential performance. Performance of any model is expected to decrease when applied to subjects with demographic characteristics that differ significantly from those of the calibration dataset. For this reason, it was decided to combine test and calibration datasets in order to develop a revised Indirect RIP model, the goal being to further improve generalizability as a result of the more diverse calibration subject pool. However, an additional test dataset was not available and any potential improvement in model performance could not be confirmed.

## Summary and conclusions

A study was conducted to compare various methods for estimation of $\dot{V}_E$ for the purpose of controlling the sampling flow rate of a PSP. The application of direct and indirect calibration methods to a recently commercialized portable RIP system was compared with $\dot{V}_E$ estimates resulting from previously developed indirect calibration equations—the Hart and Satoh models. Results indicated that despite design features intended to simplify calibration and reduce the likelihood of band slippage, direct calibration of the portable RIP system produced inconsistent and highly variable results for estimation of $\dot{V}_E$. Studies in which the accuracy of the portable RIP system was demonstrated have employed more complex direct calibration protocols requiring

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a laboratory-based pneumotachograph, which defeats the desired goal of establishing a simple field-ready direct calibration method.

Indirect calibration models based on the demographic characteristics of subjects and real time output data available from the portable plethysmograph produced estimates of $\dot{V}_E$ that were highly correlated with the reference method ($R^2 = 0.80$–$0.88$) and yielded estimates of total volume that on average were within 10% of actual values for a test dataset. Although the Indirect RIP model was not shown to be significantly better than the Satoh model, it is likely that with a larger calibration dataset and given the availability of motion index data, the relative performance of the RIP model would improve further. In addition, the packaging of the portable RIP system facilitates the collection and transmission of the heart rate data required for any indirect method through the use of the integrated garment, transducers, and recording unit with serial output port. The resulting unit was found to be an appropriate means for generating the data required for development of indirect calibration models used in conjunction with a PSP.

## Disclaimer

The findings and conclusions in this report are those of the authors and do not necessarily represent the official position of the Centers for Disease Control and Prevention.

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