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Noninvasive Methods for Measuring Ventilation in Mobile Subjects

Measurements of Ventilation in Freely Ranging Subjects

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Assessment of Heart Rate As a Predictor of Ventilation

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Includes the Commentary of the Institute's Health Review Committee

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HEI HEALTH EFFECTS INSTITUTE

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HEI Statement

Synopsis of Research Report Number 59

Noninvasive Methods for Measuring Ventilation in Mobile Subjects

BACKGROUND

Ventilation, a measure of the frequency and depth of breathing, is an important determinant of the amounts of indoor and outdoor air pollutants that enter the respiratory tract. Increasing ventilation (for example, through exercise) increases the amounts, or doses, of inhaled air pollutants delivered to the respiratory tract. Such increases may influence the magnitude of adverse effects experienced by healthy individuals and by individuals particularly susceptible to the effects of air pollutants.

Estimating doses of air pollutants is one facet of risk assessment, a quantitative approach for evaluating human health hazards from pollutants. The amount of a pollutant absorbed by or deposited in an exposed individual over a specific time period is referred to as dose. To determine inhaled dose accurately, data about ventilation, ambient exposure conditions, and exposure duration are needed. Although exposure conditions and duration are relatively easy to determine, there are few ventilatory data for people moving freely at home or at work. This is because standard equipment for measuring ventilation is not readily movable and requires a mouthpiece or face mask. As a result, pollutant doses often have been estimated from exposure conditions and duration alone. The Health Effects Institute sponsored two studies to develop and test methods for measuring ventilation in freely mobile subjects. These noninvasive methods do not require a mouthpiece or face mask, both of which are recognized as contributors to ventilatory measurement errors.

APPROACHES AND RESULTS

Ventilation Estimated from Body Surface Displacement Measurements

Drs. Dennis McCool and Domyung Paek measured ventilation with a body surface displacement (BSD) model. Each subject wore wide elastic bands containing coated wire coils around the chest and abdomen and had special magnets affixed to the breastbone and navel. Changes in electrical signals from these devices indicated dimensional changes in the subject's body that were associated with breathing. After the BSD signals were calibrated with data from a spirometer (standard equipment for measuring breathing parameters), subsequent BSD measurements yielded data about a subject's breathing patterns, breath frequency, and ventilation.

In laboratory studies, the investigators compared BSD data from 10 subjects with spirometric data obtained during upper and lower body work tasks, including lifting, pulling, and cycling. They also examined the influence of a spirometer mouthpiece on ventilation measurements. To evaluate the feasibility of using heart rate to predict ventilation, the investigators first plotted a ventilation–heart rate calibration curve for each subject based on data from a progressive exertion cycling test. They then used this curve to estimate ventilation from heart rate data alone, and compared these ventilation data with ventilation data obtained by BSD and spirometry. Finally, they tested their BSD model in a field study by monitoring nine vocational school students. The BSD equipment was placed on a cart to facilitate mobility of the tethered subjects during a classroom session and an auto body repair workshop session.

The laboratory data demonstrated that the BSD model provided ventilation data comparable to spirometry data. The specific work task influenced statistical correlations between BSD and spirometric data. Rhythmic breathing during cycling correlated the best, whereas erratic breathing during lifting correlated the worst. The investigators verified previous reports that the presence of a mouthpiece increases the volume of air inhaled per breath and decreases breath frequency. They also concluded that heart rate can be an inaccurate predictor of ventilation during the low activity levels that constitute much of daily life; transient ventilation increases detected by BSD or spirometry during low activity were not matched by similar increases in heart rate. Finally, the investigators' field study demonstrated the feasibility of using the BSD model to measure ventilation accurately and noninvasively in mobile subjects.

Ventilation Estimated from Heart Rate

Dr. Jonathan Samet and colleagues wanted to develop methods for estimating ventilation from heart rate for future epidemiologic studies. Their 58 subjects included healthy adults and children, and adults with heart disease, lung disease, or asthma. First, the investigators collected spirometric and heart monitor data in the laboratory to plot ventilation–heart rate curves for each subject during cycling, vacuuming, and lifting. They then used heart monitor data to validate the accuracy of the Heartwatch, a portable, commercial device combining a small transmitter worn on the subject's chest with a wristwatch-style receiver that records heart rate. Because route of breathing affects lung pollutant dose, they also used a partitioned face mask to determine the proportion of oral versus nasal breathing. With increased oral breathing during exercise, some inhaled air bypasses the air-scrubbing mechanisms in the nasal passages and can increase pollutant dose to the lower respiratory tract. Finally, the investigators conducted a field study to estimate ventilation from Heartwatch data using a heart rate–ventilation calibration curve from a progressive exertion cycling test; they then categorized these ventilation data by activity using records maintained by the subjects.

Data from the laboratory studies indicated that ventilation increased faster than heart rate when subjects performed upper body exercise compared with lower body exercise. Because most daily activities do not involve upper body exertion, the investigators concluded that heart rate could be used to estimate ventilation in field studies. Predictably, they reported that most subjects shifted from nasal to oral breathing with increasing exercise intensity. Using Heartwatch data from their field study, the investigators provided ventilation estimates categorized according to subject age, gender, health status, and activities. Dr. Samet and colleagues concluded that heart rate monitoring presents a feasible approach for estimating ventilation in the community setting.

PERSPECTIVES ON THE TWO METHODS

The strength of both field studies was that they revealed problems that might have gone unrecognized had methods testing been restricted to the laboratory. Both groups concurred that, due to considerable intersubject variability, establishing a heart rate–ventilation calibration curve for each subject was essential. They also agreed that the curves used in their studies were based on exercise intensities far exceeding those observed in the field studies; future studies should establish heart rate–ventilation curves based on lower activity ranges.

The two investigator groups disagreed, however, on the utility of heart rate for estimating ventilation. Dr. Samet stated that, with appropriate ventilation–heart rate curves, heart rate could reliably predict ventilation. Dr. McCool concluded that, although averaged heart rate values might correlate with averaged ventilation values, brief increases in ventilation did not always correspond with increases in heart rate. This mismatch could introduce errors into ventilation estimates based on heart rate alone. Future studies are needed to address this problem.

In summary, these two studies provide promising methods for measuring ventilation noninvasively. Because the BSD equipment imposes some mobility restrictions, Dr. McCool's approach may find ready application in small occupational studies. Dr. Samet's methods appear suitable for larger epidemiological studies if a heart rate–ventilation calibration curve is first established for each subject. These studies contribute toward improvements in methods for estimating inhaled pollutant dose and advance risk assessment methodology.

This Statement is a summary, prepared by the Health Effects Institute (HEI) and approved by the Board of Directors, of two research projects sponsored by HEI from 1989 to 1991. Drs. Dennis McCool and Domyung Paek from the Memorial Hospital of Rhode Island, in Pawtucket, RI, conducted the first study, Measurements of Ventilation in Freely Ranging Subjects, and Dr. Jonathan M. Samet and his colleagues from the University of New Mexico, in Albuquerque, NM, conducted the second study, Assessment of Heart Rate as a Predictor of Ventilation. The following Research Report contains both the detailed Investigators' Reports and a Commentary on the studies prepared by the Institute's Health Review Committee.

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The Commentary on the Investigators' Report is prepared by the HEI Health Review Committee and staff. Its purpose is to place the study into a broader scientific context, to point out its strengths and limitations, and to discuss the remaining uncertainties and the implications of the findings for public health.

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Measurements of Ventilation in Freely Ranging Subjects

F. Dennis McCool and Domyung Paek

ABSTRACT

Both the level of ventilation and breathing pattern (breathing frequency, inspiratory time, and tidal volume) have an important influence on particle deposition and gas uptake in the lungs. Accordingly, a description of these measures is needed to assess better the dose of particulate deposit and gas uptake in the lungs during varied activities. The long-term objectives of this study were to develop a means of measuring minute ventilation in the field by using body surface displacements, and to evaluate the utility of heart rate as an index of minute ventilation.

By using respiratory inductance plethysmographic belts and magnetometers placed on the rib cage and abdomen, ventilation and breathing pattern can be noninvasively measured in mobile individuals. Our specific aims were (1) to validate measurements of ventilation using body surface displacement; (2) to describe breathing patterns in subjects performing a variety of daily activities in the laboratory; (3) to analyze relationships between minute ventilation and heart rate; and (4) to measure ventilation in the field with one technique utilizing body surface displacements and another based upon heart rate.

We found that values of tidal volume, inspiratory time, and breathing frequency derived from body surface displacement measurements correlated well with those determined spirometrically during a variety of activities. The coefficient of determination for tidal volume was 0.97 ± 0.2 for cycling, 0.93 ± 0.07 for arm cranking, 0.91 ± 0.05 for pulling, and 0.84 ± 0.12 for lifting. Our experiments showed that the breathing pattern was altered by the use of a mouthpiece and varied according to the type of activity. The use of a mouthpiece increased tidal volume by 34%, decreased the breathing frequency by 10%, and increased minute ventilation by 16%. There was more variability of these parameters during lifting and pulling activities than during cycling. The ventilation–heart rate relationship varied from subject to subject and was altered by the use of a mouthpiece. We found that ventilation measured in the field from body surface displacement correlated well with ventilation measured using the pneumotachograph (R^2 = 0.89). However, measurements of ventilation derived from heart rate were not as accurate as those derived from body surface displacements.

We concluded that minute ventilation can be measured accurately using body surface displacements in the laboratory and in the field. Heart rate can also be utilized, but factors affecting the minute ventilation–heart rate relationship, such as the use of a mouthpiece and range of heart rate, must be addressed to obtain more accurate estimates of minute ventilation.

This Investigators' Report is one part of the Health Effects Institute Research Report Number 59, which also includes an Investigators' Report by Samet and colleagues, a Commentary on both Investigators' Reports by the Health Review Committee, and an HEI Statement about these research projects. Correspondence concerning the Investigators' Report by McCool and Paek may be addressed to Dr. F. Dennis McCool, Pulmonary Division, Memorial Hospital of Rhode Island, 111 Brewter Street, Pawtucket, RI 02860.

Although this document was produced with partial funding by the United States Environmental Protection Agency under assistance agreement 816285 to the Health Effects Institute, it has not been subjected to the Agency's peer and administrative review and therefore may not necessarily reflect the views of the Agency, and no official endorsement should be inferred. The contents of this document also have not been reviewed by private party institutions including those that support the Health Effects Institute; therefore, it may not reflect the views or policies of these parties and no endorsement by them should be inferred.

INTRODUCTION

The dose of particles and gases that is retained in the lungs and airways is determined primarily by four factors: the concentration of particles and gases in the ambient air, their deposition and absorption fractions, their clearance, and the individual's level of ventilation. As the level of ventilation increases, airway and parenchymal damage from inhaled toxic particles and gases may be potentiated by increasing the total amount of contaminant inhaled (Horvath 1981) or by altering the breathing pattern (tidal volume [V_T]*, breathing frequency [f], and inspiratory time [T_I]) (Morgan et al. 1984, Bennett et al. 1985; Brain et al. 1988), thereby changing the distribution and deposition fraction of those particles inhaled. The role of ventilation and breathing pattern in estimating the dose of pollutants to the lungs, however, has been largely ignored because the breathing patterns and levels of ventilation adopted during specific physical tasks are unknown. Accordingly, a de-

* A list of abbreviations appears at the end of this report for your reference.

scription of these roles is needed to make a better assessment of the dose of gas and particles deposited in the lung.

Standard methods to measure ventilation and breathing pattern require the use of a face mask or mouthpiece. These measuring devices are known, however, to alter breathing patterns and minute ventilation ($\dot{V}_E$). Specifically, V_T and $\dot{V}_E$ increase, whereas breathing frequency decreases during quiet breathing (Gilbert et al. 1972; Dolfin et al. 1983; Rodenstein et al. 1985; Perez and Tobin 1985). With exercise, the effects of a mouthpiece are more varied; tidal volume either increases (Sackner et al. 1980a) or decreases (Stark et al. 1988). However, exercise studies have been limited to evaluating breathing during cycling or treadmill exercise; other typical physical tasks that may be encountered in daily settings have not been evaluated. Because a noninvasive means of measuring ventilation has been developed in subjects whose spinal attitude is changing (McCool et al. 1986; Paek et al. 1990), we have the capability to study natural (no mouthpiece) ventilation and breathing patterns during a variety of activities. This also has provided an opportunity to compare measurements of ventilation derived from heart rate (HR) with those derived from body surface displacement (BSD) measurements.

Heart rate increases as a linear function of oxygen consumption. Because ventilation also increases as a function of oxygen consumption, HR has been used to assess ventilation indirectly. Oxygen consumption or ventilation cannot be directly measured in the field for extended periods of time; however, with current technology, measurements of HR can be obtained easily in the field and can be collected for prolonged periods. Therefore, HR has been used in the field by several investigators as a surrogate measurement of ventilation (Astrand 1967; Harber et al. 1983; Manning and Griggs 1984; Louhevaara et al. 1985; Nielson and Meyer 1987). However, the $\dot{V}_E$-HR relationship is affected by many factors, such as gender (Astrand 1960; Higgs et al. 1967), age (Durnin and Mikulicic 1956), and physical fitness (Edwards et al. 1969), and therefore needs to be calibrated for each individual. In addition, the calibration of $\dot{V}_E$-HR relationships has been typically based upon measurements of $\dot{V}_E$ made utilizing a mouthpiece. Because these devices alter ventilation, but not the circulatory system's response to exercise, the $\dot{V}_E$-HR relationship obtained using a mouthpiece may be inappropriate for estimating $\dot{V}_E$ in the unencumbered (no-mouthpiece) state.

Measuring $\dot{V}_E$ without the use of a mouthpiece has been accomplished by using pneumobelts, magnetometers, or respiratory inductive plethysmographic (RIP) belts in normal subjects (Bendixen et al. 1964; Konno and Mead 1967; Gilbert et al. 1972; Askanazi et al. 1980). By using RIP belts, changes in the cross-sectional area of the rib cage and abdomen can be assessed. In contrast, with magnetometers, displacements of the rib cage or abdomen are measured in a single dimension. These previous approaches of measuring $\dot{V}_E$ are based on a two-degrees-of-freedom model (Konno and Mead 1967) in which the volume displacement of the respiratory system is equal to the sum of the volume displacements of the rib cage and abdomen, so that:

$$V_T = \alpha\Delta RC + \beta\Delta Ab, \qquad (1)$$

where ΔRC and ΔAb represent displacements or changes in the cross-sectional areas of the rib cage and abdomen, respectively, and α and β are volume-motion coefficients. With this method, ventilation can be estimated to within 10% of the minute ventilation measured at the mouth, as long as the subject is confined to one body position.

This two-variable approach, however, is severely limited by the changes in posture and spinal flexion that may occur in freely ranging subjects. Smith and Mead (1986) have demonstrated that changes in spinal flexion while lung volume is held constant can result in artifacts of volume change as great as 50% of the vital capacity. Paek and colleagues (1990) similarly found that the error related to displacement of the rib cage averaged 28% of vital capacity. In addition, they noted that changes in the volume-motion coefficients resulted in errors that averaged approximately 15% of the V_T. The systematic relationship of these errors with the degree of spinal flexion provided a mechanism whereby the addition of a factor that was proportional to the changes in spinal flexion could be added to the sum of ΔRC and ΔAb to improve volume estimates. The change in the axial dimension of the abdomen (distance between the midsternum [above the xiphoid process] and umbilicus [ΔXi]) proved to be such a factor. It is now feasible to make measurements of ventilation from BSD in freely ranging subjects by using the following three-variable approach:

$$V_T = \alpha\Delta RC + \beta\Delta Ab + \gamma\Delta Xi, \qquad (2)$$

where γ is the a volume-motion coefficient for the axial dimension.

SPECIFIC AIMS

Our specific aims were to:

1. Validate a model that uses BSD to measure ventilation;
2. Evaluate the mouthpiece and activity dependency of breathing patterns (T_I, V_T, and f);
3. Examine $\dot{V}_E$-HR relationships; and
4. Measure ventilation using BSD and HR in the field.

Table 1. Characteristics of Study Subjects

Subject	Age (years)	Gender	Height (cm)	Weight (kg)	FVC (L)	FVC (%)[a]	FEV_1 (L)	FEV_1 (%)[a]
Laboratory Study								
1	28	M	193	86.2	6.60	106	4.37	97
2	29	M	168	56.7	4.35	92	3.59	95
3	28	M	164	59.9	4.05	105	3.74	104
4	31	M	160	54.4	3.87	93	3.15	93
5	29	M	180	79.4	5.18	93	4.22	97
6	28	M	180	77.1	5.50	103	4.32	99
7	25	F	165	56.7	3.95	88	3.19	87
8	27	M	178	66.7	5.37	102	4.28	100
9	38	M	193	79.4	5.52	93	4.23	88
10	33	F	161	60.3	4.45	108	3.54	105
Field Study								
11	15	M	168	61.2	4.07	110	3.83	124
12	15	M	169	59.0	3.97	112	3.47	115
13	15	M	170	88.0	4.30	121	3.28	109
14	16	M	183	81.6	4.07	93	3.71	100
15	16	M	160	68.9	4.39	140	3.53	135
16	20	M	163	69.9	4.58	119	4.15	132
17	16	M	174	69.4	4.18	109	3.72	115
18	16	M	166	70.3	3.83	108	3.34	112
19	17	M	168	57.6	2.79	75	2.76	89

[a] The percentages predicted are derived from baseline data from a study by Crapo and associates (1981).

METHODS AND STUDY DESIGN

MEASUREMENTS OF CHEST WALL MOTION, TIDAL VOLUME, AND HEART RATE

A total of 19 subjects participated in the study. Their physical characteristics are summarized in Table 1. Respiratory inductive plethysmograph belts (Respitrace belts; Ambulatory Monitoring, Ardsley, NY) were used to measure ΔRC and ΔAb. The Respitrace belts were positioned around the rib cage at the level of the nipples and around the abdomen below the costal margin at the level of the umbilicus. Both Respitrace belts were held in place by adhesive tape and an elastic mesh (Figure 1). The location of the belts was marked on the skin and periodically checked to assure stability.

In addition, a pair of magnetometer coils was used to measure ΔXi. One coil was placed on the rib cage at the midsternum, and the other coil was attached to the middle of a 12-inch ruler, the ends of which were attached horizontally to the lateral abdominal walls at the level of the umbilicus. The ruler was used to minimize artifacts secondary to soft tissue–induced magnetometer coil rotation (Figure 1). The subjects experienced no discomfort wearing the Respitrace belts or magnetometers.

For experiments conducted in the laboratory, V_T was measured spirometrically as subjects breathed on a circuit connected to a 10-L wedge spirometer fitted with a carbon dioxide scrubber. The inspiratory resistance of the system was 0.7 cm water at 1 L/sec airflow, and the expiratory resistance was 1.5 cm water at 1 L/sec airflow. Prior to each run, the spirometer was filled with 7 liters of 100% oxygen. The signals from the magnetometer, Respitrace belts, and spirometer were displayed on an oscilloscope and simultaneously sampled at a frequency of 50 Hz per channel and stored on disk for further analysis. The spirometer's volume

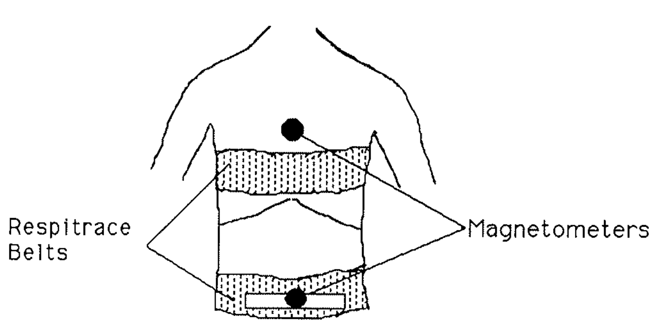

Figure 1. Schematic of the Respitrace belt and magnetometer positions on the rib cage and abdomen.

signal was used to calibrate the signals derived from BSD measurements (see Analysis: Specific Aims 1 and 2 section). Once calibrated, volume could be measured noninvasively, i.e., without a mouthpiece. For the field study, V_T was measured using a mouthpiece connected to a pneumotachograph. The flow signal from the pneumotachograph was integrated by computer to give V_T, which was used to calibrate the signals derived from BSD measurements. Heart rate was measured with electrodes placed over the precordium.

LABORATORY STUDIES

Protocol: Specific Aims 1 and 2. Validation of Body Surface Displacement Model to Measure Ventilation; Evaluation of Mouthpiece and Activity Dependency on Breathing Pattern

After the magnetometers and Respitrace belts were positioned, all subjects performed four different activities: lifting, pulling, arm cranking, and cycling.

The lifting activity was evaluated for two different work loads. It consisted of lifting a plastic jug that was either empty (light lifting) or filled with 16 kg of water (heavy lifting). The jug was lifted from the floor to a table that was 90 cm high, and then lowered to the floor every five seconds. This primarily provided exercise of the upper extremities.

The pulling activity consisted of pulling a rope that was attached to a pulley and a set of weights. The weights were raised 1 m every five seconds. The work load was varied by altering the weight attached to the pulley. Three different work loads were evaluated, with weights ranging from 15 to 35 pounds. Subjects performed this activity primarily by utilizing the trunk and lower extremity muscles; the activity also provided isometric exercise of the upper extremities.

Arm cranking was performed on a mechanically braked cycle ergometer (Monark model 818) at three different work loads. The cycle was positioned so that the subjects could sit in a chair and pedal with their arms rather than their legs. The arm cranking frequency was kept constant (40 rpm), and the work load was altered by changing the braking resistance over the range of 0.5 to 1.2 kp•m.

Subjects then were asked to cycle using their legs on the same mechanically braked cycle ergometer at three different work loads. The predicted maximal oxygen uptake was calculated from the weight and age of the subjects (Astrand 1960). The work loads then were chosen to yield approximately 20%, 40%, and 60% of the predicted maximal oxygen uptake (American College of Sports Medicine 1980). During the cycling runs, the pedaling frequency was kept constant for each subject, and only the braking resistance was changed to alter the work load.

The subjects performed each activity for eight to ten minutes. After four to six minutes, during which the subject reached a steady state, measurements of changes in chest wall dimension were obtained for another four minutes. During the last four minutes, half of the data were collected while subjects were breathing through a mouthpiece attached to a spirometer, and the other half were collected while the subjects were breathing without a mouthpiece. Each run was duplicated and the on- and off-mouthpiece data collection sequences were reversed. More than 15 minutes were allowed as a rest period between runs. To complete this protocol, subjects participated in two sessions on two separate days. We then calculated V_T with the magnetometer and calculated Respitrace signals by utilizing the previously described model (Equation 2).

Analysis: Specific Aims 1 and 2

The Respitrace, magnetometer, and volume signals were analyzed in the following sequence: (1) the volume-motion coefficients were determined for each subject; (2) the volume-motion coefficients were used to construct spirograms: (3) values for V_T, T_I, and total breath time (T_T) were calculated from the spirograms; and (4) the values of V_T, T_I, and f derived from BSD measurements were compared with those values obtained from the spirometer.

The volume-motion coefficients were calculated from data obtained while the subjects were breathing on the spirometer (in the laboratory) or through a pneumotachograph (in the field). End-inspiration and end-expiration were determined by picking the peaks and nadirs of the volume tracing. Tidal volume was calculated as the difference in volume between end-inspiration and end-expiration. Next, at end-inspiration and end-expiration, the simultaneous values of the chest wall dimensions were obtained. Then, by using multiple linear regression, each change in chest wall dimension was regressed against the corresponding change in spirometer volume for all breaths (the range of breaths analyzed was 25 to 92 for each period) while subjects breathed on the spirometer. The slope of the volume–chest wall dimension relationship represented a volume-motion coefficient (α, β, or γ) that was utilized in the three-variable model (Equation 2) to calculate V_T for each breath during both the on- and off-mouthpiece periods. Tidal volume was calculated as the change in volume between end-inspiration and end-expiration, which then was signified by reversal of the change in chest wall volume for more than 200 msec. Both T_I and T_T were calculated for each breath. The mean values of V_T, T_I, and f were calculated from pooled values for all breaths during the on- and off-mouthpiece periods. Breathing frequency was calculated as $(1/T_T) \times 60$. The $\dot{V}_E$ was calculated as the product of the mean value of V_T and f for these parameters derived from the spirometer ($\dot{V}_E$[actual]) and from BSD measurements ($\dot{V}_E$[BSD]). Using an IBM 286AT (IBM Corp., Armonk, NY) for data analysis, 20 to 25 minutes were needed to run a sequence of programs that started with the raw data and provided values for V_T and T_I for a two-minute period of data collection.

To evaluate the accuracy of the three-variable model, the calculated V_T, T_I, and f for each breath was compared with the actual V_T, T_I, and f measured simultaneously by the spirometer. The average values of V_T, T_I, and f derived from BSDs were used to assess the effect of a mouthpiece on ventilation and breathing patterns by comparing on- and off-mouthpiece periods, and to assess variations in breathing patterns between activities.

The accuracy of the BSD model for measuring ventilation and breathing pattern was assessed by comparing the results with spirometric data. Correlation coefficients obtained from the linear regression of V_T, T_I, and f data from the BSD measurements were compared with those coefficients determined from the spirometric data for each breath during the on-mouthpiece period. The significance of the differences in the V_T, T_I, and f parameters between on- and off-mouthpiece periods was assessed using a *t* test for paired variates. The variability of breathing patterns among activities was assessed by using a *t* test for paired variates to compare the coefficient of variation for V_T, T_I, and f obtained in one activity with the coefficient of variation of the corresponding variables obtained in another activity.

Protocol: Specific Aim 3. Evaluating the Volume of Expired Air Per Minute–Heart Rate Relationship

Subjects 1 through 10 performed lifting, pulling, arm cranking, and steady-state cycling activities as described previously. Heart rate was measured with electrodes placed over the precordium, and BSDs were measured while subjects breathed on (with mouthpiece) and off (no mouthpiece) a spirometer circuit. Subjects 11 through 19 (those used in the field study) performed a progressive exercise test. During this test, the subjects first sat on the bicycle and rested for five minutes. They then began to cycle at a fixed pedaling frequency of 60 rpm. The work load was increased by 10 w/min, up to 150 w, by changing braking resistance. The data were collected over a complete run, which consisted of five minutes of rest and 15 minutes of cycling. Each run was performed twice, once while breathing through a mouthpiece and the other time without a mouthpiece.

Analysis: Specific Aim 3

Plots of $\dot{V}_E$-HR were constructed during the on- and off-mouthpiece periods by using the ventilation values calculated from BSDs. Analysis of variance (ANOVA) was performed on these data to examine the effects of different activities and the use of a mouthpiece on the $\dot{V}_E$-HR relationship. With the progressive exercise test (subjects 11 through 19), $\dot{V}_E$(BSD) was measured every minute. The $\dot{V}_E$-HR relationship was described by two different models using the least squares method. There was a linear model:

$$\dot{V}_E = \text{HR} \times \text{slope} + \text{intercept} \tag{3}$$

and an exponential model

$$\dot{V}_E = \exp(\text{HR} \times \text{slope} + \text{intercept}). \tag{4}$$

The effects of a mouthpiece on the $\dot{V}_E$-HR relationship were evaluated by comparing the difference in the slopes and intercepts between the on- and off-mouthpiece periods by ANOVA for both the linear and exponential models. The mouthpiece effect also was evaluated by comparing differences in $\dot{V}_E$ between on- and off-mouthpiece periods at high (140 bpm) and low (80 bpm) HRs with a paired *t* test.

FIELD STUDY

Protocol: Specific Aim 4

Nine healthy male volunteers participated in this part of the study (Table 1, subjects 11 through 19). All of our subjects were students at a vocational high school studying automotive body repair. The chest-wall and HR signals were monitored for approximately two hours for each subject in the field. The field activities were divided into classroom, workshop, and laboratory sessions (Figure 2). During the classroom session, HR, ΔRC, ΔAb, and ΔXi were measured continuously for one hour. During this time the students were seated and attending a lecture. Tidal volume was measured using a pneumotachograph for five minutes every one-half hour.

For the workshop session, the same parameters were measured during light work such as painting, sanding, and welding. Light work continued for approximately one hour. All necessary equipment needed to measure and record

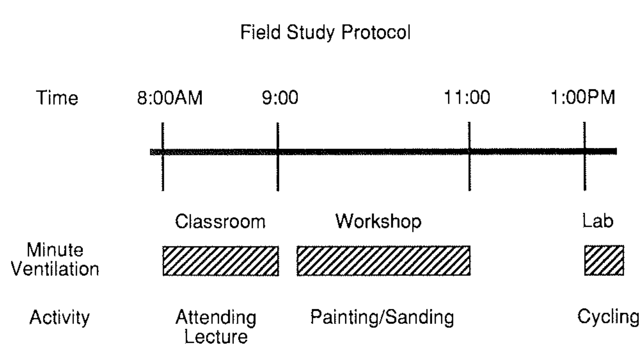

Figure 2. Schematic of the protocol used for the field study.

HR, ΔRC, ΔAb, and ΔXi was placed on a cart that could be pushed throughout the workshop, following the subjects as they engaged in the various work tasks.

The subjects then were taken to the laboratory, where they performed cycling activities. This consisted of a progressive exercise test performed with a cycle ergometer (Monark 818). As previously described, these cycling activities were performed with and without a mouthpiece. The $\dot{V}_E$(BSD) was calculated for both the on- and off-mouthpiece periods and was used to evaluate the effect of a mouthpiece on the $\dot{V}_E$-HR relationship.

Analysis: Specific Aim 4

Tidal volume measurements were obtained every one-half hour from the integrated flow signal of the pneumotachograph and were used to calibrate the BSD signals in the method described in the Analysis: Specific Aims 1 and 2 section. Once calibrated, V_T was also calculated from BSD by adding the ΔRC, ΔAb, and ΔXi dimensions (Equation 2). The $\dot{V}_E$(BSD) then was calculated as the product of the mean V_T and f.

Ventilation from HR ($\dot{V}_E$-HR) was calculated by first constructing the $\dot{V}_E$-HR relationship for each subject from the data obtained during cycling in the laboratory. These data were fit into linear (Equation 3) and exponential models (Equation 4), and the slope and intercept of the $\dot{V}_E$-HR relationship were calculated. The equations then were solved for $\dot{V}_E$($\dot{V}_E$-HR) by using the values of HR obtained in the field and the slopes and intercepts obtained during cycling in the laboratory. The $\dot{V}_E$-HR relationship was calculated using three methods:

1. The $\dot{V}_E$-HR(full) was the ventilation calculated from the slope and intercept of the $\dot{V}_E$-HR relationship over the full range of HRs.
2. The $\dot{V}_E$-HR(low) was the ventilation calculated from the slope and intercept of the $\dot{V}_E$-HR relationship over a range of HRs that was similar to the lower range noted in the field (60–90 bpm).
3. The $\dot{V}_E$-HR(exp) was the ventilation calculated from an exponential fit of the $\dot{V}_E$-HR relationship over a full range of HRs.

Plots of HR, $\dot{V}_E$(BSD), and $\dot{V}_E$-HR by the three methods with respect to time were constructed for the classroom and workshop sessions for each subject.

RESULTS

VALIDATION OF THE MODEL UTILIZING BODY SURFACE DISPLACEMENTS TO MEASURE VENTILATION

The values for V_T, T_I, and f derived from BSD measurements correlated well with those values determined utilizing the wedge spirometer (Table 2). The coefficient of determination (R^2) for V_T was 0.97 ± 0.02 for cycling, 0.93 ± 0.07 for arm cranking, 0.91 ± 0.05 for pulling, and 0.84 ± 0.12 for lifting. In general, as the activity required more change in posture, the R^2 value decreased. When expressed as a percentage of those values obtained spirometrically (Table 3), the average values of V_T measured by BSD were within 10% of those values obtained from the spirometer during all

Table 2. Coefficient of Determination Values[a] for Correlation Between Body Surface Displacement and Spirometric Measurements

Activity	Parameter	Mean ± SD	Range
Cycling	V_T	0.97 ± 0.02	0.92–0.98
	T_I	0.88 ± 0.01	0.66–0.97
	f	0.98 ± 0.02	0.94–0.99
Arm cranking	V_T	0.93 ± 0.07	0.67–0.98
	T_I	0.79 ± 0.14	0.47–0.98
	f	0.98 ± 0.01	0.96–0.99
Pulling	V_T	0.91 ± 0.05	0.62–0.97
	T_I	0.81 ± 0.06	0.75–0.94
	f	0.93 ± 0.06	0.86–0.99
Lifting	V_T	0.84 ± 0.12	0.82–0.96
	T_I	0.85 ± 0.06	0.71–0.87
	f	0.96 ± 0.04	0.82–0.97

[a] R^2 values were obtained by pooling data over all runs.

Table 3. Difference in Percentage Between Body Surface Displacement and Spirometric Measurements

Activity	Parameter	Mean ± SD[a] (%)	Range[b] (%)
Bicycling	V_T	1.98 ± 1.71	−5.65–7.96
	T_I	4.13 ± 3.12	−2.82–13.72
	f	0.16 ± 0.31	−0.33–1.97
Arm cranking	V_T	2.88 ± 3.08	−9.57–6.40
	T_I	5.38 ± 3.86	−4.55–15.36
	f	0.23 ± 0.42	−1.57–0.21
Pulling	V_T	2.04 ± 5.08	−6.78– −4.51
	T_I	6.28 ± 4.38	−4.55–15.36
	f	0.28 ± 0.18	−1.57–0.21
Lifting	V_T	2.88 ± 1.72	−4.81–9.25
	T_I	4.85 ± 2.83	−1.66–15.81
	f	0.27 ± 0.40	−0.50–0.61

[a] Mean ± SD of the absolute value of the difference.
[b] Range of the actual difference.

Table 4. Changes in Percentage in Breathing Pattern and Ventilation During Use of a Mouthpiece

Parameter	Mean ± SD (%)	Range (%)
V_T	$34.0^a \pm 25.6$	−25.5–92.4
f	$-10.2^b \pm 26.7$	−52.5–108.2
$\dot{V}_E$	$16.5^a \pm 28.0$	−41.9–153.8
T_I	$19.2^a \pm 30.1$	−50.3–89.0
T_T	$20.4^a \pm 32.6$	−52.0–110.5
T_I/T_T	-0.1 ± 7.4	−17.5–19.2

[a] $p < 0.01$ when compared with on- and off-mouthpiece periods by paired *t* test.

[b] $p < 0.05$ when compared with on- and off-mouthpiece periods by paired *t* test.

runs, and within 5% for 97 out of a total of 116 runs. For the group as a whole, there were no systematic differences between BSD measurements and spirometric values of V_T. The calculated values of T_I usually overestimated the wedge spirometer–derived value of T_I and were not as accurate (on average, 5% of the value measured spirometrically) as values for V_T (on average, 2% of the value measured spirometrically) or f (on average, 0.3% of the value measured spirometrically).

MOUTHPIECE AND ACTIVITY DEPENDENCY OF BREATHING PATTERN

Mean values of V_T, T_I, f, duty cycle (ratio of T_I to T_T [T_I/T_T]), and $\dot{V}_E$ were calculated from BSD during the on- and off-mouthpiece periods of each run. There were significant mouthpiece-induced changes in these variables. On average, during the on-mouthpiece period, $\dot{V}_E$ increased by 16%, V_T by 34%, and T_I by 19% (Table 4). In contrast, the duty cycle was unchanged and f decreased by 10%. These mouthpiece-induced changes were noted during all activities (Table 5) and during both light and heavy exercise. We also noted an activity dependency of breathing patterns.

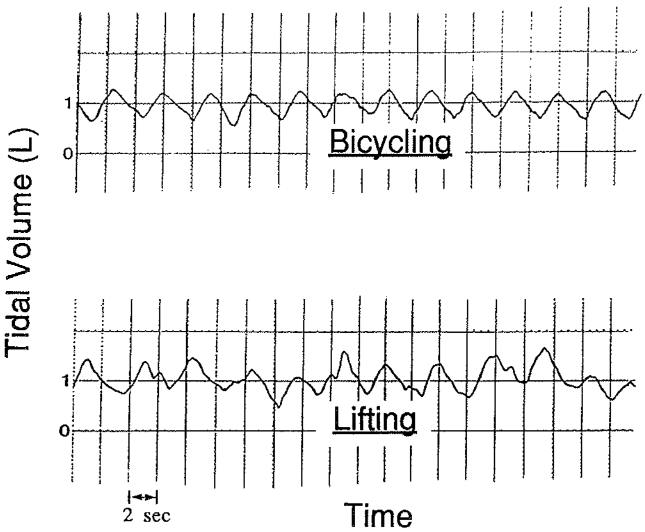

Figure 3. Spirogram obtained from BSDs during the off-mouthpiece period while a subject is cycling and lifting. Despite similar $\dot{V}_E$ values, there is more variability in V_T, T_I, and FRC during the lifting activity.

Representative spirograms obtained during the off-mouthpiece periods of cycling and lifting activities are shown for one subject (Figure 3). These spirograms illustrate the pronounced variability of V_T, T_I and functional residual capacity during activities such as lifting, and the monotonous V_Ts that occur with cycling. This variability in V_T and f for different activities was noted for all subjects. Although the mean values of V_T and T_I did not differ among activities (Figure 4), the coefficients of variation did differ. These differences were most striking when comparing lifting to cycling (coefficients of variation for V_T, T_I, and f of 39% vs. 20%, 38% vs. 21%, and 30% vs. 15% for lifting and cycling, respectively). Thus, although the levels of ventilation attained during the different activities may be similar, the physical maneuvers used to stimulate ventilation importantly influenced the breathing pattern.

Table 5. Changes in Percentage in Breathing Patterns During Different Activities and Use of a Mouthpiece[a]

Parameter	Cycling	Arm Cranking	Lifting	Pulling
V_T	29.2 ± 22.8^b	40.3 ± 28.5^b	33.3 ± 23.8^b	43.9 ± 30.9^b
f	-12.2 ± 20.3^b	-3.7 ± 41.1	-6.3 ± 26.6	-23.4 ± 20.1^b
$\dot{V}_E$	11.7 ± 24.9^c	37.4 ± 41.1	-6.3 ± 26.6	-23.4 ± 20.1^b
T_I	18.5 ± 26.9^b	9.6 ± 33.4^c	17.5 ± 35.7^b	33.7 ± 29.0^b
T_T	19.7 ± 27.0^b	8.7 ± 35.7^c	17.9 ± 43.2^b	38.1 ± 31.7^b
T_I/T_T	-0.6 ± 7.2	1.5 ± 6.8	1.4 ± 8.0	-2.6 ± 7.8

[a] Values given are means ± SD.

[b] $p < 0.01$ when compared with on- and off-mouthpiece periods by paired *t* test.

[c] $p < 0.05$ when compared with on- and off-mouthpiece periods by paired *t* test.

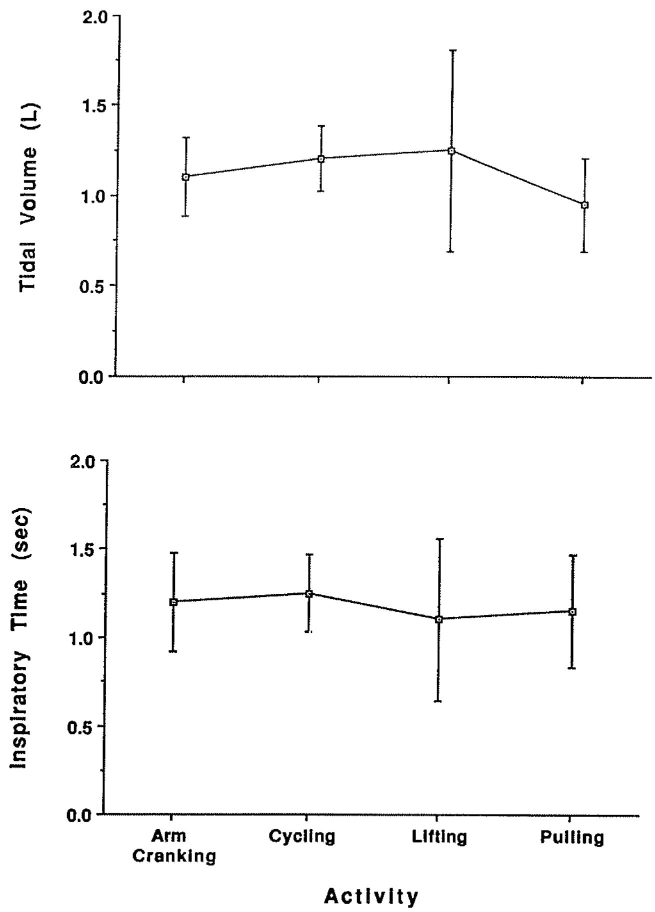

Figure 4. **Mean and standard deviation of V_T for the varied activities at similar mean levels of ventilation.** Although the V_T is similar among activities for the group, there is more variation in V_T and T_I during the lifting and pulling activities.

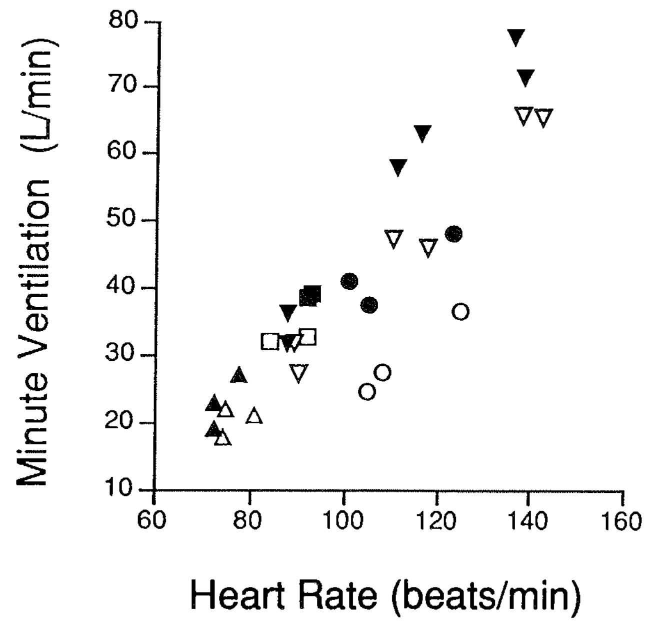

Figure 5. **Plot of $\dot{V}_E$-HR for a typical subject doing four different types of activity:** arm cranking (●, ○), cycling (▽, ▼), lifting (□, ■), and pulling (△, ▲) during on-mouthpiece (filled symbols) and off-mouthpiece (open symbols) measurements.

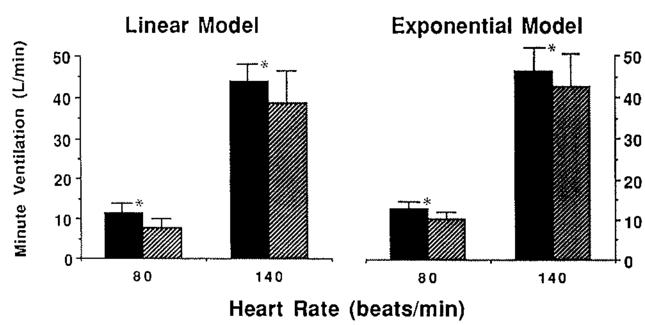

Figure 6. **Minute ventilation calculated using a linear model and an exponential model of the $\dot{V}_E$-HR relationship for each subject while cycling.** For both models, $\dot{V}_E$ is overestimated during the on-mouthpiece (filled bars) period for the low HRs and high HRs. Data are pooled from all subjects. Hatched bars = off-mouthpiece period.

VENTILATION–HEART RATE RELATIONSHIPS

The $\dot{V}_E$-HR relationship was described well by both the linear (R^2 = 0.88 ± 0.08 for the group, mean ± SD) and exponential models (R^2 = 0.84 ± 0.09 for the group). Activities performed using a mouthpiece were accompanied by higher levels of ventilation over the full range of HRs for each activity studied. This effect was noted for both linear and exponential models; an example of the mouthpiece effect is demonstrated in Figure 5 for one subject. For the group, the increase in ventilation associated with using a mouthpiece was significant ($p<0.001$). However, the effect of activity on this relationship did not reach statistical significance (p = 0.07).

Figure 6 summarizes the effect of a mouthpiece on the $\dot{V}_E$-HR relationship at low and high HRs, using both the linear and exponential models. For both models, there was a significant effect of the mouthpiece on ventilation at the low and high HR ranges ($p<0.01$), although the relative mouthpiece effect was more pronounced at the lower HRs. The increment in ventilation induced by a mouthpiece in this range (80 bpm) was as great as 111% (mean = 57 ± 15%, when using the linear model), whereas at the greater HR range of 140, the relative increase in ventilation was attenuated (mean = 15 ± 10%, when using the linear model). However, the absolute increase in ventilation at the low and high HR ranges was nearly the same, thereby suggesting that the mouthpiece shifted these relationships in a parallel fashion. Indeed, for group data, the slopes of the $\dot{V}_E$-HR relationships were nearly parallel when subjects breathed on and off the mouthpiece (Table 6).

Table 6. Individual Relationships Between Minute Ventilation and Heart Rate During Progressive Work Load Cycling For Field Study Subjects

	On Mouthpiece		Off Mouthpiece	
Subject	Slope	R^2	Slope	R^2
Linear Model				
11	0.44	0.87	0.43	0.83
12	0.39	0.85	0.44	0.96
13	0.54	0.89	0.61	0.90
14	0.71	0.94	0.70	0.94
15	0.73	0.89	0.68	0.91
16	0.60	0.92	0.52	0.91
17	0.38	0.93	0.25	0.78
18	0.32	0.92	0.27	0.95
19	0.30	0.93	0.22	0.73
Mean ± SD	0.49 ± 0.16	0.90 ± 0.03	0.46 ± 0.18	0.88 ± 0.08
Exponential Model				
11	0.017	0.71	0.021	0.77
12	0.021	0.83	0.031	0.92
13	0.022	0.85	0.024	0.93
14	0.026	0.91	0.025	0.93
15	0.024	0.82	0.028	0.77
16	0.023	0.92	0.025	0.85
17	0.015	0.95	0.013	0.75
18	0.019	0.97	0.018	0.95
19	0.013	0.92	0.012	0.72
Mean ± SD	0.020 ± 0.004	0.88 ± 0.08	0.022 ± 0.006	0.84 ± 0.09

Table 7. Mean Values of Minute Ventilation Obtained During the Five-Minute Calibration Periods

Subject	$\dot{V}_E$(actual) (L/min)	$\dot{V}_E$(BSD) (L/min)	R^2	Error[a] (%)	$\dot{V}_E$-HR(low) (L/min)	R^2	Error[a] (%)
11	13.1	14.7	0.86	12	9.5	0.16	27
12	12.4	10.6	0.93	14	1.2	0.32	90
13	19.3	21.6	0.88	13	14.7	0.10	24
14	18.9	22.9	0.83	21	18.0	0.46	5
15	11.6	13.2	0.88	14	7.0	0.34	40
16	19.3	19.2	0.81	0	9.7	0.74	50
17	12.1	12.3	0.85	2	13.9	0.27	15
18	6.1	8.3	0.83	36	4.9	0.76	20
19	13.1	14.5	0.95	11	9.7	0.82	26
Mean	15.3	14.0	0.87	14	9.8	0.42	33

[a] Error is percentage of change of $\dot{V}_E$(BSD) or $\dot{V}_E$-HR(low) when compared with $\dot{V}_E$(actual) from pneumotachograph.

MEASUREMENT OF VENTILATION USING BODY SURFACE DISPLACEMENT AND HEART RATE IN THE FIELD

To assess the accuracy of measurements of $\dot{V}_E$ based upon the BSD model in the field, values of ventilation derived from BSD measurements were compared with the actual values of ventilation obtained with the pneumotachograph. (Actual ventilation was measured during the calibration period, which lasted five minutes and occurred at one-half-hour intervals.) The $\dot{V}_E$(BSD) correlated well with $\dot{V}_E$(actual) values (R^2 = 0.87) and were within 13.6% of the $\dot{V}_E$(actual) (Table 7). In contrast, the best model for $\dot{V}_E$-HR ($\dot{V}_E$-HR[low]) did not correlate well with the $\dot{V}_E$(actual) (R^2 = 0.42) and was within only 33% of the $\dot{V}_E$(actual). The

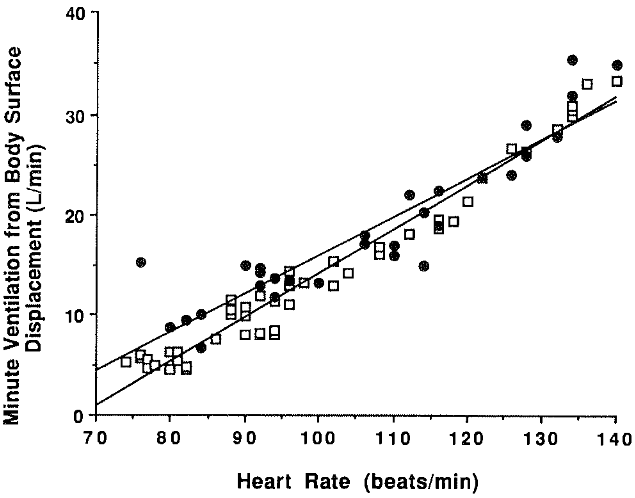

Figure 7. The $\dot{V}_E$-HR relationship for one subject. Ventilation was derived from BSD ($\dot{V}_E$[BSD]) while breathing on (●) and off (□) a mouthpiece. The slope (linear regression) of the off-mouthpiece relationship is used to predict $\dot{V}_E$ from HR measurements obtained in the field.

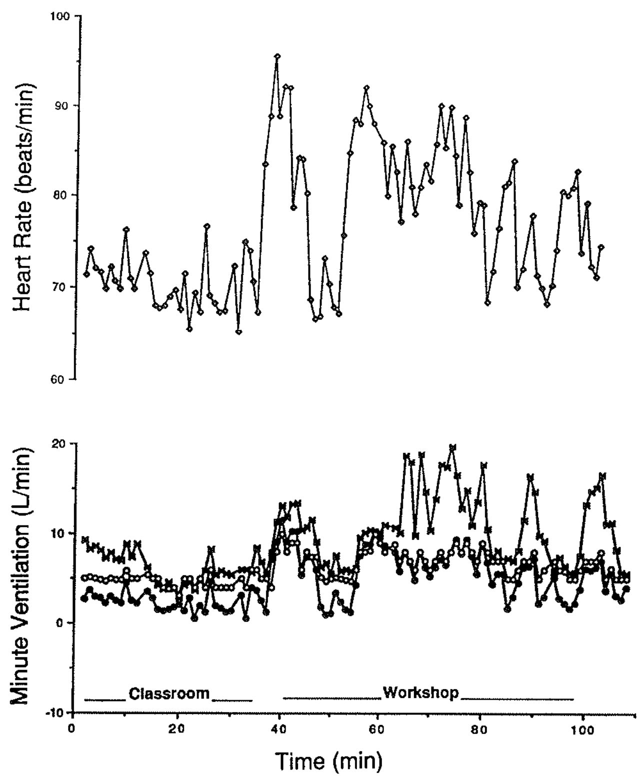

Figure 8. The time course of changes in HR (top graph) **and $\dot{V}_E$** (bottom graph) **obtained in the field during the classroom (0 to 40 minutes) and workshop sessions (40 to 100 minutes) for one subject.** $\dot{V}_E$-HR(BSD) (×) paralleled the changes in HR (◇) and, as expected, was greater during the workshop session than during the classroom session. Changes in $\dot{V}_E$-HR(low) (○) did not parallel changes in HR as well as $\dot{V}_E$(BSD). $\dot{V}_E$-HR(full) (●) was not as accurate as values obtained using $\dot{V}_E$-HR(low), and occasionally gave negative values of ventilation (data not shown).

limited accuracy of $\dot{V}_E$-HR during the calibration period can be explained in part by the intermittent deep breaths that the subjects were instructed to take during this time. The deep breaths provided a wide range of volumes over which the volume-motion coefficients could be calculated. These breaths would have increased ventilation but not HR, thereby uncoupling the $\dot{V}_E$-HR relationship and making any prediction of $\dot{V}_E$ derived from HR inaccurate during the calibration period.

All values of ventilation derived from HR data obtained in the field were based upon the $\dot{V}_E$-HR relationship characterized in the laboratory during cycling. An example of the $\dot{V}_E$-HR relationship from one subject and the values of ventilation derived from it during the field sessions are shown in Figures 7 and 8, respectively. The changes in HR, $\dot{V}_E$(BSD), $\dot{V}_E$-HR(low), and $\dot{V}_E$-HR(full) were plotted every two minutes for each subject during the classroom and workshop tasks. When examining changes in HR with regard to time (Figure 8), we found that the mean (± SD) range of HR was narrower (64 ± 9 to 96 ± 10) in the field study activities than the range noted during the cycling tests performed subsequently in the laboratory (71 ± 8 to 153 ± 7) ($p<0.01$) (Table 8). Furthermore, for most subjects, the resting HR was lower in the field than in the laboratory. In general, changes in $\dot{V}_E$(BSD) were qualitatively similar to the changes in HR.

The $\dot{V}_E$-HR(full) values were only within 45% of the values determined by BSD. For three of nine subjects, $\dot{V}_E$-HR(full) had negative values of ventilation. The inaccuracy of $\dot{V}_E$-HR(full) may be related to applying slopes and intercepts derived over the full range of HR to a range of HR that is lower in the field. Indeed, the slope and intercept of the $\dot{V}_E$-HR relationship over the full range of HR differed from those over the lower range of HR (0.27 L/beat and −11 L/min vs. 0.61 L/beat and 49 L/min) ($p<0.001$). The estimates of ventilation derived from HR were more accurate when the slopes and intercepts (Equation 3) were obtained from a calibration curve derived from a range of HRs similar to that measured in the field ($\dot{V}_E$-HR[low]). When using the linear model over a similar range of HRs or when using the exponential model, $\dot{V}_E$-HR(low) was within 21% of $\dot{V}_E$(BSD), and $\dot{V}_E$-HR(exp) was within 19% of $\dot{V}_E$(BSD) (Table 9).

Values of ventilation derived from BSD in the field were qualitatively similar to simultaneous changes in HR. In contrast, transients in $\dot{V}_E$ were not reflected as well by $\dot{V}_E$-HR(low) as they were by $\dot{V}_E$(BSD) (Figure 8). This can be seen as less variation in $\dot{V}_E$-HR(low) (SD = 1.9 L/min) when compared with $\dot{V}_E$(BSD) (SD = 4.2 L/min) (Table 9). Just as the mean HR was greater in the field during the workshop session than during the classroom session,

Table 8. Range of Heart Rate During Tasks in the Field Study and During Subsequent Cycling Activity in the Laboratory

Subject	Minimum HR Recorded in Field	Maximum HR Recorded in Field	Minimum HR Recorded on Cycle	Maximum HR Recorded on Cycle
11	59	91	62	162
12	65	95	74	140
13	80	104	88	146
14	57	85	60	130
15	71	101	76	160
16	53	78	68	156
17	74	107	72	144
18	65	90	68	172
19	52	108	76	171
Mean ± SD	64 ± 9	95 ± 10	72 ± 8	153 ± 14

Table 9. Mean Values of Minute Ventilation Obtained During the Classroom and Workshop Session[a]

Subject	$\dot{V}_E$(BSD) (L/min)	$\dot{V}_E$-HR(low) (L/min)	Error[b] (%)	$\dot{V}_E$-HR(full) (L/min)	Error[b] (%)	$\dot{V}_E$-HR(exp) (L/min)	Error[b] (%)	R^{2c}
11	11.4 ± 4.5	10.5 ± 1.5	8	6.7 ± 3.4	41	10.7 ± 0.9	6	0.51
12	9.3 ± 4.0	4.7 ± 2.8	49	3.8 ± 3.4	59	5.4 ± 1.8	41	0.75
13	16.3 ± 6.7	16.0 ± 1.5	2	12.7 ± 2.7	22	16.3 ± 1.2	0	0.72
14	12.9 ± 5.0	9.7 ± 3.3	25	6.0 ± 4.6	53	11.0 ± 2.3	15	0.54
15	10.9 ± 4.9	10.0 ± 4.0	8	9.2 ± 4.5	16	10.2 ± 2.7	6	0.70
16	6.0 ± 3.1	5.6 ± 1.9	7	1.2 ± 3.3	80	7.0 ± 1.1	16	0.70
17	9.2 ± 4.1	13.4 ± 0.3	45	15.5 ± 1.5	68	13.3 ± 0.2	45	0.75
18	8.1 ± 1.5	5.6 ± 0.9	31	4.5 ± 1.3	44	6.1 ± 0.6	25	0.51
19	9.0 ± 3.9	10.2 ± 1.1	13	7.3 ± 3.3	19	10.3 ± 0.9	14	0.67
Mean	10.3 ± 4.2	9.5 ± 1.9	21 ± 17	7.4 ± 3.4	45	10.0 ± 1.3	19 ± 16	0.64

[a] Values given are means ± SD.

[b] Error is the percentage of change of $\dot{V}_E$-HR(low), $\dot{V}_E$-HR(full), or $\dot{V}_E$-HR(exp) when compared with $\dot{V}_E$(BSD).

[c] Coefficient of determination for the relation between $\dot{V}_E$-HR(exp) and $\dot{V}_E$(BSD).

$\dot{V}_E$(BSD) was greater during the workshop session when the subjects were engaged in light physical activity.

DISCUSSION

This study has enabled us to (1) validate a model used to measure ventilation from BSD, (2) demonstrate that breathing patterns vary with activities, (3) evaluate the effects of a mouthpiece on breathing pattern and minute ventilation, (4) describe factors that alter the $\dot{V}_E$-HR relationship, (5) evaluate the limitations of using HR in the field as a surrogate for measurements of ventilation, and (6) explore the utility of measurements of BSD to determine ventilation in the field.

VALIDATION OF THE USE OF BODY SURFACE DISPLACEMENTS TO MEASURE VENTILATION

Previous investigators have used either pneumobelts, magnetometers, or RIP belts to measure noninvasively tidal volume and minute ventilation in normal subjects (Konno and Mead 1967; Stagg et al. 1978; Sackner et al. 1980b). These approaches are based on a two-degrees-of-freedom model (Konno and Mead 1967), in which the volume displacement of the respiratory system is equal to the sum of the volume displacements of the rib cage and abdomen (Equation 1). This two-variable approach is severely limited by changes in posture and spinal flexion that may occur in freely ranging subjects. Smith and Mead (1986) and Paek and colleagues (1990) have demonstrated that changes in RC and Ab dimensions as great as 50% of vital capacity may occur during isovolume spinal flexion. Accordingly, we utilized a measure of spinal flexion (xiphi-abdominal distance) to develop a three-degrees-of-freedom model (McCool et al. 1986; Paek et al. 1990) (Equation 3). As we have shown previously, the three-degrees-of-freedom model was more accurate during activities that incorporated changes in spinal attitude or body position ($R^2 = 0.97 \pm 0.15$ for the three-degrees-of-freedom model, and $R^2 = 0.83 \pm 0.15$ for the two-degrees-of-freedom model), whereas both models accurately predicted V_T when body position was unchanged (McCool et al. 1986).

The average values of V_T, T_I, and f estimated from BSD correlated well with spirometric values of V_T, T_I, and f (Tables 2 and 3). The mean values of V_T were within 5% of spirometry values, and the mean values of T_I were within 10% of spirometric values. Although there were no systematic differences between V_Ts measured by spirometry and by BSD, T_Is were usually more prolonged when measured by BSD. This systematic overestimation was due mainly to the definition of a breath. Inspiration was defined as the interval between the lowest and highest points in the spirogram, and any fluctuations or noise in the preinspiratory baseline signal occasionally made the lowest point of the signal occur earlier than the actual initiation of the breath. Thus, the three-variable model of BSD can be used adequately to assess breathing patterns and ventilation in the laboratory.

MOUTHPIECE AND ACTIVITY DEPENDENCY OF BREATHING PATTERN

Effects of a mouthpiece on breathing pattern have been studied during quiet breathing (Gilbert et al. 1972; Askanazi et al. 1980; Sackner et al. 1980a; Dolfin et al. 1983; Rodenstein et al. 1985), bicycle ergometry (Sackner et al. 1980a), and carbon dioxide rebreathing (Newton et al. 1983; Weissman et al. 1984). There is general agreement that using a mouthpiece and nose clip will increase V_T and $\dot{V}_E$ with variable effects on f. In our study, only data from BSD was used to compare the on- and off-mouthpiece breathing patterns and the activity dependency of breathing patterns. Thus, any systematic error in our modeling would have been minimized in this comparison. Our findings of an increase in V_T ($34 \pm 26\%$) and a decrease in f ($10 \pm 27\%$) when subjects breathed through the mouthpiece is similar in direction and magnitude to previous findings. The increase in V_T more than offset the reduction of f, with minute ventilation greater (17%) during the on-mouthpiece period. The changes in magnitude of V_T and $\dot{V}_E$ during activities primarily involving the upper limbs (lifting and pulling) were similar to those noted during lower extremity exercise. Thus, the respiratory apparatus used to measure ventilation will itself decrease f and increase V_T and $\dot{V}_E$ over a range of activities and activity intensities.

The mouthpiece effect on V_T was attenuated at the higher levels of ventilation. At the highest work loads, the increase in V_T during the on-mouthpiece period was almost negated by the decrease in f. The attenuation of the mouthpiece effect on $\dot{V}_E$ and V_T at higher levels of ventilation may reflect the alinearity of the V_T-$\dot{V}_E$ relationship that has previously been described during exercise (Hey et al. 1966). As respiratory drive and $\dot{V}_E$ increase, V_T increases linearly to approximately one-half of vital capacity. Subsequently, the increase in ventilation is primarily attributed to an increase in f, not in V_T.

When analyzing ventilation in terms of respiratory drive (V_T/T_I) and respiratory timing (T_I/T_T), we found that the mouthpiece primarily increased drive, whereas timing generally did not change. What increases drive is speculative. One study suggests that nasal or oral receptors are irritated by the nose clip or mouthpiece (Dolfin et al. 1983), whereas others suggest that oral breathing itself, not the irritation, is the responsible mechanism (Hirsch and Bishop 1982; Perez and Tobin 1985; Rodenstein et al. 1985).

When comparing different activities that had nearly equivalent levels of $\dot{V}_E$, we found more variability in V_T and T_I during lifting than during cycling. The lack of variation in V_T when cycling may be due in part to entrainment of breathing with pedaling frequency (Bechbache and Duffin 1977; Hill et al. 1988; Takano 1988). Entrainment of breathing with a constant pedaling frequency would fix T_I and T_T. Because $\dot{V}_E$ remains constant at a given work rate, V_T would be similar from breath to breath. In contrast, during a lifting activity, breathing was interrupted by the activity as the size and timing of each breath became more variable. The marked variations in V_T and T_I for a given individual during the lifting and pulling activities indicate that people engaged in different physical activities may have different breathing patterns, which depend upon the type of activity and are independent of activity intensity.

VENTILATION–HEART RATE RELATIONSHIPS

Cycling on a bicycle ergometer is the usual method employed to describe the $\dot{V}_E$-HR and $\dot{V}_E$–oxygen consumption relationships for an individual. The types of tasks encountered in the field are, however, diverse, and may differ from cycling in terms of muscle recruitment patterns and metabolic costs. This raises questions about the usefulness of a $\dot{V}_E$-HR relationship calibrated during cycling and then applied to other activities that may be encountered in the field.

This issue was addressed in part by Petrofsky and Lind (1978), who compared lifting to cycling and observed that, for a given oxygen consumption, there was no difference in ventilation and HR between cycling and lifting. However, several studies have reported that, at a given level of oxygen consumption, exercise involving small muscle groups, such as arm cranking, results in higher levels of ventilation and HR than those activities involving relatively large muscle groups, such as cycling (Bobbert 1960; Bevegaard et al. 1966; Stenberg et al. 1967; Jensen 1972). Our results evaluating four different types of activities (cycling, arm cranking, lifting, and pulling) indicate that the $\dot{V}_E$-HR relationship was not specific to activity. This finding suggests that the $\dot{V}_E$-HR relationship calibrated during one type of activity can be used to predict ventilation during another type of activity. Our results of the activity specificity of this relationship, however, should be interpreted with caution. The range of ventilation and HR that we obtained during each of these activities was not as broad as the range obtained during cycling. As Samet and associates have shown in the accompanying Investigators' Report, Assessment of Heart Rate As a Predictor of Ventilation (Samet et al. 1993), the $\dot{V}_E$-HR relationship for activities other than cycling that use broader ranges of HR and $\dot{V}_E$ may differ from that derived during cycling.

The use of a mouthpiece strikingly altered the $\dot{V}_E$-HR relationship for most subjects and for the group as a whole. The increase in ventilation when using a mouthpiece while breathing at rest has been previously described; our results extend these findings to a variety of activities (Gilbert et al. 1972). The mechanism of this mouthpiece-induced increase in ventilation remains unknown. It is unlikely that the increased resistance imposed by a mouthpiece will increase energy consumption, and thereby ventilation, because the mouthpiece was attached to large-caliber tubes, and the increased resistance was minimal. Moreover, increases in dead space with the use of a mouthpiece cannot fully explain the increase in ventilation for a given HR. Sackner and coworkers (1980a) measured the effective dead space volume contributed by a mouthpiece and observed that ventilation still increased when using a mouthpiece, even after the dead space effect was eliminated. Our findings of a nearly parallel shift of the $\dot{V}_E$-HR relationship with the use of a mouthpiece also do not support this dead space explanation. At low levels of ventilation (low HR range of the $\dot{V}_E$-HR relationship), the dead space ventilation is a greater fraction of V_T than at higher levels of ventilation. Consequently, the slope of the $\dot{V}_E$-HR relationship would be less during mouthpiece breathing. For the above reasons, the mouthpiece effect cannot be explained by the minimal increases in dead space volume related to breathing through a mouthpiece. Other possible factors that could increase respiratory drive and alter breathing pattern include irritation of nasal or oral receptors by the mouthpiece or nose clip, as previously described.

The $\dot{V}_E$-HR relationship can be adequately described with either a linear or exponential model. However, the efficiency of these models may differ when used to predict ventilation from HR in the field. In contrast to the linear model, when using the exponential model at the lower HR range, increments in ventilation are blunted, even with modest changes of HR. Because most daily tasks are performed over a low range of HRs, the linear model, which most accurately describes the $\dot{V}_E$-HR relationship over the lower range of HRs, would best predict $\dot{V}_E$ in the field.

MEASUREMENT OF VENTILATION USING BODY SURFACE DISPLACEMENT AND HEART RATE IN THE FIELD

Ventilation is not measured directly in the field because of the impracticality of performing continual measurements of airflow at the mouth in ambulatory subjects. Because the technology is readily available to measure HR continually in the field, and because changes in the HR continually parallel changes in metabolic rate, investigators have used HR to measure $\dot{V}_E$. However, there are several factors that affect the $\dot{V}_E$-HR calibration curves and therefore limit their applicability to the field. These factors include: (1) the use of a mouthpiece; (2) the lower ranges of HR found in the field, which require extrapolation of the $\dot{V}_E$-HR relationship to lower ranges of HR than those obtained in the laboratory; (3) the effect of work load on the slope of the $\dot{V}_E$-HR relationship; and (4) the uncoupling of HR from ventilation, such as that which occurs with voluntary increases in $\dot{V}_E$ or in patients who take cardiac medication such as beta blockers. Some of these factors can be addressed so that HR can be used as a more accurate index of ventilation in the field. First, the mouthpiece effect can be addressed by using BSD to measure ventilation during the $\dot{V}_E$-HR calibration procedure in the laboratory. Second, differences in the range of HRs between cycling in the laboratory and tasks in the field can be addressed by analyzing the calibration curve over a range comparable with that seen in the field. Third, assurance that the subject is well relaxed in the laboratory prior to any testing may alleviate any increase in basal HR related to anxiety.

When using $\dot{V}_E$-HR calibration curves derived without the use of a mouthpiece and applying them to HR data obtained in the field, we found that the exponential model and the linear model that was derived from the low range of HRs ($\dot{V}_E$-HR[low]) gave better estimates of ventilation than those derived from the full range of HRs ($\dot{V}_E$-HR[full]). Although these were the most accurate models derived from HRs, they were not as accurate as the $\dot{V}_E$(BSD), and they were unable to pick up rapid transients of ventilation as well as the $\dot{V}_E$(BSD) did. As Dr. Samet's study reveals (see accompanying Investigators' Report), HR can be measured easily and accurately in the field for prolonged periods of time using a lightweight portable device. The changes in HR measured with this device parallel changes in $\dot{V}_E$, as shown in Figure 8, but these measurements may be inaccurate. However, as we have demonstrated, the accuracy of using HR to measure $\dot{V}_E$ can be improved when the $\dot{V}_E$-HR relationship is characterized without using a mouthpiece and the regression equation for $\dot{V}_E$ is derived over the same range of HRs that would be encountered in the field.

This study has demonstrated the feasibility of using BSD to determine ventilation in the field. The advantages of this technique, when compared with the use of HR, include a more accurate assessment of $\dot{V}_E$ and the ability to provide measurements of V_T and f. The disadvantage of this technique is that, at this time, it is cumbersome, technically more demanding, and therefore more time consuming. Furthermore, the RIP belts may slip on the soft tissues of the chest wall, resulting in artifacts. Although this could pose more of a problem when fitting the rib cage belt for women, we found no greater difficulty in securing the Respitrace belts for the women in this study. However, recent advances in technology may make noninvasive ambulatory monitoring of $\dot{V}_E$ less cumbersome. New RIP technology makes it possible to power RIP belts with a battery for up to eight hours. If the belts are calibrated in two or three different body positions and a device is used to indicate in which body posture the activity is performed, RIP belts may be used in conjunction with a recorder to measure $\dot{V}_E$. Although the currently available magnetometers are bulky, they can be modified for prolonged use in the field by decreasing the size of the device and their power requirements. By taking advantage of lower noise circuits, the signal and the electrical power needed to elicit the signal can be reduced while still preserving an acceptable signal-to-noise ratio. Further amelioration of power demands can be accomplished by gating the signals. The advantage of a magnetometer system over RIP belts is the continual recording of ΔXi.

CONCLUSIONS

Using a respiratory apparatus such as a mouthpiece or nose clips can increase ventilation and alter breathing patterns (V_T, T_I, and f) at rest and during a variety of activities. In addition, because the breathing pattern depends on the activity being performed, there may be more variability in V_T and T_I for one activity than for another (e.g., lifting vs. cycling). The mouthpiece effect on ventilation limits the utility of HR as a predictor of $\dot{V}_E$ in the field. Other factors limiting the use of HR in the field include differences in the range of HRs in the laboratory and during tasks in the field, the need to extrapolate the HR data in the laboratory to lower values in the field, and differences in slope of the $\dot{V}_E$-HR relationship over low and high ranges of HRs. Estimates of ventilation from HR can be optimized by addressing these factors; however, they still are less accurate than estimates based on BSD and fail to pick up transients of ventilation and ventilation derived from BSD. Furthermore, ventilation estimates using BSD can be analyzed in terms of breathing pattern, and consequently, changes in respiratory drive and timing. In addition to ventilation, these are important factors that may influence gas and particle deposition in the lungs. If it is possible to use new technology to monitor $\dot{V}_E$ and breathing patterns noninvasively in the field for six to eight hours, accurate assessments of $\dot{V}_E$ during environmental exposures to particles and gases both at home and in the work place would be possible.

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ABBREVIATIONS

ΔAb	change in abdominal cross-sectional area during a breath
bpm	beats per minute
BSD	body surface displacement
f	breathing frequency
FEV_1	forced expiratory volume in one second
FVC	forced vital capacity
HR	heart rate
kp•m	kilopond•meter
ΔRC	change in rib cage cross-sectional area during a breath
R^2	coefficient of determination
RIP	respiratory inductive plethysmographic (belt)
T_I	inspiratory time
T_I/T_T	ratio of inspiratory time to total breath time
T_T	total breath time
$\dot{V}_E$	volume of air expired per minute
$\dot{V}_E$(actual)	ventilation calculated from average values of V_T and f derived from the spirometer or pneumotachograph
$\dot{V}_E$(BSD)	ventilation derived from body surface displacement measurements
$\dot{V}_E$-HR(exp)	ventilation derived from heart rate measurements using an exponential model of the ventilation–heart rate relationship

over the full range of heart rate elicited in the laboratory

$\dot{V}_E$-HR(full) ventilation derived from heart rate measurements using a linear model of the ventilation–heart rate relationship over the full range of heart rates elicited in the laboratory

$\dot{V}_E$-HR(low) ventilation derived from heart rate measurements using a linear model of the ventilation–heart rate relationship over only the low range of heart rates elicited in the laboratory

V_T tidal volume

ΔXi change in the axial (midsternum [above the xiphoid process] to umbilicus)

Assessment of Heart Rate As a Predictor of Ventilation

Jonathan M. Samet, William E. Lambert, David S. James, Christine M. Mermier, and Thomas W. Chick

ABSTRACT

The rate of ventilation and route of breathing (i.e., nasal versus oronasal) are potential determinants of pollutant doses to target sites in the lung. However, the lack of accurate methods for ambulatory measurement of ventilation has hindered estimation of exposure and dose in freely ranging individuals, complicating the interpretation of the relationships among exposure, dose, and response in epidemiological studies. The goal of this project was to develop and validate a method of monitoring ventilation for large-scale epidemiologic investigations. We estimated ventilation for individual subjects from ambulatory heart rate monitoring, using the relationship between ventilation and heart rate that had been obtained during exercise testing.

Fifty-eight subjects participated in the study, which included healthy adults and children, and subjects with lung and heart disease. Subjects performed cycle exercise and tasks involving lifting and vacuuming. Work loads of progressive and variable order were used in the testing. Conventional methods were used to measure heart rate and total ventilation, and a sampling mask was developed to measure the partitioning of breathing between oral and nasal routes. The minute ventilation–heart rate relation was evaluated under steady-state and varying work loads. In a second phase, subjects wore wristwatch monitors that recorded their heart rates, minute by minute, throughout the day. Subjects recorded activities, locations, and levels of exertion. Two 16-hour monitoring periods were obtained from each subject.

The laboratory findings documented considerable intersubject variability in the minute ventilation–heart rate relation with a two- to five-fold range in the coefficients describing the change in ventilation relative to heart rate. This variation implies that individual testing is required to derive accurate predictive equations. Minute ventilation–heart rate regressions for the maximal progressive exercise test and for the test with a nonprogressive submaximal work load sequence were comparable, indicating that varying the sequence of work loads does not substantially affect the minute ventilation–to–heart rate ratio. During upper body work (e.g., lifting), the minute ventilation–to–heart rate ratio was one-third greater than during lower body exercise. Diverse patterns of partitioning breathing between oral and nasal routes were observed with increasing oral ventilation in most subjects as work load increased. In the field, heart rate and activity patterns were monitored successfully in adults and children with low rates of instrument failure and noncompliance. The laboratory data base, in combination with the Heartwatch recordings, was used to estimate ventilation by type of activity and location. The distribution of heart rate values showed less than 10% of the values were above 100 beats per minute; with the exception of vigorous exercise activities, the 95th percentile of estimated ventilation was approximately 20 L/min for healthy men and women.

We have found that heart rate monitoring can be used feasibly to estimate ventilation in the field. However, the minute ventilation–heart rate relation needs to be established in the laboratory for each subject with emphasis on the lower range of work loads appropriate for most activities in the community setting. Our pilot data suggest that few nonoccupational activities, other than vigorous exercise, are associated substantially with elevated heart rates and ventilation.

This Investigators' Report is one part of the Health Effects Institute Research Report Number 59, which also includes an Investigators' Report by McCool and colleagues, a Commentary on both Investigators' Reports by the HEI Health Review Committee, and an HEI Statement about these research projects. Correspondence regarding the Investigators' Report by Samet and associates may be addressed to Dr. Jonathan M. Samet, Professor of Medicine, New Mexico Tumor Registry, University of New Mexico Medical Center, 900 Camino de Salud NE, Albuquerque, NM 87131-5306.

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INTRODUCTION

Vehicle exhaust is a complex mixture comprised of hundreds of compounds that are emitted as gases and liquid and solid aerosols (Bates and Watson 1988). As the exhaust cools and its components age, vapors may be adsorbed onto particles, and droplets may be formed by condensation. An extensive literature indicates that the delivery of particles and gases to tissues in the respiratory tract depends not only on inhaled concentrations of the pollutants, but also on reactivity and solubility for gases (Overton and Miller 1988; Ultman 1988), and on physical characteristics for particles (Brain and Valberg 1979). Pollutant doses to target tissues also vary with the level of volume of expired air per minute (minute ventilation) ($\dot{V}_E$)*, the extent of oral versus nasal breathing, and personal characteristics, including age and the presence of abnormalities or disease in the airways or the lung parenchyma (Schlesinger 1988).

The present air quality standards for pollutants resulting from vehicle exhaust establish maximum emission levels at the tailpipe and maximum permissible concentrations in ambient air. These standards have been formulated on the basis of information derived from laboratory investigations in in vivo and in vitro models, controlled human exposure studies, and epidemiologic research. Exposure, rather than the dose delivered to target tissues, has been considered more commonly in the animal and human investigations. However, the dose delivered to sites of injury in the lung is a more appropriate indicator of potential hazard than concentration measured in inhaled air (National Research Council 1991).

Thus, to date, few epidemiological studies have included estimates of pollutant dose delivered to target structures in the respiratory tract. Most often, exposure measures have been used as surrogates for dose, and misclassification of dose has been a likely consequence. For many pollutants, the lack of information on levels and patterns of ventilation introduces substantial uncertainty in exposure-dose relations in epidemiological studies and risk assessments. The development and application of methods to measure the level and route of ventilation in ambulatory subjects would improve the understanding of the health effects associated with inhaling vehicular emissions. In combination with new techniques for personal exposure assessment (National Research Council 1991), the estimation of ventilation and characterization of breathing route could lead to more refined estimates of lung dose for individual subjects in epidemiologic studies.

Passive monitors and portable continuous instrumentation for monitoring personal and small area exposures are now available for carbon monoxide, nitrogen dioxide, volatile organic compounds, and particles (Sexton and Ryan 1988; National Research Council 1991), and surveys of personal exposure have been conducted on large numbers of people (Akland et al. 1985; Quackenboss et al. 1986; Wallace 1987). Time-activity patterns in populations have been measured accurately by using written diary records and 24-hour recall interviews (Robinson 1988). Similarly, levels of physical activity have been assessed by questionnaire methods in epidemiological studies (Washburn and Montoye 1986).

SPECIFIC AIMS

In response to the Health Effects Institute's Request for Applications 88–1, this pilot project was designed to develop methods for use in field studies to estimate $\dot{V}_E$ associated with specific activities and to assign the route of breathing as nasal alone, combined oral and nasal, or oral alone. The rationale for each of the specific aims follows.

The previous epidemiological studies of Raizenne and Spengler (1989) and Shamoo and coworkers (1990) documented the feasibility of estimating $\dot{V}_E$ from ambulatory recordings of heart rate (HR). However, in these studies, determinants of the minute ventilation–heart rate ($\dot{V}_E$-HR) relation had not been characterized completely. Our research focused on quantification of the $\dot{V}_E$-HR relation and assessment of its variability with different types of exercise in the laboratory. The primary research questions addressed in the laboratory research were:

1. Does the $\dot{V}_E$-HR relation differ between lower body exercise (e.g., walking or riding a cycle) and upper body exercise (e.g., arm work or lifting)?
2. Does the $\dot{V}_E$-HR relation differ between exercise using progressively increasing work loads and exercise in which the order of work load presentation is varied (i.e., nonprogressive)?
3. Does the $\dot{V}_E$-HR relation differ between individuals?
4. Does the $\dot{V}_E$-HR relation vary on the basis of personal characteristics such as gender, age, level of fitness, and health status?

We chose to use standard exercise testing protocols involving the lower and upper body, and progressive and nonprogressive work loads to characterize the $\dot{V}_E$-HR relation for healthy children and adults and for persons with asthma, chronic obstructive pulmonary disease, and ischemic heart disease. These data were used to develop predictive equations to estimate $\dot{V}_E$ from average minute HRs.

A secondary objective of the laboratory phase of the research was to characterize the partitioning of ventilation between the nose and mouth in relation to the total ventilation. These data were to be used to estimate the proportion of oral breathing in the field setting. A mask was developed to separate mouth and nose breathing and to sample inspiratory ventilation during exercise. The variation in parti-

* A list of abbreviations appears at the end of this report for your reference.

tioning was evaluated in relation to personal characteristics of the subjects; these included gender, age, level of fitness, nasal health history, and nasal airway resistance as measured by rhinomanometry. We also proposed to develop questionnaires to assess nasal diseases and symptoms. Two questionnaires were developed to address acute and chronic nasal symptoms, structural nasal problems, and nasal and sinus conditions such as allergic rhinitis and chronic sinusitis.

A third objective was to develop methods for monitoring HR and activities in the field. We chose to use an inexpensive wristwatch-style monitor (Heartwatch, Computer Instruments Corp., Hempstead, NY) to measure HR. The accuracy of the Heartwatch was verified by comparing its readings with HRs measured by conventional ambulatory electrocardiogram (ECG) recorders. A time-activity diary was developed to monitor types of activity and levels of exertion. Two days of monitoring were obtained for each subject. Using the predictive equations developed in the laboratory, the $\dot{V}_E$ was estimated by classes of activity, age group, gender, and health status.

Thus, the objectives of this research were to develop and test the complement of methods needed to estimate the level and pattern of ventilation in the community setting. The hypotheses and study design focused on characterizing the utility of the approach and factors determining the variability of the $\dot{V}_E$-HR relation. The methods were tested on men, women, and children. Substantial information was collected on activities and HRs, and estimates of ambulatory ventilation were derived from the data. However, it was not a goal of this project to assemble a representative sample of subjects to assure generalization to the population.

METHODS AND STUDY DESIGN

SUBJECT SELECTION

Fifty-eight subjects were recruited for laboratory testing and ambulatory monitoring, and full data were obtained from 56 subjects (Table 1). Males and females were selected to represent the full age spectrum, from school-age child to older adult. A small number of subjects with diseases of interest in air pollution research were also included in the sample (i.e., patients with asthma, chronic obstructive pulmonary disease, or ischemic heart disease). Because a population-based sample was not needed to fulfill the aims of the project, a variety of approaches was used to enlist volunteers. Subjects were recruited from advertisements posted at the University of New Mexico and the Albuquerque Veterans Administration Medical Centers, from advertisements placed in a senior citizens' newsletter, and from personal contacts of the investigators.

All potential candidates for participation were screened to obtain representation across the full age range, and inquiry was made to ensure that subjects were able to ride a stationary cycle. If subjects reported no history of heart or lung disease, they were classified as "healthy." Subjects with asthma, chronic obstructive pulmonary disease, or ischemic heart disease were recruited from patient populations at the two medical centers. The presence of these conditions was confirmed at the first laboratory visit by subject history and clinical evaluation by the project physiologist, Dr. Thomas Chick. The majority of the subjects were able to complete the full protocol's three days of laboratory testing and two days of ambulatory monitoring. Attrition was

Table 1. Subject Characteristics[a]

Group	n	Age Range	Body Mass Index (kg/m^2)	$\dot{V}_{O_2}$ Maximum (mg/kg/min)	FEV_1 (% Predicted)[b]
Healthy men	15	18–72	23.6 (3.4)	45.7 (16.7)	97.4 (11.7)
Healthy boys	6	10–17	18.8 (3.0)	46.6 (5.8)	99.7 (4.3)
Healthy women	16[c]	21–72	23.4 (2.1)	32.2 (8.9)	110.5 (10.7)
Healthy girls	6	7–17	18.3 (3.6)	38.0 (5.0)	104.5 (11.8)
Subjects with asthma	5	11–43	19.9 (4.0)	41.7 (6.7)	98.6 (15.3)
Subjects with chronic obstructive pulmonary disease	5	55–70	25.3 (3.6)	16.7 (4.6)	32.4 (9.5)
Subjects with ischemic heart disease	5	53–72	28.7 (5.2)	21.8 (5.7)	86.5 (20.7)
Total	58				

[a] All values are group means, with standard deviations in parentheses.

[b] Prediction equations for adults are from Crapo and associates (1981); equations for children are from Knudson and coworkers (1976).

[c] Number includes two subjects who dropped out of the study after completing day 1.

limited to two subjects who moved out of town after completing the first day of laboratory tests. No subjects were lost due to noncompliance or declining to participate further. Subjects who completed the protocol were given a participation award of $100.

LABORATORY METHODS

Research subjects came to the Veterans Administration Medical Center for exercise and pulmonary function testing on three separate days (Table 2).

First Day of Testing

On the first day, informed consent was obtained, and each subject completed a standardized respiratory symptoms questionnaire based on the questionnaires developed for children and adults by the American Thoracic Society (Ferris 1978) and a second questionnaire that characterized upper respiratory health during the previous week and for each of the days of testing. This second questionnaire was administered at the beginning of each day of laboratory testing. Height, weight, resting HR, and blood pressure were measured. Standard spirometry was performed with measurements of maximum voluntary ventilation (MVV), forced expiratory volume in one second, peak expiratory flow, and forced vital capacity. To determine nasal airway resistance (NAR), posterior rhinomanometry was performed (see below). After rhinomanometry, subjects performed a graded maximal exercise test on a cycle ergometer (Model ER1, Erich Jaeger, Rockford, IL).

Table 2. Schedule of Laboratory Testing and Field Monitoring

Laboratory Testing

Day One
- Informed consent
- Respiratory health history questionnaire
- Questionnaire on upper respiratory disease
- Questionnaire on today's nasal symptoms
- Height and weight
- Rhinomanometry
- Spirometry
- Basal ventilation and heart rate monitoring
- Progressive maximal exercise test on cycle ergometer

Day Two
- Questionnaire on today's nasal symptoms
- Rhinomanometry
- Basal ventilation and heart rate monitoring
- Submaximal exercise test on cycle ergometer to measure oronasal partitioning of ventilation
- Vacuuming task exercise test

Day Three
- Questionnaire on today's nasal symptoms
- Basal ventilation and heart rate monitoring
- Nonprogressive submaximal exercise test on cycle ergometer
- Lifting exercise test

Field Monitoring

Day One
- Heartwatch monitoring of heart rate for 16 hours
- Activity diary

Day Two
- Heartwatch monitoring of heart rate for 16 hours
- Activity diary

Maximum Exercise Test Protocol

Each subject breathed through a mouthpiece and wore a noseclip. Respiratory gases were collected in a bag and analyzed with a Jaeger Incarepulmobile system. Prior to each exercise testing session, the metabolic cart was calibrated using certified gas standards for oxygen, carbon dioxide, and nitrogen (Med Point Technologies, Cucamonga, CA). Minute ventilation, oxygen uptake ($\dot{V}_{O_2}$), carbon dioxide output ($\dot{V}_{CO_2}$), respiratory exchange ratio (R), tidal volume (V_T), respiratory frequency (f), and HR were integrated over 15-second intervals. Before performing the exercise test, each subject sat quietly in a chair, and basal metabolic data, ventilation, and HR were measured. Basal measurements were not a part of the original battery of tests and were added to the protocol halfway through data collection. Thus, basal measurements are only available for 21 subjects, primarily children and subjects with heart and lung disease who were tested later in the project. During the exercise test, the work load was increased according to the subject's age, gender, and fitness level. During the first three minutes of the test, the subjects pedaled with no resistance. Resistance was increased progressively in one-minute intervals. For fit men, the work load was increased 25 watts per minute; for relatively unfit men, the work load was increased 20 watts per minute. Similarly, for fit and unfit women, the work load was increased 20 and 15 watts per minute, respectively. For boys and girls, work loads were increased by increments of 8, 10, or 15 watts per minute. The subjects were instructed to pedal at a rate of 60 revolutions per minute; the allowable range was 50 to 70 revolutions per minute. The test was terminated when the subjects could no longer maintain 50 revolutions per minute. The maximum work load achieved was used to determine the work loads applied in subsequent tests. A Borg scale was administered at the completion of each exercise test (American College of Sports Medicine 1991) (Table 3); the subjects rated their highest level of exertion.

Table 3. Borg Scale for Rating Perceived Level of Exertion[a]

Level	Description	Example
0	Nothing at all	Lying down
0.5	Very, very light	Sitting at rest
1	Very light	Standing still
2	Light	
3	Moderate	
4	Somewhat heavy	
5	Heavy	
6		
7		
8		
9	Very, very heavy	Almost maximal
10	Maximal	Cannot maintain for any length of time

[a] Adapted from American College of Sports Medicine (1991).

Second Day of Testing

On the second day, the subjects performed a simulated household task, pushing a vacuum cleaner over a carpet. This was a progressive test in which the work load was increased by adding weight to the vacuum cleaner and speeding up the cadence of the motion. One kilogram of weight was added each minute, and the cadence was increased in increments from 20 strokes per minute to 35 strokes per minute in the final stages of the test. In stages 2, 3, and 4, subjects moved the vacuum cleaner by arm motion only. In stages 5 through 12, subjects were instructed to step forward and then backward with each stroke of their arms. Exercise was terminated at 50% of maximum HR reserve ([maximal HR − resting HR] × 0.50 + resting HR). Respiratory gases and ventilation were sampled through a mouthpiece, and HR was continuously monitored. After a half-hour rest period for recovery, rhinomanometry was performed to characterize day-to-day variability in NAR. A second exercise test was performed on this day. Each subject rode the stationary cycle while wearing a face mask to sample inspiratory ventilation by nasal and oral routes.

Partitioning of Ventilation

A nasal continuous positive airway pressure mask (Respironics, Murrysville, PA) was modified with a plastic partition to separate the nose from the mouth. Three sizes of masks were developed to accommodate various face sizes. The mask was held on the subject's face by a system of adjustable Velcro straps. The adequacy of mask seals was checked before the beginning of each test. Subjects exercised on a cycle ergometer. The work load was increased in five-minute stages, and the load increments were twice as large as those used in the maximal exercise test. The test was terminated either at 85% of maximum HR or 20 L/min of oral respiration, whichever condition occurred first. Subjects were told to breathe normally during the exercise and were not informed of the test objective.

A single pneumotachograph was used to measure inspiratory flows during the oral and nasal exercise test. Thus, serial and not simultaneous measurements were obtained. The nasal and oral compartments were sampled alternatively by switching valves to direct flow to the pneumotachometer. Each stage of the graded exercise test lasted for five minutes. The first two minutes of each stage were used to achieve physiological steady state at the new work load level, and recordings were disregarded during this interval. From 2 to 2.75 minutes within a stage, total ventilation was measured by combining flow from both the oral and the nasal ports on the mask. After 2.75 minutes, the valves were switched, and only the oral flow was sampled. After 3.75 minutes, the valves were switched again, and only nasal flow was sampled. After 4.75 minutes, the valves were switched back to combined oral and nasal flow, and the load increment was increased.

Third Day of Testing

On the third day, the subjects performed another upper-body exercise test, lifting. Keeping cadence to a metronome, each subject raised a canvas bag to which weight was added progressively. During each minute of the test, 1 kg of weight was added. The speed of the lifting sequence started at 25 lifts per minute and was increased to 30 lifts per minute in the seventh minute of exercise. The test was terminated at 50% of maximum HR reserve. Each subject breathed through a mouthpiece during the lifting test. After one-half hour for recovery, another exercise test was conducted on the cycle ergometer to assess the $\dot{V}_E$-HR relation under varying submaximal work loads. As opposed to steadily increasing work loads, the order of presentation was nonprogressive and followed 0%, 30%, 10%, 50%, 20%, and 40% of maximum work load achieved on the maximal exercise test. Respiration was sampled through a mouthpiece during the varying work load test.

Rhinomanometry

Posterior rhinomanometry was performed to determine NAR. Before each series of measurements, pressure and flow measurements were calibrated using a water manometer and rotameter, respectively. In conducting the measurement, air flow was delivered to the nasal airway via a nasal continuous positive airway pressure mask held over the nose by the subject. A 2-cm-i.d. hose connected a port on the mask to a pneumotachometer. Pressure measurements were obtained through a separate port on the mask and through a 0.4-cm-i.d. tube held as far back as possible in the mouth. The difference in the measurements from the pressure transducer and the flow measurements were recorded on an X-Y recorder. Subjects were asked to take a slow tidal breath while the mask and oral tube were in place. Ten measurements were performed per day on the first eight subjects to characterize variability; in the remaining subjects, a minimum of five curves per test day were obtained. Pressure measurements were read from the inspiratory portion of the tracings at flow increments of 0.05 L/sec (2 mm on the chart paper). The origin was determined by the point at which the curve crossed the zero-flow line. Nasal airway resistance was calculated at a flow of 0.25 L/sec.

Laboratory Validation of Heartwatch Measurements

The accuracy of Heartwatch pulse data was evaluated by comparing measurements made by this device with measurements from a conventional ECG. Comparisons were made in the laboratory during exercise on the cycle ergometer, and in the field, against ambulatory ECG recordings.

In the laboratory, the Heartwatch was worn by 10 of our research subjects during graded exercise tests; these measurements were compared with HRs measured on the Jaeger Incarepulmobile system. The HR displayed on the liquid crystal diode of the Heartwatch was updated every five seconds. This value was compared with HRs averaged over 15-second intervals on the Jaeger system. Ten observations were sampled from each of 10 exercise tests, providing 100 comparison observations.

FIELD MONITORING OF ACTIVITIES AND HEART RATE

Field Validation of the Heartwatch Measurements

To validate the accuracy of the Heartwatch under realistic conditions representative of the field setting, a Heartwatch was worn by 10 patients at the Veterans Administration Medical Center who were undergoing ambulatory ECG monitoring for clinical purposes. These patients were not subjects in our $\dot{V}_E$-HR research program. The patients selected for monitoring were relatively free of ventricular and supraventricular ectopy and had no known conduction defects. The continuous ECG recording was sampled at 10 points throughout the day. One-minute tracings were printed out, and the QRS complexes within the interval were counted by hand and compared with the one-minute average measured for the same period by the Heartwatch.

Field Monitoring of Heart Rate

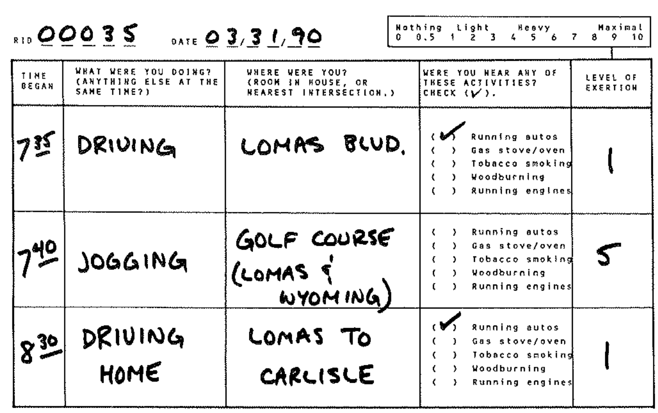
On two separate days, subjects wore the Heartwatch for 16-hour periods to monitor HR during daily activities. The Heartwatch is comprised of two components: a small transmitter (137 × 30 × 12 mm) and a wristwatch-style receiver. Both components are powered by lithium batteries. The transmitter is worn on the chest, at the base of the sternum over the xyphoid process. Two disposable electrodes are used to attach the transmitter. Heart rate is monitored by detecting the R wave of the ECG. Pulse signals are transmitted to the receiver, in which a microprocessor calculates the mean minute HR and stores the rates in random access memory. During the monitoring period, the Heartwatch recorded minute average HRs; these recordings later were transferred to a desktop computer.

While wearing the Heartwatch, the subjects were free to pursue a wide range of activities, but were cautioned not to immerse the Heartwatch in water. The memory capacity of the Heartwatch is limited to 16 hours, and, therefore, the subjects removed the monitors at the end of their waking days. When the transmitter and electrodes were removed from the subject's chest, transmission was stopped, and after 15 minutes of receiving no signal, the monitor automatically turned off.

Time-Activity Diary Records

During ambulatory monitoring, subjects maintained a written record of activities in a logbook (Figure 1). The subjects were directed to record all activities lasting 15 minutes or longer. Written instructions on the diary use were printed on the inside of the front cover of the diary, along with an example. Subjects also used the diary to record their location (indoors, outdoors, and in transit), and proximity to sources of air pollutants. For each entry, subjects completed the Borg scale, which is a subjective rating of the level of exertion (Table 3).

Activities recorded in the diary were coded according to a standard system used in national time-activity studies (Robinson 1988), and a three-digit classification system was


RID 00035 DATE 03/31/90

Nothing 0 0.5 Light 1 2 3 4 Heavy 5 6 7 8 Maximal 9 10

TIME BEGAN	WHAT WERE YOU DOING? (ANYTHING ELSE AT THE SAME TIME?)	WHERE WERE YOU? (ROOM IN HOUSE, OR NEAREST INTERSECTION.)	WERE YOU NEAR ANY OF THESE ACTIVITIES? CHECK (✓).	LEVEL OF EXERTION
7:35	DRIVING	LOMAS BLVD.	(✓) Running autos ( ) Gas stove/oven ( ) Tobacco smoking ( ) Woodburning ( ) Running engines	1
7:40	JOGGING	GOLF COURSE (LOMAS & WYOMING)	( ) Running autos ( ) Gas stove/oven ( ) Tobacco smoking ( ) Woodburning ( ) Running engines	5
8:30	DRIVING HOME	LOMAS TO CARLISLE	(✓) Running autos ( ) Gas stove/oven ( ) Tobacco smoking ( ) Woodburning ( ) Running engines	1

Figure 1. Sample page from time-activity diary.

developed to code locations. Each diary was coded by two observers; discordant codes were reviewed and corrected by a third observer.

QUALITY ASSURANCE AND QUALITY CONTROL

The data were collected in accordance with the general directives of the Health Effects Institute. The quality assurance and quality control program included a written protocol, standard operating procedures, documentation of qualifications and training of personnel, written and computerized records, and standardized data-processing and verification procedures. All raw written and digital records were archived, as were processed and analytical files.

STATISTICAL METHODS AND DATA ANALYSIS

Analyses of the exercise test data primarily involved conventional parametric statistical methods, including linear regression and analysis of variance. The analyses were performed using PC-SAS software (Version 6.04, Statistical Analysis Software Institute, Cary, NC) (1985) and Lotus 1-2-3 software (Release 3, Lotus Development Corp., Cambridge, MA).

Analysis of the Minute Ventilation–Heart Rate Relation

The relation between $\dot{V}_E$ and HR was characterized using ordinary least squares regression methods. Minute ventilation measurements were transformed to the natural log due to the curvilinear nature of the relationship between HR and ventilation (Astrand and Rodahl 1977). A regression equation by subject was constructed for each exercise test. The effects of exercise test type, age, and gender on the estimated slopes of the $\dot{V}_E$-HR regressions were tested using the *t* test and analysis of variance.

Analysis of Oral Versus Nasal Selection of Breathing and Nasal Airway Resistance

The relation of oral $\dot{V}_E$ to total $\dot{V}_E$ was evaluated with linear regression analysis. Total $\dot{V}_E$ during the exercise test was adjusted to the percentage of MVV to standardize the measurement of total $\dot{V}_E$ for all subjects. Oral $\dot{V}_E$ was regressed on total $\dot{V}_E$ at different percentages of MVV.

For the rhinomanometry curves, the constants K_1 and K_2 from Rohrer's equation (Pressure = $[K_1 \times \text{flow}] + [K_2 \times \text{flow}^2]$) (Cockcroft et al. 1979) were determined using linear regression analysis of pressure divided by flow versus flow. For both sessions for each individual, the mean and SD were obtained by using the estimate for resistance at a flow of 0.25 L/sec from each regression equation. To test for differences in the median NAR and in the SD of NAR, the nonparametric Kruskal-Wallis test was used. For paired comparisons of NAR and the SD of NAR, log-transformed values were used to produce a more normal distribution.

Prediction of Ventilation from Ambulatory Heart Rate Recordings

Ambulatory HR data and diary information on the type of activity were used to predict $\dot{V}_E$ in the field setting. The $\dot{V}_E$-HR relation derived from the maximal exercise test for each individual was used to predict $\dot{V}_E$ for all activities.

RESULTS

The presentation of results follows the organization of the major components of the study: the laboratory results followed by the results of the HR monitoring in the field. The major components of the laboratory testing included a description of the $\dot{V}_E$-HR relation by type of exercise and subject characteristics, characterization of oral and nasal breathing patterns, measurement of NAR, and assessment of the accuracy of the Heartwatch. The major components of the ambulatory monitoring were a description of activity patterns, a description of HR by type of activity, and the estimation of $\dot{V}_E$.

SUBJECT CHARACTERISTICS

The $\dot{V}_E$-HR relations were based on 58 subjects who ranged in age from 7 to 72 years (Table 1). On average, the healthy subjects showed a high level of aerobic fitness and had spirometric values near 100% of the predicted values. The body mass index of most healthy subjects was within the normal range (18–26 kg/m^2) (American College of Sports Medicine 1991). The subjects with heart and lung disease showed a lower level of aerobic fitness and ventilatory function than the healthy subjects. The subjects with chronic obstructive pulmonary disease and ischemic heart disease had higher body mass index values than the healthy subjects.

VENTILATION–HEART RATE RELATION

The regression analyses for healthy men and women are described in Table 4. The distributions of slopes describing the $\dot{V}_E$-HR relation of healthy subjects for each type of exercise test are illustrated in Figures 2 and 3 for men and women, respectively. On the average, $\dot{V}_E$ increased more quickly than HR for exercises involving the upper body, such as lifting and vacuuming, than for lower body exercises. This difference in the $\dot{V}_E$-HR relation between upper body activity and cycling was more apparent in healthy women than in healthy men, especially during the vacuum-

Table 4. Statistics for Regressions of Natural Log of Ventilation[a] on Heart Rate by Subject Group and Type of Exercise

		Progressive Maximal Cycling				Nonprogressive Submaximal Lifting				Progressive Load Lifting				Progressive Load Vacuuming			
		Slope		R^2		Slope		R^2		Slope		R^2		Slope		R^2	
Group	*n*	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD
Healthy men	15	0.022	0.004	0.97	0.02	0.023	0.005	0.89	0.06	0.033	0.013	0.73	0.14	0.044	0.021	0.62	0.22
Healthy boys	5	0.019	0.002	0.96	0.02	0.017	0.004	0.74	0.21	0.022	0.006	0.51	0.35	0.030	0.008	0.55	0.30
Healthy women	15	0.022	0.004	0.96	0.03	0.020	0.006	0.81	0.18	0.030	0.013	0.66	0.19	0.045	0.015	0.69	0.13
Healthy girls	5	0.020	0.004	0.95	0.04	0.017	0.005	0.74	0.25	0.026	0.010	0.52	0.18	0.036	0.008	0.54	0.22
Subjects with asthma	5	0.022	0.001	0.97	0.01	0.021	0.002	0.88	0.03	0.032	0.016	0.82	0.04	0.032	0.019	0.36	0.25
Subjects with COPD	5	0.018	0.003	0.96	0.03	0.017	0.009	0.72	0.41	0.021	0.008	0.82	0.14	0.025	0.007	0.82	0.13
Subjects with ischemic heart disease	5	0.023	0.003	0.94	0.04	0.036	0.008	0.82	0.12	0.035	0.015	0.59	0.18	0.050	0.020	0.72	0.07

[a] Values for natural log of ventilation were calculated in L/min. Heart rate was measured in bpm.

ing task (Figures 2 and 3). The regression coefficients obtained with the progressive maximal exercise cycling test and the nonprogressive submaximal cycling tests were not significantly different (p = 0.397) (Table 5). However, all other paired comparisons of the slope coefficients obtained with the various exercise tests showed statistically significant differences ($p < 0.001$).

Substantial interindividual variability in the $\dot{V}_E$-HR relation was observed (Figures 4, 5, and 6). The slopes of the regression of $\dot{V}_E$ on HR also varied across categories of gender and age. Within specific strata defined by age, gender, and type of exercise test, substantial variation was evident among individual subjects. For example, Figure 7 shows the results for adult males during the progressive maximal exercise test.

The estimated $\dot{V}_E$s from the $\dot{V}_E$-HR regressions for the different types of exercise tests at given HRs were compared for healthy adult men and women (Table 6). The slopes comparing the ventilation estimates from the progressive maximal and nonprogressive submaximal cycle tests, as well as vacuuming and lifting tests, were close to 1.0 for HRs ranging from 60 to 140 beats per minute (bpm). However, the slopes were substantially less than 1.0 for maximal cycle exercise, compared with vacuuming and lifting tests.

For 21 subjects, measurements of the $\dot{V}_E$-HR relation were obtained while the subjects were seated at rest. These resting measurements were compared with preexercise measurements of $\dot{V}_E$ and HR obtained while subjects stood at rest for three minutes before beginning the vacuuming and lifting tests. The HR and $\dot{V}_E$ measurements were substan-

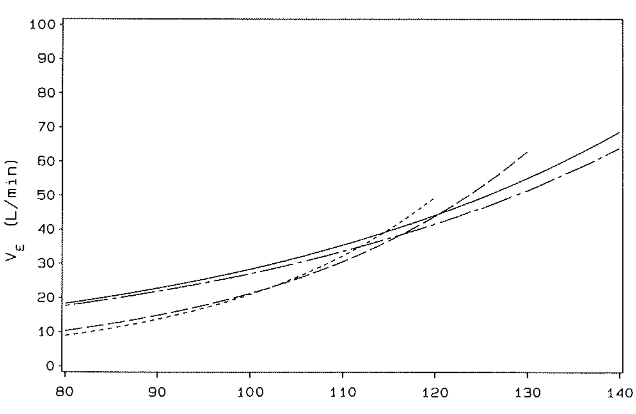


Figure 2. Plot for mean of regressions of $\dot{V}_E$ on HR by type of exercise test for 15 healthy men. Solid line = progressive maximal cycling test; dot dash line = non progressive submaximal cycling test; short dash line = progressive load vacuuming test; long dash line = progressive load lifting test.

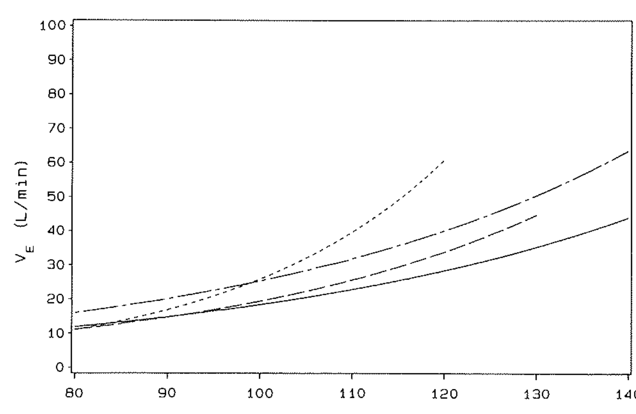


Figure 3. Plot of mean regressions on $\dot{V}_E$ on HR by type of exercise for 15 healthy women. Solid line = progressive maximal cycling test; dot dash line = nonprogressive submaximal cycling test; short dash line = progressive load vacuuming test; long dash line = progressive load lifting test.

Table 5. Comparison of Differences in Individual Ventilation-on–Heart Rate Regression Slopes for Healthy Adults

Exercise Test	n	Mean Difference ($\times 10^3$)	SD
Progressive maximal cycling vs. progressive load vacuuming	41	− 20.0[a]	15.2
Progressive maximal cycling vs. progressive load lifting	42	− 8.3[a]	11.2
Progressive maximal cycling vs. nonprogressive submaximal cycling	42	0.5	4.1
Progressive load vacuuming vs. progressive load lifting	41	12.3[a]	14.6
Progressive load vacuuming vs. nonprogressive submaximal cycling	41	20.4[a]	14.7
Progressive load lifting vs. nonprogressive submaximal cycling	42	8.8[a]	11.6

[a] Significant by paired t test, $p < 0.001$.

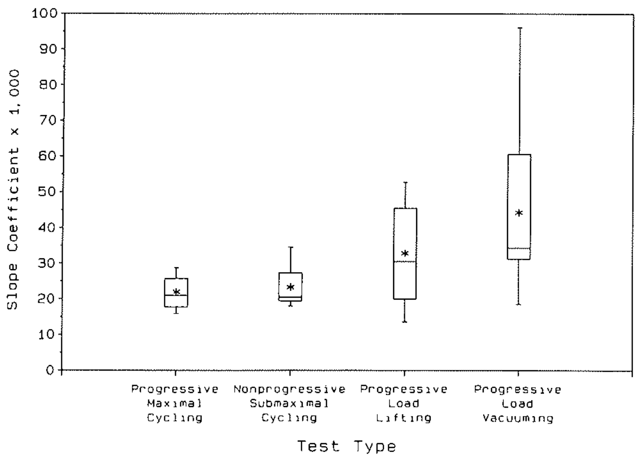


Figure 4. Distribution of the slope coefficients for 15 healthy men from regressions of natural log transformed $\dot{V}_E$ on HR. The box plots show the 25th and 75th percentiles as the bottom and top of the boxes, respectively. The medians and means are indicated by the horizontal lines and asterisks, respectively, in the boxes. The vertical lines show the range of the data to 1.5 times with interquartile range.

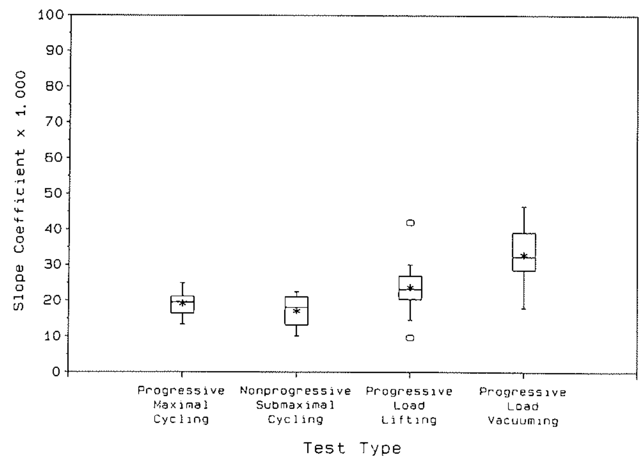


Figure 6. Distribution of the slope coefficients for 10 boys and girls aged 7 to 17 years from regressions of natural log transformed $\dot{V}_E$ on HR. The box plots show the 25th and 75th percentiles as the bottom and top of the boxes, respectively. The medians and means are indicated by the horizontal lines and asterisks, respectively, in the boxes. The vertical lines show the range of the data to 1.5 times the interquartile range and the circles represent outliers beyond this range.

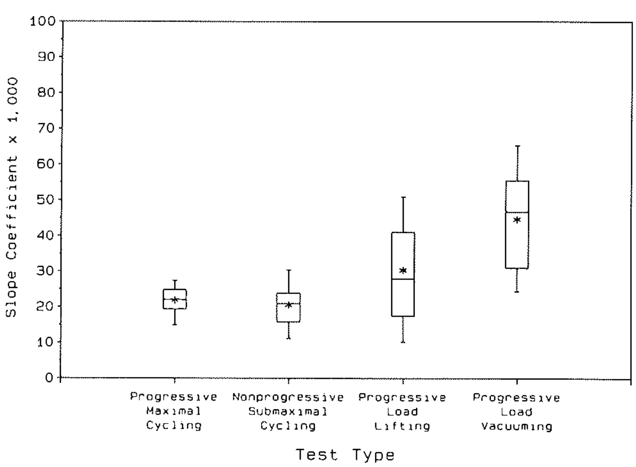


Figure 5. Distribution of the slope coefficients for 15 healthy women from regressions of natural log transformed $\dot{V}_E$ on HR. The box plots show the 25th and 75th percentiles as the bottom and top of the boxes, respectively. The medians and means are indicated by the horizontal lines and asterisks, respectively, in the boxes. The vertical lines show the range of the data to 1.5 times the interquartile range.

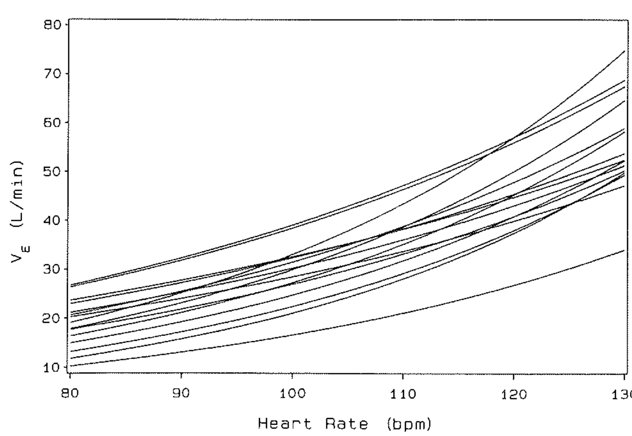


Figure 7. Plot of fitted regression lines of $\dot{V}_E$ on HR for 15 individual healthy men performing cycle exercise to maximal capacity.

Table 6. Regression Slopes for Estimated Minute Ventilation Values Based on Ventilation–Heart Rate Relations Derived from Different Pairs of Exercise Tests

Heart Rate (bpm)	Progressive Maximal Cycling vs. Nonprogressive Submaximal Cycling		Progressive Maximal Cycling vs. Progressive Load Lifting		Progressive Load Vacuuming vs. Progressive Load Lifting		Progressive Maximal Cycling vs. Progressive Load Vacuuming	
	Slope	R^2	Slope	R^2	Slope	R^2	Slope	R^2
60	0.93	0.59	0.13	0.00	0.77	0.17	0.22	0.27
80	0.98	0.61	0.29	0.08	0.98	0.26	0.29	0.36
100	0.93	0.58	0.49	0.30	1.25	0.45	0.31	0.42
120	0.75	0.49	0.44	0.44	1.18	0.50	0.23	0.33
140	0.59	0.43	0.32	0.44	1.01	0.43	0.15	0.22

tially higher for standing at rest before exercise than for sitting measurements (Table 7).

NASAL AIRWAY RESISTANCE

The values of NAR were similar on the two test days (Table 8). The median NAR for all subjects from both test days was 1.15 cm H_2O/L/sec. Nasal airway resistance decreased with age, but did not vary with other demographic and health characteristics of the subjects (Table 9).

Table 7. Comparison of Resting Minute Ventilation and Heart Rate Measurements Between Sitting and Standing Postures[a]

Variable[b]	Mean	SD	Minimum	Maximum
$\dot{V}_E$ Basal − $\dot{V}_E$ Vac	−1.7	2.7	−8.9	2.1
HR Basal − HR Vac	−8.9	10.2	−29.4	10.1
$\dot{V}_E$ Basal − $\dot{V}_E$ Lift	−2.4	2.1	−8.4	0.7
HR Basal − HR Lift	−11.2	10.9	−34.7	0.6

[a] n = 21 subjects.
[b] $\dot{V}_E$ is measured in L/min. HR is measured in bpm. Basal = sitting at rest; Vac = standing at rest before vacuuming test; Lift = standing at rest before lifting test.

Table 8. Nasal Airway Resistance for Test Days 1 and 2[a]

	Nasal Airway Resistance (cm H_2O/L/sec)	
	Day 1 Mean	Day 2 Mean
Mean	1.46	1.61
SD	1.09	1.54
Median	1.09	1.15
IQR	0.94	1.03
Minimum	0.24	0.24
Maximum	5.41	9.62

[a] n = 56 subjects.

We also did not find significant associations of NAR with current symptom status or with past history of nasal disease (Tables 10 and 11). For 14 of the 25 questions (primarily those on nasal surgery, nasal polyps, and the use of medication), too few positive responses were recorded to allow for adequate analysis.

We also assessed the variability in NAR. Within-subject variability was measured by examining changes in NAR and its SD between the two test days. The average individual difference between NAR on the two test days was 0.15 ± 1.23 cm H_2O/L/sec. The individual average percentage change in the NAR between test day 1 and test day 2 was 6%. The individual median SD of NAR was not significantly different between test day 1 and test day 2 (by Kruskal-Wallis test, $p = 0.21$).

Between-subject variability was examined by determin-

Table 9. Nasal Airway Resistance by Subject Characteristics

Characteristic	n	Median (IQR) (cm H_2O/L/sec)
Gender		
Male	32	1.09 (0.79)
Female	24	1.30 (1.22)
Age (years)		
7–13	11	1.77 (1.74)
16–60	28	1.13 (0.86)
61–72	18	0.87 (0.60)[a]
Health status		
Normal	41	1.16 (0.98)
Asthma	5	1.88 (1.30)
Chronic obstructive pulmonary disease	5	0.85 (1.11)
Ischemic heart disease	5	0.81 (0.59)
Smoking status		
Nonsmoker	50	1.13 (1.02)
Smoker	6	1.26 (0.87)

[a] p = 0.007 (Kruskal-Wallis test).

Table 10. Nasal Airway Resistance by Upper Respiratory Symptoms

Symptom	Day 1 n	Day 1 Median (IQR) (cm H_2O/L/sec)	Day 2 n	Day 2 Median (IQR) (cm H_2O/L/sec)
Cold in last week				
Yes	13	1.69 (1.19)	13	1.88 (1.29)
No	43	1.46 (1.21)	43	1.52 (1.41)
Nose congested				
Yes	15	1.98 (1.23)	9	1.35 (0.62)
No	41	1.33 (1.16)	47	1.65 (1.48)
Nasal discharge				
Yes	14	1.98 (1.30)	20	1.40 (0.90)
No	41[a]	1.37 (1.16)	36	1.72 (1.59)

[a] One missing response.

ing the difference in the SD of NAR on a single day between various subject groups. There was a significant decrease in the SD of NAR with increasing age (by Kruskal-Wallis test, $p = 0.03$) (James et al. 1993). There were no other significant differences in the between-subject SDs of NAR for the two test days by different subject characteristics. We also did not find significant differences in the SD of NAR between subjects from any of the symptom questions (James et al. 1993).

PARTITIONING OF BREATHING BETWEEN THE ORAL AND NASAL ROUTES

In past investigations, the partitioning of breathing between the oral and nasal routes has been described by identifying a "switch point," a level of ventilation at which breathing becomes predominantly oral. However, not all persons achieve a switch point and this simplistic approach reduces a complex and continuous pattern of changing ventilatory route into a simple dichotomous index (Figure 8). For example, Figure 8 shows the percentages of adult males

Table 11. Nasal Airway Resistance by Chronic Nasal Conditions

Nasal Condition	n	Median (IQR) (cm H_2O/L/sec)
Hay fever		
Yes	12	1.30 (1.18)
No	44	1.13 (0.92)
Sinus trouble		
Yes	24	1.04 (0.87)
No	32	1.24 (1.17)
Broken nose[a]		
Yes	8	1.19 (0.89)
No	47	1.11 (1.00)
Frequent nasal discharge		
Yes	24	1.09 (0.94)
No	32	1.16 (1.06)
Frequent nasal congestion		
Yes	18	1.16 (0.90)
No	37	1.11 (1.06)

[a] One missing response.

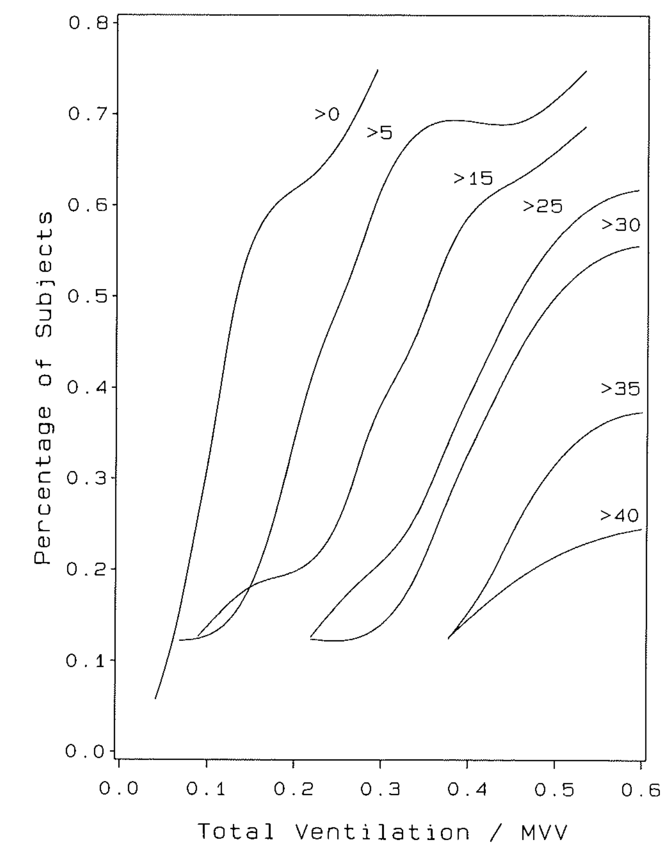


Figure 8. Orally inspired ventilation (V_I) at various levels of total V_I expressed as percentage of MVV during exercise for 12 healthy men. The curves represent the proportions of subjects whose oral V_I was 0, 5, 15, 25, 30, 35, or 40 L/min or greater. For example, 50% of subjects breathed at least 25 L/min through the oral route at 45% MVV, but only 25% of subjects breathed at least 35 L/min orally at this level of MVV.

Table 12. Pattern of Breathing at Rest and During Exercise

Resting $\dot{V}_E$	Exercise $\dot{V}_E$	n
Nasal[a]	Nasal[a]	6
Nasal[a]	Oronasal[b]	19
Oronasal[b]	Oronasal[b]	28
Oral[c]	Oral[c]	3

[a] Oral $\dot{V}_E$ less than 1 L/min, and nasal $\dot{V}_E$ greater than 1 L/min.
[b] Oral $\dot{V}_E$ greater than 1 L/min, and nasal $\dot{V}_E$ greater than 1 L/min.
[c] Oral $\dot{V}_E$ greater than 1 L/min, and nasal $\dot{V}_E$ less than 1 L/min.

achieving various levels of oral breathing relative to total ventilation. Thus, 40% of the men in the study breathed at least 15 L/min by mouth at a total ventilation corresponding to 30% of MVV. Accordingly, we estimated the proportion of ventilation by the oral route, as a percentage of MVV, and also characterized the overall pattern of breathing between the oral and nasal routes during the test.

Initially, subjects were categorized into four groups based on the overall pattern of ventilation during the test (Table 12). Most subjects had a mixed pattern of ventilation, but a few used the nasal or oral route exclusively. Subjects in the nasal only and oral only groups had lower MVVs than those with mixed patterns (Table 13). The three subjects in the oral-oral group were in the highest category of NAR.

Oral ventilation in relation to MVV was calculated for inspired ventilation up to 50% of MVV (Table 14 and Figure 9). The patterns were similar for adult males and females. As total ventilation increased, the percentage of oral flow increased progressively for adults, but flattened for children. Absolute flow rates at different levels of ventilation were estimated for children, men, and women (Figure 10). The projected linear increases of absolute flow rates were similar for men and women, as would be anticipated from the analysis in Figure 9; however, absolute flow rates increased more steeply with ventilation than did relative flow rates. Subjects with asthma had a high proportion of ventilation through the mouth, but there were only four subjects with asthma (Table 14). The pattern of oral breathing did not vary with different levels of fitness or with different nasal symptoms and conditions.

VALIDATION OF HEARTWATCH MEASUREMENTS

The recording accuracy of the Heartwatch was evaluated under controlled conditions in the laboratory and under field conditions. During exercise testing in the laboratory,

Table 13. Pattern of Breathing by Subject Characteristics

Group or Characteristic	n	Nasal-Nasal (n = 6)	Nasal-Oronasal (n = 19)	Oronasal-Oronasal (n = 28)	Oral-Oral (n = 3)
Healthy men	16	0	7	8	1
Healthy boys	5	1	1	2	1
Healthy women	15	3	6	6	0
Healthy girls	5	1	1	3	0
Subjects with asthma	5	0	2	2	1
Subjects with chronic obstructive pulmonary disease	5	1	1	3	0
Subjects with ischemic heart disease	5	0	1	4	0
Fitness[a]					
< 0.8	7	1	2	4	0
0.8–1.2	31	5	9	14	3
> 1.2	18	0	8	10	0
Maximum voluntary ventilation[b]					
Mean		74.5	128.5	129.1	80.0
SD		20.2	38.3	55.0	53.4
Minimum		53	61	39	43
Maximum		104	195	245	141
Nasal airway resistance[c]					
< 0.75	14	2	8	4	0
0.75–1.78	28	3	8	17	0
> 1.78	7	1	3	0	3

[a] Fitness measured by $\dot{V}_{O_2}$max/$\dot{V}_{O_2}$max predicted.
[b] Maximum voluntary ventilation measured in L/min.
[c] Nasal airway resistance measured in cm H_2O/L/sec.

Table 14. Proportion of Oral Breathing[a] During Exercise at Various Percentages of Maximum Voluntary Ventilation

Group or Characteristic	n	Mean Oral $\dot{V}_E$/Total $\dot{V}_E$					
		5% MVV	10% MVV	20% MVV	30% MVV	40% MVV	50% MVV
All subjects	51	0.14	0.24	0.35	0.41	0.46	0.50
Healthy men	16	0.16	0.19	0.35	0.48	0.56	0.61
Healthy boys	5	0.40	0.47	0.45	0.47	0.50	0.51
Healthy women	15	0.09	0.18	0.27	0.33	0.40	0.47
Healthy girls	5	0.00	0.21	0.39	0.34	0.32	0.31
Subjects with asthma	4	0.58	0.57	0.56	0.62	0.64	0.66
Subjects with chronic obstructive pulmonary disease	5	0.00	0.20	0.33	0.38	0.44	0.51
Subjects with ischemic heart disease	5	0.04	0.25	0.41	0.47	0.50	0.52
Fitness[a]							
< 0.8	7	0.00	0.13	0.28	0.34	0.40	0.46
0.8–1.2	30	0.18	0.28	0.38	0.42	0.46	0.47
> 1.2	14	0.14	0.22	0.31	0.42	0.51	0.60
Ventilation pattern							
Nasal-nasal	6	0.00	0.00	0.00	0.00	0.00	0.00
Nasal-oronasal	18	0.00	0.02	0.12	0.24	0.36	0.47
Oronasal-oronasal	24	0.19	0.39	0.53	0.56	0.59	0.60
Oral-oral	3	1.00	1.00	1.00	1.00	1.00	1.00
Cold in last week							
Yes	10	0.29	0.35	0.43	0.42	0.45	0.50
No	41	0.11	0.22	0.33	0.41	0.47	0.51
History of hay fever							
Yes	11	0.16	0.20	0.32	0.44	0.52	0.59
No	40	0.14	0.26	0.36	0.40	0.45	0.48

[a] Fitness measured by $\dot{V}_{O_2}$max/$\dot{V}_{O_2}$max predicted.

HRs from the Heartwatch were compared with those recorded by the Jaeger system. Fifty-five comparison observations were obtained for each of 10 subjects. Linear regression analysis indicated close agreement between the measurements of the two systems. The regression coefficients for individual subjects ranged from 0.96 to 1.05, and the R^2 ranged from 0.88 to 1.00 (Table 15). The linear regression on all observations combined was $HR_{Jaeger} = -1.07 + 1.01 \times HR_{Heartwatch}$ ($R^2 = 0.98$).

To assess the accuracy of the Heartwatch under condi-

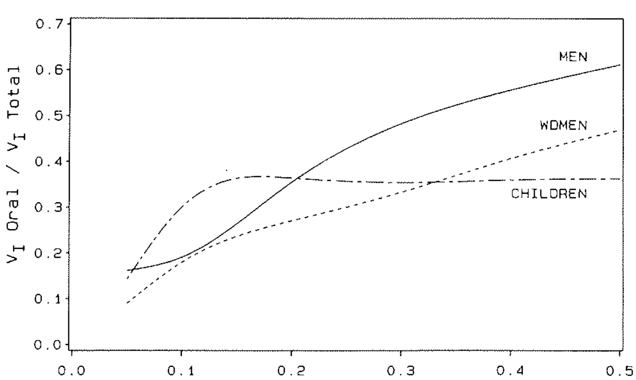


Figure 9. Fraction of orally inspired ventilation relative to total V_I expressed as percentage of MVV for healthy men, women, and children.

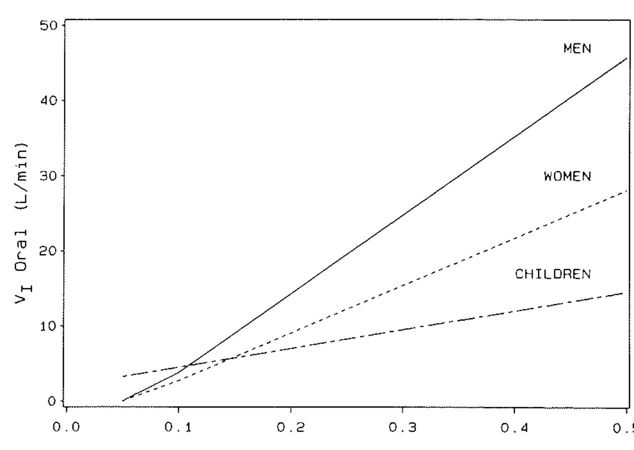


Figure 10. Orally inspired ventilation (V_I) at various levels of total V_I expressed as a percentage of MVV for healthy men, women, and children.

Table 15. Individual Regressions of Heartwatch on Jaeger System Heart Rates During Exercise Testing[a]

Subject Number	Slope	Intercept	R^2
1	0.98	2.13	0.98
2	0.99	1.00	1.00
3	0.99	0.57	0.99
4	0.97	2.76	0.97
5	1.00	0.10	0.97
6	0.96	3.59	0.98
7	0.96	3.59	1.00
8	0.97	3.18	1.00
9	1.05	- 7.77	0.88
10	1.01	1.52	0.96

[a] Values are based on 55 determinations per subject.

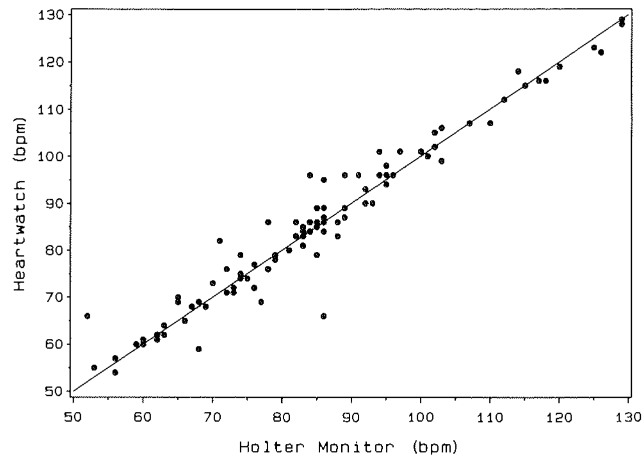


Figure 11. Scatterplot and linear regression for mean HRs (bpm) measured by the Holter monitors and Heartwatches.

tions typical of field monitoring, the Heartwatch was worn by 10 patients at the Veterans Administration Hospital during ambulatory ECG monitoring (Holter monitoring). During the comparison monitoring, the Heartwatch showed little loss of data except for recordings from one obese patient for whom electrode contact was lost intermittently. The quality of three of the Holter tracings was poor; however, reliable HR measurements could usually be obtained. Ten separate nonconsecutive minutes of monitoring data were selected randomly for each subject, yielding a total of 99 pairs of one-minute HRs for comparison.

The mean minute HRs measured by the Heartwatch and the Holter monitor were in agreement (Figure 11). No significant difference was observed in the mean HRs measured by the two systems (t = 1.18, p = 0.24). Eighty-eight of the 100 observations were within 4 bpm, and only four observations differed by more than 10 bpm. The difference in measured HR did not appear to be related to the magnitude of the HR, and linear regression analysis indicated good concordance ($HR_{Holter} = 3.04 + 0.97 \times HR_{Heartwatch}$; R^2 = 0.95).

AMBULATORY HEART RATE MONITORING

We attempted to obtain two 16-hour periods of HR and activity monitoring from each subject. However, fewer days of data were obtained because of monitor failure and subject noncompliance. Thus, 110 person-days of HR monitoring were collected from 57 subjects. Only one day of monitoring data could be obtained for four subjects.

One of the objectives of the research program was to evaluate the quality of the data collected with the Heartwatch and the activity diary. Typically, the Heartwatch was put on and started by the subject on the morning after visiting the laboratory for exercise testing. Each subject was shown how to apply the electrodes to his or her chest and how to attach the transmitter. Following a written set of instructions, the subject pressed buttons on the Heartwatch to begin recording the HR. Ten of 57 subjects experienced difficulties with starting the Heartwatch, and monitoring days had to be repeated. The sequence of buttons to be pushed is somewhat complex, and the display on the watch face is difficult for many people to understand. When the device was attached and started by a technician, no recording problems occurred. Loss of HR data also occurred on four occasions because buttons were pushed during the monitoring period, stopping the recording of data. Usually the pushing of buttons was accidental; however, some subjects pushed buttons when attempting to correct signal interference.

In general, the Heartwatch recordings were relatively free of artifact caused by poor contact of the electrodes with the skin and signal interference caused by strong FM sources (e.g., electric motors and video display terminals). Electrode sites were cleaned carefully and slightly abraded with an alcohol swab before electrodes were attached. In all but two instances, the transmitter remained securely attached to the subjects, even during intensive exercise and play. Transient loss of electrode contact due to movement, obesity, and undergarments resulted in a loss of signal and a 0 bpm measurement for the interval.

Signal interference caused by the proximity of FM sources was typically brief. Interference was usually registered as a high HR, and an alarm was sounded by the Heartwatch warning the wearer to move away from the source of interference. Artifact caused by signal interference is manifest as a nonsensical HR in excess of 200 bpm. Thus, HRs less than 40 bpm and greater than 200 bpm were considered to be artifact and were deleted from the time series. These records constituted less than 0.5% of the total monitoring time.

Table 16. Distribution of Heart Rates[a] by Age, Gender, and Health Status

Group	Mean	SD	Percentile 25	50	75	95
Healthy men	79.9	19.9	66	76	87	117
Healthy boys	92.3	21.3	80	91	103	127
Healthy women	82.0	16.2	71	80	90	110
Healthy girls	99.5	19.1	86	97	110	134
Subjects with asthma	87.0	21.2	74	83	95	127
Subjects with chronic obstructive pulmonary disease	88.7	14.7	77	87	99	114
Subjects with ischemic heart disease	85.0	11.2	77	84	92	104

[a] All HR values are given in bpm.

The activity diary was used properly by most subjects. However, two types of problems inherent with this type of written record were observed: nonsequential recording of activities and the omission of transitional activities (i.e., no record of car use between changing locations). These inconsistencies occurred in 22 of 110 diaries, and were attributable to 12 subjects. The impact of these inconsistencies was minimized by the project staff, who reviewed the diaries with the subjects and made corrections when the Heartwatches were retrieved.

The distribution of minute HRs is presented in Table 16. Children, in general, displayed higher HRs than adults. Heart rates for healthy adults were not different from HRs for adults with lung or heart disease.

Using diary information, HR measurements were grouped by class of activity and location. The distribution of HRs recorded during the 10 most frequently reported activities are presented in Table 17 for men, and Appendix A for women, boys and girls, and subjects with heart and lung disease. In general, lower HRs were associated with predominantly sedentary activities such as attending classes, household paperwork, and watching television. Higher HRs were associated with bicycling. When organized by location (Table 18; Appendix B), lower HRs were observed indoors (at home); higher HRs were observed outdoors. On the average, almost 50% of the monitored time was spent indoors (at home), and about 10% of the time was spent outdoors.

To estimate $\dot{V}_E$ from HR, we used the regression equations for each subject derived from measurements obtained during exercise reaching up to maximal work capacity on the cycle ergometer. This approach was used even though some activities involved upper body work and most activities were at much lower exertion rates than maximal capacity. The justification for this approach was based on the pilot monitoring of HR and activities. Upper body activities were infrequently reported (Table 17; Appendix A). Less than 5% of the ambulatory monitoring was for activities that could be classified as primarily involving the upper body (Table 19). Furthermore, HRs measured in the community setting usually ranged from 60 to 120 bpm (Table 17; Appendix A). Regressions from the full range of HRs were

Table 17. Distribution of Heart Rates for the Ten Most Frequent Activities Among Healthy Men[a]

Activity	*n*	Total Time (minutes)	Mean	SD	Percentile 25	50	75	95
Main job	5	1,867	74.3	13.3	64	74	83	95
TV viewing	10	1,852	67.6	12.8	59	66	76	90
Meals at home	15	1,657	76.2	12.8	67	76	83	98
Reading	8	1,316	71.0	9.1	64	71	77	86
Personal hygiene	14	888	80.0	12.4	72	78	88	100
Studying	2	691	76.8	11.7	68	77	85	96
Attending classes	1	641	67.0	15.2	55	62	78	94
Travel for goods	7	561	68.7	14.4	57	66	76	98
Household paperwork	4	505	66.7	9.0	62	66	72	84
Bicycling	6	495	127.1	23.2	112	131	145	157

[a] All HR values are given in bpm.

Table 18. Distribution of Heart Rates by Major Location Classes for Healthy Men[a]

Location	n	Total Time (minutes)	Mean	SD	Percentile 25	Percentile 50	Percentile 75	Percentile 95
Indoors, home	15	11,590	73.0	13.4	63	72	81	96
Indoors, work	5	2,187	77.9	12.3	69	78	86	98
Indoors, other	15	4,462	80.3	21.0	66	77	90	121
Outdoors	14	2,837	102.0	29.2	80	97	123	154
In transit	15	2,677	79.8	16.5	68	79	89	108

[a] All HR values are given in bpm.

used, rather than truncating the data to the heart range typical of most activities, i.e., less than 120 bpm. Using conventional protocols for exercise testing limited the number of observations at low levels of exercise. Comparison between $\dot{V}_E$ estimates based on the full range of HRs and $\dot{V}_E$ estimates based on HRs up to 120 bpm showed a small positive bias, on the average, for the regression based on the full HR data (Table 20). Some variation among the subjects was evident. Using each subject's $\dot{V}_E$-HR regression derived from the maximal exercise test, exhaled ventilation rates were estimated from the ambulatory HRs and organized by activity (Table 21; Appendix C) and location (Table 22; Appendix D).

As expected, the range of $\dot{V}_E$ and the pattern of relative ventilation across the various classes of activities and locations was similar to that observed for HR. The lowest estimated ventilation rates were associated with home activities including reading and television viewing. The highest estimates of ventilation were obtained for bicycling. To standardize the estimates of ventilation across individuals, ventilation also was calculated as a percentage of MVV and categorized by activities (Table 23; Appendix E) and locations (Table 24; Appendix F).

Table 19. Numbers of Subjects Who Reported Engaging in Activities Primarily Involving the Upper Body and the Total Time Spent in These Activities[a]

Activity	n	Total Time (minutes)
Meal preparation	41	2,001
Serving food, setting table, putting groceries away	11	280
Doing dishes	23	639
Meal cleanup	14	247
Work around the house	3	36
Indoor chores, washing windows, vacuuming	27	1,870
Washing clothes	8	580
Other clothes care	2	33
Indoor repairs, painting	4	598
Baby care	2	132

[a] Based on 110 16-hour monitoring periods obtained from 57 subjects.

DISCUSSION

This multifaceted project was designed to develop field-applicable methods for assessing total ventilation and route of ventilation. The need to refine the understanding of exposure-dose relations for inhaled pollutants provided a rationale for the investigation and for future application of the new methods in representative samples of the population. The methods examined included the use of HR to estimate ventilation, the measurement of breathing through the oral and nasal routes, and the feasibility of HR monitoring and activity recording in the field.

ESTIMATION OF VENTILATION USING HEART RATE

Pollutant doses to the lung are potentially affected by the amount of air inhaled and the partitioning of breathing between the oral and nasal routes. Thus, information on total

Table 20. Difference Between Minute Ventilation Estimates Based on Full Range of Heart Rates and Minute Ventilation Estimates Based Only on Heart Rates Less Than 120 bpm[a]

Heart Rate (bpm)	Estimated $\dot{V}_E$-HR(full) – $\dot{V}_E$-HR(60–120) (L/min) Mean	SD	Median	Minimum	Maximum
60	0.4	2.4	0.7	−7.6	4.2
70	0.4	2.3	0.7	−7.0	4.6
80	0.4	2.1	0.4	−6.1	4.8
90	0.2	1.7	0.3	−4.7	4.3
100	−0.2	1.1	−0.2	−2.8	2.8
110	−0.9	1.2	−0.6	−4.6	0.9
120	−2.0	3.1	−1.1	−9.3	3.5

[a] Heart rate data were obtained from 31 healthy men and women, aged 18 to 72 years, during the maximal cycling exercise test.

Table 21. Distribution of Estimated Minute Ventilation[a] for the Ten Most Frequent Activities Among Healthy Men

Activity	n	Total Time (minutes)	Mean	SD	Minute Ventilation (L/min) Percentile 25	50	75	95
Main job	5	1,867	17.8	10.5	13.1	16.8	20.9	27.0
TV viewing	10	1,852	14.5	6.4	9.1	14.4	18.0	25.0
Meals at home	15	1,657	18.7	7.6	14.0	18.6	21.9	30.2
Reading	8	1,316	13.1	5.0	8.3	13.0	17.4	21.1
Personal hygiene	14	888	19.2	7.9	12.6	18.6	23.4	34.2
Studying	2	691	22.4	4.5	18.7	22.2	25.5	30.2
Attending classes	1	641	19.9	5.3	15.9	17.8	22.9	29.6
Travel for goods	7	561	16.7	7.8	13.3	14.5	18.5	27.1
Household paperwork	4	505	10.5	3.5	7.7	9.4	13.2	17.2
Bicycling	6	495	65.0	29.5	40.8	63.8	87.8	113.0

[a] Minute ventilation was estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table 22. Distribution of Estimated Minute Ventilation[a] by Major Location Classes for Healthy Men

Location	n	Total Time (minutes)	Mean	SD	Minute Ventilation (L/min) Percentile 25	50	75	95
Indoors, home	15	11,590	16.6	7.3	11.6	16.3	20.5	28.2
Indoors, work	5	2,187	17.9	5.6	13.4	17.4	21.6	27.3
Indoors, other	15	4,462	22.7	15.3	14.2	19.0	26.4	46.2
Outdoors	14	2,837	36.7	26.7	18.7	27.3	44.6	96.9
In transit	15	2,677	20.0	12.2	13.5	17.8	23.3	35.8

[a] Minute ventilation was estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table 23. Distribution of Estimated Minute Ventilation[a] Expressed as Percentage of Maximum Voluntary Ventilation for the Ten Most Frequent Activities Among Healthy Men

Activity	n	Total Time (minutes)	Mean	SD	% Maximum Voluntary Ventilation Percentile 25	50	75	95
Main job	5	1,867	10.6	7.7	7.0	9.5	12.6	18.6
TV viewing	10	1,852	10.0	5.2	5.5	8.1	14.2	19.5
Meals at home	15	1,657	12.6	5.8	9.4	11.4	14.8	23.3
Reading	8	1,316	10.6	5.1	5.4	11.0	14.4	18.6
Personal hygiene	14	888	11.9	5.6	8.3	11.2	14.2	21.5
Studying	2	691	10.3	3.2	7.8	9.4	12.1	16.9
Attending classes	1	641	11.7	3.2	9.4	10.5	13.6	17.4
Travel for goods	7	561	12.8	5.8	10.2	10.9	13.6	26.5
Household paperwork	4	505	9.0	4.8	5.7	6.6	12.0	18.2
Bicycling	6	495	32.0	14.6	18.8	31.0	43.5	55.7

[a] Minute ventilation was estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table 24. Distribution of Estimated Minute Ventilation[a] Expressed as the Percentage of Maximum Voluntary Ventilation by Major Location Classes for Healthy Men

					% Maximum Voluntary Ventilation Percentile			
Location	*n*	Total Time (minutes)	Mean	SD	25	50	75	95
Indoors, home	15	11,590	11.0	5.7	7.0	10.5	13.5	21.1
Indoors, work	5	2,187	10.5	3.9	7.2	10.3	12.7	17.3
Indoors, other	15	4,462	13.9	9.2	9.1	11.9	16.1	28.9
Outdoors	14	2,837	19.1	13.2	9.9	14.4	23.8	47.6
In transit	15	2,677	13.4	8.0	9.3	12.3	16.0	26.7

[a] Minute ventilation was estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise tests for each individual.

ventilation and route of breathing might be utilized in epidemiologic studies to estimate the pollutant dose delivered to the lung and thereby address the dose-response relation rather than the exposure-response relation. If obtained from representative samples of the population, such information might also be incorporated into population models for predicting dose. Epidemiologic studies largely have addressed exposure and not dose, and limited data from clinical studies have been used in population exposure models to estimate lung doses of pollutants to urban populations (Johnson and Paul 1983; Paul and Johnson 1985; Ott et al. 1988; Wallace et al. 1988). These clinical data were obtained from small samples of subjects, and the types of exercise were limited to those that could be simulated in the laboratory setting. Consequently, there is growing recognition of the need for direct measurement of breathing patterns in ambulatory populations to improve the assessment of the risks of air pollution (Sexton and Ryan 1988). Ideally, the levels and patterns of ventilation would be measured on a population-based sample to obtain representative values for men and women, children, and susceptible groups during a wide range of urban activities.

To date, a simple but accurate method has not been available to measure ventilation directly in ambulatory subjects. For example, induction plethysmography can provide measurements of thoracic movement and $\dot{V}_E$ (Sackner et al. 1980). When carefully calibrated against conventional spirometry, this monitoring approach can provide accurate measurements of $\dot{V}_E$ in the laboratory; however, measurements of breathing patterns under active conditions are much less accurate because of interference from body movements and postural changes. New instrumentation that uses body displacement to estimate $\dot{V}_E$ has been described in recent years (Sackner et al. 1980; McCool et al. 1986). The instrumentation has not yet been miniaturized, and the monitoring approach may be too expensive to deploy on larger numbers of people. In another approach, the subject breathes through a face mask, and $\dot{V}_E$ and $\dot{V}_{O_2}$ are recorded to a data logger (Ikegami et al. 1988). This system seems too cumbersome to be worn for long periods and ventilation may be influenced by the face mask.

Heart rate monitoring represents a potentially feasible approach for estimating ventilation in the community setting. Heart rate is coupled to metabolic activity and oxygen consumption, and ambulatory monitoring of HR has been used to estimate energy expenditure (Astrand and Rodahl 1977; Harber et al. 1984). Although HR is influenced by factors other than physical exertion, the major determinant over averaging times of one minute or longer is oxygen consumption. The correlation between oxygen consumption and ventilation is also high; consequently, ventilation and HR are tightly associated.

In healthy persons, the $\dot{V}_E$-HR relationship is determined by the intensity of exercise, training status, posture, and the type of exercise. During graded exercise, ventilation increases in a curvilinear fashion with increasing work load, $\dot{V}_{O_2}$, and HR. Alternatively, the relation between ventilation and $\dot{V}_{O_2}$ can be depicted as biphasic, with a linear component at low to moderate exercise intensities, followed by an inflection above which ventilation increases exponentially with exercise intensity. The work load at the point of inflection is referred to as the "anaerobic threshold." Suprathreshold ventilation is stimulated by the combined effects of the metabolic demand of working muscle and the chemical stimulus of lactic acidosis.

Although the major determinant of the coupling of ventilation with HR is the intensity of exercise, other factors also affect the $\dot{V}_E$-HR relation. At any given work load, HR is lower in trained than in untrained subjects, in the supine position than in the upright posture (Sheldahl et al. 1984), and during leg exercise than during arm exercise. Supine posture and the trained state reduce the HR response to submaximal exercise by augmenting stroke volume. In upright arm exercise, both ventilation and HR are greater at a given submaximal work load than during leg exercise of the same intensity because of the lower muscle mass of the arms, the earlier onset of lactic acid production, and the reduced stroke volume with arm work, probably from reduced venous return on a gravitational basis (Freyschuss 1975; Martin et al. 1991; Toner et al. 1990). Moreover, the relation of work load to HR is identical in leg and arm work (Martin et al. 1991).

Most routine activities are at work levels well below the anaerobic threshold. Thus, HR can be used to estimate $\dot{V}_E$, particularly for the typically less strenuous activities of daily life. Small and relatively inexpensive instruments made for athletic training are now available for monitoring HR in the field setting. By combining HR recording with an activity diary, the level of ventilation associated with various activities can be determined.

The usefulness of HR recordings as a surrogate measurement of metabolic activity has been demonstrated in epidemiologic studies of physical activity and energy expenditure (Taylor et al. 1984; Patrick et al. 1986). Heart rate monitoring was first used as a surrogate measurement of ventilation and of lung dose of air pollutants by Raizenne and Spengler (1989). While attending a summer camp in southern Ontario, Canada, girls aged 7 to 14 years performed a standardized exercise test and individual $\dot{V}_E$-HR ratios were determined. Heart rates were monitored with a portable recording device during daily activities, which involved time outdoors and indoors. Ambient levels of ozone and acid aerosols were measured. Raizenne and Spengler (1989) then used a dosimetric model to calculate doses of ozone and acid aerosols for various time periods. This approach documented substantial variation among the children in estimated dose of acid aerosols delivered to the respiratory tract, even though the exposure was the same for all children. For some of the children, elevated estimates of ventilation and lung dose were observed with outdoor exercise at times of the day when ambient levels of ozone and acid aerosols were highest. These observations demonstrated the potential for moderate degrees of activity to yield acid aerosol doses in the range associated with adverse effects in controlled human exposures (Spengler et al. 1989).

Expanding on the approach of Raizenne and Spengler (1989), workers at the Rancho Los Amigos Medical Center (Shamoo et al. 1990) characterized the $\dot{V}_E$-HR relation for 20 men and women who worked outdoors in Los Angeles and hence were at risk for exposure to ambient ozone. Minute ventilation–on–heart rate regressions were developed for each subject from observations obtained during progressive exercise on a treadmill and during walking at various speeds on a level course. Ambulatory monitoring of HR was then performed, and the subjects maintained a written record of activities and exertion levels. The HRs and estimated $\dot{V}_E$s from the field monitoring have not yet been reported. However, this study demonstrated the superiority of HR recording over the perceived level of exertion as a surrogate measure of ventilation. Nine subjects underwent several training sessions to learn to recognize and classify their levels of ventilation. Relative to the estimate of $\dot{V}_E$ from the ambulatory HR recording, subjects tended to underestimate breathing rates in the intermediate classes of ventilation, and they sometimes completely omitted periods of elevated physical activity. This apparent inaccuracy in self-reporting, when viewed in the context of the often brief and transient nature of exertional activities that would have to be recorded in a written diary, suggests that the monitoring of HR would produce a more continuous and accurate record of ventilation.

McCool and Paek (1993) used the $\dot{V}_E$-HR relation, characterized during cycling, to predict $\dot{V}_E$ for nine subjects during classroom and workshop activities involving auto body repair. In the field, $\dot{V}_E$ was measured using body surface motion. The test protocol did not require a mouthpiece for measuring $\dot{V}_E$, thereby avoiding the associated upward bias.

McCool and Paek (1993) considered three different approaches for describing the $\dot{V}_E$-HR relation: linear regression based on data from the full range of HRs, linear regression based on data from the low range of HRs observed in the field, and exponential regression based on the full range of the data. The investigators noted that the subjects had lower HRs in the field than in the laboratory. The average maximum HR in the field was only 95 bpm, whereas the average was 153 bpm during cycling in the laboratory. Poor correlation between the measured ventilation and that predicted by the linear regression using the full data was attributed to the narrow ranges and low values of HRs in the field setting. In fact, negative values of $\dot{V}_E$ were predicted, apparently because the intercept of the regression model was unconstrained. Better correlations were found with the linear regression model based on the low range of HR and with the exponential model. The average errors were only about 20% and the group means did not show a pattern of upward or downward bias. Downward bias would have been anticipated because of the use of the $\dot{V}_E$-HR relation from lower body exercise to predict $\dot{V}_E$ during work involving upper and lower body movement.

In this project, we assessed the effect of the type of exercise load (lower body versus upper body) and the sequence of load presentation (progressive versus nonprogressive order of presentation) on the $\dot{V}_E$-HR relation (Table 4; Figures 2 through 6). Although the order of presentation did not affect this relation, ventilation increased more steeply with upper body exercise than with lower body exercise (Tables 5 and 6; Figures 2 through 6). The differing responses to upper versus lower body exercise, well described in the literature on work physiology (Astrand and Rodahl 1977), imply that accurate prediction of ventilation using the $\dot{V}_E$-HR relation must incorporate regression slopes appropriate for the activity. We recommend that a minimum protocol for using the $\dot{V}_E$-HR relation in the community setting would include two exercise tests, one involving upper body work and a second involving lower body work. We documented substantial variability among individuals in the $\dot{V}_E$-HR relation (Figures 4 through 6). This variability implies that exercise testing would be needed to describe the individual $\dot{V}_E$-HR relations for subjects in field studies, assuming an average relation for groups of subjects would not suffice.

BREATHING ROUTE

In addition to total ventilation, the breathing route is a potentially important determinant of lung doses of inhaled pollutants. To date, the breathing route has been studied in the laboratory, but only with a conventional exercise challenge (Niinimaa et al. 1980, 1981). Methods for assessing the oronasal distribution of ventilation have not been developed yet for the field setting. Predictors of the pattern of distribution of ventilation between the oral and nasal routes also have not been identified. If nasal symptoms or NAR were associated with the pattern of partitioning, then models could be developed to predict breathing route during usual activities.

The distribution of ventilation between the oral and nasal routes may depend on the presence of nasal diseases or pathology, or of lung diseases. Allergic rhinitis (hay fever) is common in both children and adults. Asthma and chronic obstructive pulmonary disease are also common conditions. Approximately 8% of adults and children have asthma (Evans et al. 1987; Gergen et al. 1988), and from 5% to 10% of adults have chronic obstructive pulmonary disease (U.S. Department of Health and Human Services 1984). Limited evidence indicates that the presence of respiratory disease may influence oronasal selection (Chadha et al. 1987).

In this study, we administered a questionnaire on nasal symptoms and measured NAR before the assessment of partitioning of breathing between the oral and nasal routes. We hypothesized that symptoms and NAR at the time of testing would predict the pattern of partitioning.

We investigated the variability of measurements of NAR by posterior rhinomanometry and standardized the procedure in our laboratory. In the initial phase of establishing the technique in the laboratory, we considered the flow at which NAR should be estimated and the number of curves to be obtained. A decision was made to estimate NAR at a flow of 0.25 L/sec, a level reached by 96% of the subjects. Analyses of data from eight subjects asked to produce 10 tracings showed that five satisfactory tracings would provide a sufficiently precise estimate of NAR for most subjects.

The median NAR measured in this study, 1.15 cm H_2O/L/sec, is in the range of values obtained in other studies with posterior rhinomanometry: 1.04 to 2.48 cm H_2O/L/sec (Ingelstedt et al. 1969; McLean et al. 1976; Cockcroft et al. 1979; Cole et al. 1980; Dvoracek et al. 1985). Each of these studies used different flow or pressure values to record NAR. However, the mean SD in our study, 1.33 cm H_2O/L/sec, was higher than the values reported in the other studies. The higher SD may reflect the wide age range of subjects in our study, as well as the inclusion of both healthy subjects and subjects with heart and respiratory diseases. However, the difference in NAR for comparing the two days of testing was small for most subjects. Thus, for most individuals, we achieved highly reproducible measurements of NAR.

Nevertheless, we did not find significant associations between the presence of nasal symptoms or conditions and the level of NAR (Tables 10 and 11). Subjects completed 25 questions on upper airway health and symptoms. The median NAR was frequently larger when the nasal symptom was present than when it was absent; however, because of the large between-subject variability and the limited sample size, these differences were not statistically significant (Table 10). Other studies have shown modest associations of symptoms with NAR. Gleeson and coworkers (1986) compared posterior rhinomanometry with subjective assessment of nasal patency and found a moderate correlation (R^2 = 0.47). McCaffery and Kern (1979) used anterior rhinomanometry to study 974 subjects and found that moderate and severe subjective symptoms of nasal obstruction were associated with greater NAR. Symptoms of nasal obstruction occurred primarily at total NAR above 3 cm H_2O/L/sec. Cole and coworkers (1980) found higher NAR using posterior rhinomanometry in subjects with rhinitis or fixed obstruction than in normal subjects.

Our methodology for describing route of breathing in a continuous rather than a discrete fashion differed from previous studies. Methodology for characterizing route of breathing has not been standardized; switch points from nasal to predominantly oral routes have been determined without specific criteria (Saibene et al. 1978; Niinimaa et al. 1980, 1981). Additionally, the subjects in most previous studies generally have been young, healthy volunteers (Saibene et al. 1978; Niinimaa et al. 1980, 1981). By contrast, we recruited subjects across a broad age range and included a small number of subjects with heart or lung disease (Table 1).

As in other studies, we observed that some patients have a clear switch point (Saibene et al. 1978; Niinimaa et al. 1980, 1981); however, only a minority of subjects had this pattern (Table 12). The most common pattern was oronasal breathing at rest with augmented oral breathing during exercise. By contrast, in the study of 30 subjects by Niinimaa and coworkers (1980, 1981), 67% breathed nasally at rest and oronasally with exercise. Of the remainder, 13% breathed oronasally at rest and with exercise, and 17% breathed only nasally. One subject had an irregular pattern of ventilation.

The assessment of predictors of breathing route was limited by the number of subjects. However, several associations of potential biologic significance were identified (Table 13). Subjects with asthma had a higher level of oral ventilation at rest and little increase in the proportion of oral breathing with exercise. Chadha and coworkers (1987) described the route of breathing in six asymptomatic subjects with asthma. In comparison with control subjects, subjects with asthma also had a higher average proportion of oral breathing at rest (45% versus 14%). With exercise, the proportion of oral breathing increased further in the study of Chadha and associates (1987). We also found that subjects in the upper quartile of NAR had a greater proportion of oral breathing. This pattern persisted when subjects with asthma were excluded. Chadha and colleagues (1987) reported that NAR and the percentages of nasal ventilation at rest and during exercise were not associated. However, this lack of association must be interpreted in the context of a total sample size of only 18 subjects.

HEART RATES AND ESTIMATED MINUTE VENTILATION DURING DAILY ACTIVITIES

Heart rate is known to be coupled to work load in healthy persons. An extensive literature addresses the energy costs of various activities (Astrand and Rodahl 1977), but information on distributions of HRs during daily activities has not been reported. The present pilot project demonstrates that field monitoring in combination with the recording of activities can describe the HRs during activities of sufficient duration. Ventilation associated with activity in the field setting was estimated using regression equations derived for each subject from laboratory testing.

We used the individual $\dot{V}_E$-HR relations to estimate ventilation during daily activities (Tables 21 through 24; Appendices C through F). The patterns of ventilation estimated for various locations and activities seemed appropriate for the work loads anticipated. For example, the highest $\dot{V}_E$s were found during bicycling, and the lowest were found during the most sedentary activities. This consistency of ventilatory patterns with the work load imposed supports the validity of the estimates of $\dot{V}_E$. However, we did not validate the use of the $\dot{V}_E$-HR relation using unencumbering technology such as magnetometry.

Only limited data are available on $\dot{V}_E$ in the community setting, and none have been reported in a fashion allowing direct comparison to the present study. Using magnetometers, McCool and colleagues (1986) estimated ventilation in the laboratory setting for six subjects who performed simulated household tasks. In subsequent work by McCool and Paek (1993), ventilation was measured for nine young males during approximately one hour of light industrial activity. Shamoo and associates (1990) described estimated ventilation for activities based on methods similar to the present study; full details have not yet been reported.

Although the present investigation was initiated for the purpose of developing methodology, the HR and activity monitoring add substantially to the available data on levels of activity in the community. The study showed surprisingly narrow distributions of HRs and associated estimates of $\dot{V}_E$ (Table 16). During almost all activities, except for exercise, HR values were below 100, with predicted levels of ventilation infrequently above 25 L/min.

LIMITATIONS

This project was intended to develop methodology and assess its potential for field use in a pilot study in the community setting. The laboratory methods were conventional and previously well characterized. Our use of a mouthpiece for measuring ventilation may have biased results toward larger values. $\dot{V}_E$ increases more during breathing through a mouthpiece than with unencumbered breathing (Gilbert et al. 1972; Askanazi et al. 1980). McCool and Paek (1993) directly addressed the effect of breathing through a mouthpiece on the $\dot{V}_E$-HR relation. The regression coefficients describing the incremental changes in $\dot{V}_E$ with HR were similar for data obtained with and without a mouthpiece. McCool and Paek (1991) suggested that the effect of the mouthpiece may be to shift the $\dot{V}_E$ by a relatively constant absolute amount across the range of HRs.

The partitioning of breathing between the oral and nasal routes was assessed using a face mask. The mask may have influenced ventilation by increasing respiratory dead space or through sensory stimulation (Askanazi et al. 1980). In supine subjects at rest, Askanazi and associates (1980) found that a mask with a dead space of approximately 50 mL increased ventilation by about 30%. Hirsch and Bishop (1982) assessed patterns of breathing through a face mask or a mouthpiece with a noseclip in place during at-rest breathing of air, or breathing with chemostimulation induced by inhaling air with elevated carbon dioxide or decreased oxygen. The changes in patterns of breathing during chemostimulation were different for the two modalities. The study did not compare these results to a noninvasive method for assessing ventilation. Nevertheless, the findings of these two studies suggest that our use of a face mask for assessing partitioning probably did affect breathing patterns; however, the effect of using a mask on the pattern of partitioning is uncertain.

In retrospect, the protocol did not include sufficient measurements after the establishment of basal conditions for the initial group of subjects, nor were sufficient data collected at low levels of exertion. We make this judgment in light of the pulse rate distributions obtained during the Heartwatch monitoring, which showed that most HRs were below 100 (Table 16). McCool and Paek (1993) also comment on the need for laboratory assessment of the $\dot{V}_E$-HR relation in the range of HR relevant to the community setting. To make future protocols for determining the $\dot{V}_E$-HR relation more relevant for typical activity patterns, we suggest that the progression of work load should be less steep initially.

With regard to feasibility, most subjects completed the full protocol. The Heartwatch provided valid data for almost all of the time monitored and proved to be readily handled by most subjects. The time-activity diaries required conscientious and timely completion during the day. In spite of having well-motivated volunteer subjects, inconsistencies occurred in 20% of the diaries.

The methods reported here could be refined further by categorizing activities as upper or lower body and by using the appropriate $\dot{V}_E$ relation to estimate ventilation. Oral and nasal ventilation could also be estimated using the relation between oral breathing and total ventilation obtained in the laboratory. We have developed an approach for describing oral breathing in a continuous fashion; however, additional validation is needed.

CONCLUSIONS

Data from this multicomponent project support the following conclusions in relation to the primary research objectives:

1. The $\dot{V}_E$-HR relation was found to be different for lower body and upper body exercise; the regression coefficient describing the increase of ventilation with HR was about 30% greater with upper body exercise.
2. The $\dot{V}_E$-HR relations were identical for cycling exercise test protocols with progressively increasing work load and nonprogressive presentation of work loads.
3. Substantial variation among individuals was documented in the $\dot{V}_E$-HR relation.
4. The $\dot{V}_E$-HR relation did not vary greatly across categories of gender and age, but numbers were limited within particular strata.

In the field component, we established the feasibility of using the Heartwatch to record HRs; in combination with the time and activity recording, HRs could be assigned to specific activities and ventilation estimated. The two days of HR recording showed that HRs infrequently exceeded 120 bpm. Most periods with high HRs were associated with reports of physical activity.

Conventional protocols used in exercise testing provide few measurements at low work loads. We suggest that protocols yielding more precise estimates of the regression slope of ventilation on HR are needed at low HRs. Such protocols should provide sufficient time at rest to establish a basal state and should increase the work load more slowly than in the usual protocols. Although the $\dot{V}_E$-HR relation is different for upper and lower body exercise, the impact of using separate regression slopes for the two forms of exercise may not be great in the community setting. In our limited data base of activities, predominantly upper body work was infrequent. However, some occupations may involve sustained and vigorous upper body activity.

Although our subjects were not selected at random and their activities may not be representative of the full spectrum of activities in the community, the range of measured HRs was surprisingly narrow. The subjects were volunteers, and many exercised frequently for fitness. Nevertheless, few daily activities other than exercise were associated with high HRs. It is possible that nonoccupational activities that substantially affect air pollutant doses could be ascertained with questions directed toward subjects' exercise frequency and level.

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APPENDIX A. Measured Ambulatory Heart Rates by Type of Activity for Women, Children, and Subjects with Heart and Lung Disease

Table A.1. Distribution of Heart Rates for the Ten Most Frequent Activities Among Healthy Women

					Heart Rate (bpm) Percentile			
Activity	*n*	Total Time (minutes)	Mean	SD	25	50	75	95
Main job	6	3,600	84.2	12.6	76	84	93	105
TV viewing	11	1,964	74.4	11.8	67	72	81	93
Meals at home	13	1,253	77.7	10.6	70	76	85	96
Meal preparation and cooking	13	1,035	82.3	12.0	74	83	90	103
Personal hygiene	14	710	86.5	14.6	75	85	96	113
Naps and resting	7	699	71.7	10.0	65	71	76	92
Reading	5	634	75.3	10.0	69	75	80	91
Other personal travel	11	549	80.3	14.8	71	79	87	104
Travel to and from work	6	470	84.8	16.2	74	85	94	113
Studying	3	460	73.6	8.7	68	71	78	91

Table A.2. Distribution of Heart Rates for the Ten Most Frequent Activities Among Healthy Boys

					Heart Rate (bpm) Percentile			
Activity	*n*	Total Time (minutes)	Mean	SD	25	50	75	95
TV viewing	5	929	86.1	21.2	72	89	101	120
Routine indoor chores	4	561	93.1	18.4	78	90	104	129
Attending classes	2	537	84.4	19.8	71	81	95	117
Meals at home	6	518	89.8	13.5	82	90	98	111
Computer games	2	462	89.4	11.5	82	88	95	109
Naps and resting	3	403	74.3	17.1	60	72	86	105
Shopping	5	306	100.4	12.6	93	100	106	123
Meal preparation and cooking	4	277	88.6	16.1	79	88	100	114
Studying	2	247	80.2	15.3	72	80	88	107
Outdoor play	2	238	115.0	16.4	103	114	126	145

Table A.3. Distribution of Heart Rates for the Ten Most Frequent Activities Among Healthy Girls

Activity	n	Total Time (minutes)	Mean	SD	Heart Rate (bpm) Percentile 25	50	75	95
Attending classes	4	1,275	98.6	15.7	87	98	109	126
TV viewing	6	1,213	93.0	18.3	80	91	103	127
Meals at home	6	919	94.9	14.9	85	95	104	119
Outdoor play	3	555	117.5	22.0	103	111	126	166
Personal hygiene	6	428	97.8	14.7	86	96	108	124
Routine indoor chores	5	323	110.7	15.0	102	112	121	132
Bicycling	3	284	126.8	20.5	110	129	144	158
Shopping	4	278	96.8	12.7	87	96	104	118
Racquet sports	1	234	98.5	17.5	84	99	113	125
Other passive leisure	1	227	94.4	9.1	88	94	101	110

Table A.4. Distribution of Heart Rates for the Ten Most Frequent Activities Among Women with Asthma

Activity	n	Total Time (minutes)	Mean	SD	Heart Rate (bpm) Percentile 25	50	75	95
Shopping	3	352	90.6	24.5	75	87	96	163
Meals at home	4	340	80.1	9.6	74	80	86	98
Errands	2	338	82.2	17.3	69	81	93	114
Paid work	1	329	70.2	9.0	64	68	76	89
Travel for goods	3	286	84.7	24.3	72	79	89	114
Routine indoor chores	2	285	84.7	8.4	78	85	91	98
Reading	1	244	68.1	9.5	62	65	70	92
Other household chores	1	220	79.6	7.4	74	80	85	91
TV viewing	3	219	73.6	11.7	64	71	82	94
Other personal travel	3	209	89.1	24.3	73	81	99	142

Table A.5. Distribution of Heart Rates for the Ten Most Frequent Activities Among Men with Chronic Obstructive Pulmonary Disease

Activity	n	Total Time (minutes)	Mean	SD	Heart Rate (bpm) Percentile 25	50	75	95
TV viewing	3	1,001	83.1	10.2	76	84	90	98
Relaxing	4	922	87.7	14.1	76	85	99	111
Other passive leisure	3	675	74.1	7.6	69	74	79	86
Paid work	2	662	99.1	12.1	90	99	108	120
Meals at home	5	599	91.8	10.9	84	90	100	111
Naps and resting	3	530	73.4	7.9	68	71	76	89
Personal travel	3	417	94.5	10.5	87	92	101	115
Travel for goods	4	384	91.6	16.8	79	86	103	128
Reading	3	364	84.1	12.2	73	86	95	101
Household repairs	1	264	107.7	8.1	102	108	112	121

Table A.6. Distribution of Heart Rates for the Ten Most Frequent Activities Among Men with Ischemic Heart Disease

Activity	*n*	Total Time (minutes)	Mean	SD	Heart Rate (bpm) Percentile 25	50	75	95
TV viewing	4	1,048	81.7	8.5	76	82	88	95
Meals at home	5	771	83.3	8.9	77	83	89	98
Other passive leisure	5	576	85.2	9.5	78	85	91	103
Socializing	3	429	86.9	10.0	79	86	94	104
Travel for goods	4	425	84.4	14.1	74	82	93	110
Main job	1	379	94.6	5.5	91	95	98	104
Indoor repairs	2	353	83.5	4.5	81	83	86	89
Naps and resting	3	304	74.8	8.9	67	72	81	89
Relaxing	4	301	80.7	10.2	75	81	85	93
Meals at restaurants	3	249	91.2	10.8	83	88	98	112

APPENDIX B. Measured Ambulatory Heart Rates by Location Class for Women, Children, and Subjects with Heart and Lung Disease

Table B.1. Distribution of Heart Rates by Major Location Class for Healthy Women

Location	*n*	Total Time (minutes)	Mean	SD	Heart Rate (bpm) Percentile 25	50	75	95
Indoors, home	14	10,405	78.3	15.0	68	76	86	103
Indoors, work	6	3,600	82.8	12.0	75	83	90	103
Indoors, other	14	4,079	83.6	17.0	73	82	93	110
Outdoors	13	1,339	99.6	22.4	83	95	116	140
In transit	15	2,670	83.9	16.1	73	82	92	111

Table B.2. Distribution of Heart Rates by Major Location Class for Healthy Boys

Location	*n*	Total Time (minutes)	Mean	SD	Heart Rate (bpm) Percentile 25	50	75	95
Indoors, home	6	4,625	87.0	18.0	76	87	98	117
Indoors, other	5	1,415	90.1	18.1	79	90	101	117
Outdoors	6	1,521	109.0	25.0	94	105	119	157
In transit	6	734	95.5	17.1	85	93	104	126

Table B.3. Distribution of Heart Rates by Major Location Class for Healthy Girls

Location	n	Total Time (minutes)	Mean	SD	Heart Rate (bpm) Percentile			
					25	50	75	95
Indoors, home	6	4,181	96.7	16.9	85	96	107	126
Indoors, other	6	3,153	97.2	16.6	85	95	107	127
Outdoors	6	1,315	116.1	24.0	100	113	129	163
In transit	6	994	96.7	17.1	86	94	104	127

Table B.4. Distribution of Heart Rates by Major Location Class for Women with Asthma

Location	n	Total Time (minutes)	Mean	SD	Heart Rate (bpm) Percentile			
					25	50	75	95
Indoors, home	5	1,917	79.1	11.8	71	79	87	97
Indoors, work	1	329	70.2	9.0	64	68	76	89
Indoors, other	5	1,977	87.8	24.4	73	82	95	149
Outdoors	5	214	112.4	34.1	85	99	155	169
In transit	5	1,255	85.7	21.1	73	82	92	117

Table B.5. Distribution of Heart Rates by Major Location Class for Men with Chronic Obstructive Pulmonary Disease

Location	n	Total Time (minutes)	Mean	SD	Heart Rate (bpm) Percentile			
					25	50	75	95
Indoors, home	5	4,285	86.1	14.3	74	85	95	112
Indoors, work	1	441	96.4	12.7	87	94	104	121
Indoors, other	5	880	90.3	12.6	82	89	99	115
Outdoors	5	1,126	86.3	16.0	75	82	101	113
In transit	5	1,365	95.4	13.5	85	95	105	118

Table B.6. Distribution of Heart Rates by Major Location Classes for Men with Ischemic Heart Disease

Location	n	Total Time (minutes)	Mean	SD	Heart Rate (bpm) Percentile			
					25	50	75	95
Indoors, home	5	4,046	83.0	9.1	77	83	89	98
Indoors, work	1	128	91.7	5.0	89	92	95	100
Indoors, other	5	1,427	84.2	12.4	75	83	91	106
Outdoors	5	1,467	89.1	13.0	81	89	97	112
In transit	5	1,008	88.0	13.0	77	87	96	108

APPENDIX C. Estimated Minute Ventilation (L/min) by Type of Activity for Women, Children, and Subjects with Heart and Lung Disease

Table C.1. Distribution of Estimated Minute Ventilation[a] for the Ten Most Frequent Activities Among Healthy Women

Activity	n	Total Time (minutes)	Mean	SD	Minute Ventilation (L/min) Percentile 25	50	75	95
Main job	6	3,600	14.5	5.1	10.8	13.6	17.5	23.8
TV viewing	11	1,964	9.5	5.1	7.1	8.4	11.4	15.1
Meals at home	13	1,253	12.3	3.6	9.3	12.2	14.6	18.8
Meal preparation and cooking	13	1,035	12.7	3.9	9.3	12.5	15.4	19.6
Personal hygiene	14	710	14.4	5.3	10.2	13.5	17.1	23.7
Naps and resting	7	699	9.9	3.4	7.2	9.5	12.5	15.4
Reading	5	634	10.6	4.2	7.4	9.2	13.9	17.4
Other personal travel	11	549	13.6	7.5	9.4	13.2	15.9	21.5
Travel to and from work	6	470	14.2	6.1	10.4	13.0	16.4	23.3
Studying	3	460	12.7	14.1	8.9	12.4	15.1	20.8

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table C.2. Distribution of Estimated Minute Ventilation[a] for the Ten Most Frequent Activities Among Healthy Boys

Activity	n	Total Time (minutes)	Mean	SD	Minute Ventilation (L/min) Percentile 25	50	75	95
TV viewing	5	929	13.0	4.6	9.1	12.8	15.8	21.2
Routine indoor chores	4	561	13.8	6.3	9.3	11.8	16.5	27.0
Attending classes	2	537	12.0	5.3	8.9	11.3	13.6	17.9
Meals at home	6	518	12.9	3.6	10.6	12.5	14.8	19.0
Computer games	2	462	14.1	3.9	12.2	13.4	15.3	19.2
Naps and resting	3	403	11.2	4.0	8.6	9.5	13.6	19.0
Shopping	5	306	16.4	5.7	14.5	16.2	18.6	23.3
Meal preparation and cooking	4	277	13.3	3.6	10.7	12.5	15.5	19.7
Studying	2	247	11.8	3.4	9.7	10.9	13.0	18.1
Outdoor play	2	238	17.6	6.1	14.0	16.1	19.6	28.4

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table C.3. Distribution of Estimated Minute Ventilation[a] for the Ten Most Frequent Activities Among Healthy Girls

Activity	n	Total Time (minutes)	Mean	SD	Minute Ventilation (L/min) Percentile 25	50	75	95
Attending classes	4	1,275	12.9	3.5	10.7	12.3	14.2	19.0
TV viewing	6	1,213	10.8	4.2	8.1	9.5	12.3	18.9
Meals at home	6	919	11.9	4.1	9.1	10.7	13.6	20.5
Outdoor play	3	555	14.4	7.1	10.2	12.4	15.3	30.1
Personal hygiene	6	428	16.1	4.8	12.4	15.9	18.8	24.6
Routine indoor chores	5	323	15.5	5.1	11.4	14.9	19.2	24.7
Bicycling	3	284	27.1	12.8	16.7	25.9	34.8	52.1
Shopping	4	278	14.6	5.4	10.9	13.2	16.6	25.0
Racquet sports	1	234	12.8	5.5	8.1	11.8	16.7	22.7
Other passive leisure	1	227	10.8	2.5	9.0	10.4	12.4	15.4

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table C.4. Distribution of Estimated Minute Ventilation[a] for the Ten Most Frequent Activities Among Women with Asthma

Activity	n	Total Time (minutes)	Mean	SD	Minute Ventilation (L/min) Percentile 25	50	75	95
Shopping	3	352	20.3	21.4	10.9	15.3	18.7	75.2
Meals at home	4	340	12.4	4.1	9.5	11.0	14.0	20.9
Errands	2	338	15.0	6.3	9.9	13.4	19.6	26.6
Paid work	1	329	10.4	2.2	8.9	9.7	11.5	15.2
Travel for goods	3	286	18.8	33.2	9.3	10.9	13.7	28.2
Routine indoor chores	2	285	20.4	3.4	17.7	20.4	22.6	26.4
Reading	1	244	7.6	1.9	6.5	6.9	7.8	12.5
Other household chores	1	220	11.7	1.9	10.3	11.7	13.1	14.9
TV viewing	3	219	9.6	2.4	7.6	9.1	11.0	14.6
Other personal travel	3	209	18.0	20.1	9.8	11.4	19.7	47.3

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table C.5. Distribution of Estimated Minute Ventilation[a] for the Ten Most Frequent Activities Among Men with Chronic Obstructive Pulmonary Disease

Activity	n	Total Time (minutes)	Mean	SD	Minute Ventilation (L/min) Percentile 25	50	75	95
TV viewing	3	1,001	17.8	2.3	16.2	17.5	19.0	21.6
Relaxing	4	922	21.4	5.9	16.7	20.3	25.4	31.9
Other passive leisure	3	675	18.5	2.7	17.0	18.6	19.9	22.8
Paid work	2	662	21.5	4.9	17.6	21.2	24.3	30.3
Meals at home	5	599	22.0	4.9	17.9	21.2	25.8	31.0
Naps and resting	3	530	15.2	3.3	12.8	13.9	17.8	22.0
Personal travel	3	917	22.0	5.0	17.8	20.4	26.0	31.0
Travel for goods	4	384	24.9	7.2	19.9	23.3	27.6	40.9
Reading	3	364	19.4	3.7	16.2	18.6	22.6	25.8
Household repairs	1	264	29.6	4.1	26.7	29.5	31.5	36.4

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table C.6. Distribution of Estimated Minute Ventilation[a] for the Ten Most Frequent Activities Among Men with Ischemic Heart Disease

Activity	n	Total Time (minutes)	Mean	SD	Minute Ventilation (L/min) Percentile 25	50	75	95
TV viewing	4	1,048	20.3	4.9	16.8	20.1	22.9	29.6
Meals at home	5	771	20.8	3.9	18.3	20.1	23.1	28.1
Other passive leisure	5	576	17.1	4.0	13.8	16.1	19.7	25.4
Socializing	3	429	16.5	3.4	14.0	16.0	18.6	22.9
Travel for goods	4	425	18.5	17.4	15.2	17.0	19.6	25.4
Main job	1	379	26.1	3.8	23.5	26.1	28.2	33.0
Indoor repairs	2	353	23.4	3.1	21.8	23.7	25.4	27.2
Naps and resting	3	304	17.8	4.1	15.2	16.3	18.7	24.8
Relaxing	4	301	20.9	19.7	15.7	18.7	22.9	29.2
Meals at restaurants	3	249	20.1	2.9	17.9	20.1	21.8	25.6

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

APPENDIX D. Estimated Minute Ventilation (L/min) by Location Class for Women, Children, and Subjects with Heart and Lung Disease

Table D.1. Distribution of Estimated Minute Ventilation[a] by Major Location Class for Healthy Women

Location	n	Total Time (minutes)	Mean	SD	Minute Ventilation (L/min) Percentile 25	50	75	95
Indoors, home	14	10,405	12.1	6.7	8.3	11.3	14.3	20.0
Indoors, work	6	3,600	13.6	3.9	10.8	13.1	16.1	20.0
Indoors, other	14	4,079	15.7	8.7	11.4	13.7	17.5	28.5
Outdoors	13	1,335	18.8	9.3	12.5	16.5	21.9	38.8
In transit	15	2,670	13.8	7.8	9.8	13.0	15.8	23.2

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table D.2. Distribution of Estimated Minute Ventilation[a] by Major Location Class for Healthy Boys

Location	n	Total Time (minutes)	Mean	SD	Minute Ventilation (L/min) Percentile 25	50	75	95
Indoors, home	6	4,625	13.2	4.6	9.9	12.6	15.6	21.0
Indoors, other	5	1,415	14.5	7.8	11.0	13.8	16.5	21.6
Outdoors	6	1,521	20.1	12.1	13.9	16.8	21.6	42.8
In transit	6	734	15.8	7.1	12.4	14.6	17.5	25.6

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table D.3. Distribution of Estimated Minute Ventilation[a] by Major Location Class for Healthy Girls

Location	n	Total Time (minutes)	Mean	SD	Minute Ventilation (L/min) Percentile 25	50	75	95
Indoors, home	6	4,181	13.0	5.2	9.3	11.4	15.9	22.6
Indoors, other	6	3,153	14.0	5.7	10.2	12.6	16.5	24.6
Outdoors	6	1,315	17.6	10.4	10.7	13.8	21.5	40.0
In transit	6	994	13.8	6.0	9.7	11.8	17.6	24.2

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table D.4. Distribution of Estimated Minute Ventilation[a] by Major Location Class for Women with Asthma

Location	n	Total Time (minutes)	Mean	SD	Minute Ventilation (L/min) Percentile 25	50	75	95
Indoors, home	5	1,917	13.8	8.3	9.2	12.2	18.1	23.4
Indoors, work	1	329	10.4	2.2	8.9	9.7	11.5	15.2
Indoors, other	5	1,977	19.1	19.9	9.6	12.5	18.8	68.4
Outdoors	5	214	35.1	27.3	16.0	22.3	62.2	86.1
In transit	5	1,255	17.7	22.0	10.1	12.6	19.2	33.6

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table D.5. Distribution of Estimated Minute Ventilation[a] by Major Location Class for Men with Chronic Obstructive Pulmonary Disease

Location	n	Total Time (minutes)	Mean	SD	Minute Ventilation (L/min) Percentile 25	50	75	95
Indoors, home	5	4,285	20.0	5.4	16.3	18.6	22.8	31.0
Indoors, work	1	441	20.1	4.7	16.7	18.9	22.3	29.8
Indoors, other	5	880	23.3	4.3	20.7	23.3	25.6	30.7
Outdoors	5	1,126	21.5	5.0	18.2	20.5	24.3	30.2
In transit	5	1,365	24.4	5.9	19.9	23.9	28.1	33.7

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table D.6. Distribution of Estimated Minute Ventilation[a] by Major Location Class for Men with Ischemic Heart Disease

Location	n	Total Time (minutes)	Mean	SD	Minute Ventilation (L/min) Percentile 25	50	75	95
Indoors, home	5	4,046	20.3	5.0	16.5	19.7	23.5	29.6
Indoors, work	1	128	24.1	3.0	22.3	24.2	26.1	29.7
Indoors, other	5	1,427	19.1	11.1	14.9	17.9	21.2	29.1
Outdoors	5	1,467	23.7	5.9	18.7	23.5	28.3	33.5
In transit	5	1,008	21.1	12.6	16.0	19.2	24.3	33.6

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

APPENDIX E. Estimated Minute Ventilation (Percentage of Maximum Voluntary Ventilation) by Type of Activity for Women, Children, and Subjects with Heart and Lung Disease

Table E.1. Distribution of Estimated Minute Ventilation[a] Expressed as Percentage of Maximum Voluntary Ventilation for the Ten Most Frequent Activities Among Healthy Women

Activity	*n*	Total Time (minutes)	Mean	SD	% Maximum Voluntary Ventilation Percentile 25	50	75	95
Main job	6	3,600	11.1	3.7	8.4	10.6	13.3	17.5
TV viewing	11	1,964	9.2	5.4	6.9	7.7	9.3	21.5
Meals at home	13	1,253	10.3	3.7	7.9	9.4	11.5	18.2
Meal preparation and cooking	13	1,035	10.2	2.8	8.3	9.8	11.7	14.8
Personal hygiene	14	710	12.5	4.5	9.1	11.4	14.7	21.1
Naps and resting	7	699	8.8	2.8	6.6	8.1	11.0	13.2
Reading	5	634	11.8	7.2	7.1	8.7	12.2	25.8
Other personal travel	11	549	13.8	8.5	8.5	11.5	16.8	26.3
Travel to and from work	6	470	11.5	5.0	8.3	10.3	13.7	19.2
Studying	3	460	9.1	2.3	7.3	8.4	10.5	13.5

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table E.2. Distribution of Estimated Minute Ventilation[a] Expressed as Percentage of Maximum Voluntary Ventilation for the Ten Most Frequent Activities Among Healthy Boys

Activity	*n*	Total Time (minutes)	Mean	SD	% Maximum Voluntary Ventilation Percentile 25	50	75	95
TV viewing	5	929	13.2	7.1	7.6	12.3	16.4	27.2
Routine indoor chores	4	561	13.1	6.0	8.2	11.4	15.6	25.8
Attending classes	2	537	11.6	8.3	6.8	9.6	13.9	22.3
Meals at home	6	518	14.0	7.9	9.5	10.9	14.9	30.5
Computer games	2	462	21.1	8.8	13.7	22.8	25.1	31.4
Naps and resting	3	403	8.0	3.2	5.4	7.3	9.4	14.1
Shopping	5	306	16.5	5.5	13.9	15.6	17.9	24.6
Meal preparation and cooking	4	277	11.3	3.6	9.5	10.7	13.2	17.9
Studying	2	247	8.9	3.2	7.0	8.6	10.1	14.7
Outdoor play	2	238	26.5	9.0	20.5	25.3	30.1	41.4

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table E.3. Distribution of Estimated Minute Ventilation[a] Expressed as Percentage of Maximum Voluntary Ventilation for the Ten Most Frequent Activities Among Healthy Girls

Activity	n	Total Time (minutes)	Mean	SD	% Maximum Voluntary Ventilation Percentile			
					25	50	75	95
Attending classes	4	1,275	14.3	5.8	10.6	12.5	16.2	25.8
TV viewing	6	1,213	14.2	7.1	9.8	12.4	16.5	25.9
Meals at home	6	919	13.9	5.5	10.2	12.5	16.6	23.5
Outdoor play	3	555	22.8	13.6	15.0	18.3	23.8	55.1
Personal hygiene	6	428	17.7	5.4	14.0	17.0	20.4	27.0
Routine indoor chores	5	323	17.4	4.7	13.6	16.6	20.6	26.2
Bicycling	3	284	28.3	13.0	17.7	26.8	36.2	53.7
Shopping	4	278	16.2	6.1	11.2	14.1	20.8	26.8
Racquet sports	1	234	15.1	6.5	9.5	13.9	19.8	26.8
Other passive leisure	1	227	12.8	3.0	10.6	12.3	14.7	18.2

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table E.4. Distribution of Estimated Minute Ventilation[a] Expressed as Percentage of Maximum Voluntary Ventilation for the Ten Most Frequent Activities Among Women with Asthma

Activity	n	Total Time (minutes)	Mean	SD	% Maximum Voluntary Ventilation Percentile			
					25	50	75	95
Shopping	3	352	17.4	17.7	9.8	11.5	15.8	74.0
Meals at home	4	340	10.2	3.3	7.4	10.3	12.3	16.8
Errands	2	338	10.6	4.3	7.3	9.6	13.0	19.5
Paid work	1	329	7.6	1.6	6.5	7.1	8.4	11.2
Travel for goods	3	286	13.9	19.1	8.1	9.6	11.5	24.7
Routine indoor chores	2	285	12.5	2.3	10.8	12.3	13.9	16.4
Reading	1	244	5.5	1.4	4.7	5.0	5.6	9.0
Other household chores	1	220	11.6	1.9	10.1	11.5	12.9	14.7
TV viewing	3	219	7.0	1.8	5.6	6.7	8.1	10.6
Other personal travel	3	209	15.9	15.0	9.4	11.0	14.6	45.8

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table E.5. Distribution of Estimated Minute Ventilation[a] Expressed as Percentage of Maximum Voluntary Ventilation for the Ten Most Frequent Activities Among Men with Chronic Obstructive Pulmonary Disease

Activity	n	Total Time (minutes)	Mean	SD	% Maximum Voluntary Ventilation Percentile			
					25	50	75	95
TV viewing	3	1,001	37.5	7.9	34.5	38.5	42.0	48.5
Relaxing	4	922	32.5	11.8	22.6	27.3	42.4	54.0
Other passive leisure	3	675	40.8	12.0	28.0	44.3	49.6	55.9
Paid work	2	662	35.7	9.0	28.7	34.6	40.3	51.3
Meals at home	5	599	42.7	13.9	29.7	41.4	54.9	65.2
Naps and resting	3	530	31.5	11.7	21.9	28.4	34.6	55.6
Personal travel	3	417	46.0	10.3	38.4	43.3	51.3	66.7
Travel for goods	4	384	49.2	16.3	34.3	51.9	59.6	71.4
Reading	3	364	41.9	5.4	37.9	42.1	45.7	49.7
Household repairs	1	264	56.0	7.7	50.5	55.8	59.6	68.8

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table E.6. Distribution of Estimated Minute Ventilation[a] Expressed as Percentage of Maximum Voluntary Ventilation for the Ten Most Frequent Activities Among Men with Ischemic Heart Disease

Activity	n	Total Time (minutes)	Mean	SD	% Maximum Voluntary Ventilation Percentile 25	50	75	95
TV viewing	4	1,048	20.2	6.2	15.7	20.2	24.9	29.1
Meals at home	5	771	18.0	6.6	10.6	19.3	22.8	27.6
Other passive leisure	5	576	18.8	4.5	15.9	18.9	21.3	26.9
Socializing	3	429	19.5	3.7	16.8	18.9	21.5	26.1
Travel for goods	4	425	18.9	9.6	14.2	19.2	22.3	29.1
Main job	1	379	32.3	4.7	29.1	32.3	40.0	40.9
Indoor repairs	2	353	13.0	4.1	10.9	11.7	12.5	22.3
Naps and resting	3	304	14.9	4.7	11.9	13.2	17.9	22.7
Relaxing	4	301	17.6	9.6	13.9	17.2	19.3	26.2
Meals at restaurants	3	249	24.6	3.7	21.9	24.3	26.9	31.4

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

APPENDIX F. Estimated Minute Ventilation (Percentage of Maximum Voluntary Ventilation) by Location Class for Women, Children, and Subjects with Heart and Lung Disease

Table F.1. Distribution of Estimated Minute Ventilation[a] Expressed as Percentage of Maximum Voluntary Ventilation by Major Location Class for Healthy Women

Location	n	Total Time (minutes)	Mean	SD	% Maximum Voluntary Ventilation Percentile 25	50	75	95
Indoors, home	14	10,405	10.9	6.7	7.5	9.3	11.9	22.0
Indoors, work	6	3,600	10.8	3.5	8.4	10.2	12.5	17.0
Indoors, other	14	4,079	12.8	6.0	9.5	11.7	14.0	22.9
Outdoors	13	1,335	16.9	9.6	10.6	13.4	19.5	37.6
In transit	15	2,670	11.8	7.2	8.3	10.4	13.4	21.1

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table F.2. Distribution of Estimated Minute Ventilation[a] Expressed as Percentage of Maximum Voluntary Ventilation by Major Location Classes for Healthy Boys

Location	n	Total Time (minutes)	Mean	SD	% Maximum Voluntary Ventilation Percentile 25	50	75	95
Indoors, home	6	4,625	13.6	7.7	8.4	11.6	16.8	28.6
Indoors, other	5	1,415	15.1	9.0	8.8	13.6	18.5	30.0
Outdoors	6	1,521	21.7	16.6	13.3	17.1	25.2	46.9
In transit	6	734	17.9	10.3	11.5	15.6	23.6	31.5

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table F.3. Distribution of Estimated Minute Ventilation[a] Expressed as Percentage of Maximum Voluntary Ventilation by Major Location Classes for Healthy Girls

Location	n	Total Time (minutes)	Mean	SD	% Maximum Voluntary Ventilation Percentile			
					25	50	75	95
Indoors, home	6	4,181	14.8	5.7	10.9	13.7	17.7	25.4
Indoors, other	6	3,153	16.1	7.0	11.2	14.4	19.4	28.2
Outdoors	6	1,315	22.7	13.7	13.9	18.5	26.2	52.9
In transit	6	994	15.7	6.6	10.9	13.9	19.2	27.4

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table F.4. Distribution of Estimated Minute Ventilation[a] Expressed as Percentage of Maximum Voluntary Ventilation by Major Location Classes for Women with Asthma

Location	n	Total Time (minutes)	Mean	SD	% Maximum Voluntary Ventilation Percentile			
					25	50	75	95
Indoors, home	5	1,917	9.9	4.9	7.1	10.0	12.1	14.8
Indoors, work	1	329	7.6	1.6	6.5	7.1	8.4	11.2
Indoors, other	5	1,977	14.1	13.5	8.0	10.2	13.2	46.0
Outdoors	5	214	32.1	28.2	11.9	16.5	61.2	84.7
In transit	5	1,255	13.0	13.6	8.1	10.2	13.2	23.6

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table F.5. Distribution of Estimated Minute Ventilation[a] Expressed as Percentage of Maximum Voluntary Ventilation by Major Location Classes for Men with Chronic Obstructive Pulmonary Disease

Location	n	Total Time (minutes)	Mean	SD	% Maximum Voluntary Ventilation Percentile			
					25	50	75	95
Indoors, home	5	4,285	38.8	13.0	27.5	38.6	47.4	61.5
Indoors, work	1	441	32.7	7.7	27.3	30.7	36.4	48.5
Indoors, other	5	880	46.5	14.8	36.4	46.5	58.2	68.2
Outdoors	5	1,126	41.2	12.5	32.9	41.7	49.6	62.3
In transit	5	1,365	47.9	12.8	38.9	46.7	55.6	68.2

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

Table F.6. Distribution of Estimated Minute Ventilation[a] Expressed as Percentage of Maximum Voluntary Ventilation by Major Location Classes for Men with Ischemic Heart Disease

Location	n	Total Time (minutes)	Mean	SD	% Maximum Voluntary Ventilation Percentile			
					25	50	75	95
Indoors, home	5	4,046	18.6	5.8	14.6	18.8	22.8	28.2
Indoors, work	1	128	29.9	3.7	27.6	29.9	32.3	36.9
Indoors, other	5	1,427	19.5	7.8	15.0	19.1	23.6	29.1
Outdoors	5	1,467	20.6	9.4	12.5	19.4	28.1	36.9
In transit	5	1,008	21.0	8.8	15.7	20.3	26.0	34.1

[a] Minute ventilation estimated from ambulatory heart rates using the $\dot{V}_E$-on-HR regressions from the maximal exercise test for each individual.

ABOUT THE AUTHORS

Jonathan Samet, M.D., the principal investigator for these studies, is Professor of Medicine and Chief of the Pulmonary and Critical Care Division at the University of New Mexico Hospital. He is also the Director of Epidemiology at the New Mexico Tumor Registry. He received his M.D. degree from the University of Rochester School of Medicine and Dentistry and received clinical training in internal medicine and the subspecialty of pulmonary diseases. He completed a three-year fellowship in clinical epidemiology at the Channing Laboratory, Harvard Medical School, and has an M.S. in epidemiology from the Harvard School of Public Health. His research interests focus on environmental and occupational epidemiology and the effects of chemical and radioactive agents on the lung.

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PUBLICATIONS RESULTING FROM THIS RESEARCH

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ABBREVIATIONS

bpm	beats per minute
ECG	electrocardiogram
f	respiratory frequency
HR	heart rate
IQR	interquartile range
MVV	maximum voluntary ventilation
NAR	nasal airway resistance
R	respiratory exchange ratio
$\dot{V}_{CO_2}$	volume of carbon dioxide expired per minute
$\dot{V}_E$	volume of air expired per minute (minute ventilation)
$\dot{V}_E$-HR	minute ventilation–to–heart rate relation
$\dot{V}_E$-HR(full)	minute ventilation estimates based on full range of heart rates
$\dot{V}_E$-HR(60–120)	minute ventilation estimates based on heart rates in the range of 60 to 120 bpm
$\dot{V}_I$	volume of air inspired per minute
$\dot{V}_{O_2}$	volume of oxygen uptake per minute
V_T	tidal volume

INTRODUCTION

A Request for Applications (RFA 88–1) that solicited proposals for "Ozone and Carbon Monoxide: Assessment of Population Exposure and Dose" was issued by the Health Effects Institute (HEI) in the summer of 1988. In response to this RFA, Dr. F. Dennis McCool from the Memorial Hospital of Rhode Island, in Pawtucket, RI submitted a proposal to HEI, entitled "Measurements of Ventilation in Freely Ranging Subjects." Under this same RFA, Dr. Jonathan M. Samet from the University of New Mexico, in Albuquerque, NM, submitted a proposal entitled "Ventilation, Activity, and Lung Doses of Air Pollutants." Dr. McCool's study began in July 1989 and ended in December 1990; total expenditures were $100,866. Dr. Samet's study began in July 1989 and ended in March 1991; total expenditures were $132,866. Both Investigators' Reports were received in June 1991, and both were accepted by the Health Review Committee in April 1992.

Because both studies dealt with the development of methods for measuring ventilation noninvasively, the Health Review Committee decided to publish both reports in a single HEI Research Report with one Health Review Committee Commentary addressing both studies. This approach provides a ready comparison between the protocols, results, and conclusions for these two studies and a more comprehensive presentation of current knowledge about noninvasive measurements of ventilation. The Health Review Committee's Commentary is intended to place the Investigators' Reports in perspective, as an aid to the sponsors of HEI and to the public. During HEI's review of these studies, Drs. Samet and McCool had an opportunity to review each other's report. During this review process, the Health Review Committee and the investigators also had the opportunity to exchange comments and to clarify issues in the Investigators' Reports and in the Health Review Committee's Commentary.

REGULATORY BACKGROUND

The U.S. Environmental Protection Agency (EPA) sets standards for criteria pollutants under Section 202 of the Clean Air Act, as amended in 1990. Section 202(a)(1) directs the Administrator to "prescribe (and from time to time revise) . . . standards applicable to the emission of any air pollutant from any class or classes of new motor vehicles or new motor vehicle engines, which in his judgment cause, or contribute to, air pollution which may reasonably be anticipated to endanger public health or welfare." Sections 202(a), (b)(1), (g), and (h), and Sections 207(c)(4), (5), and (6) impose specific requirements for reducing motor vehicle emissions of certain oxidants (and other pollutants) and, in some cases, provide the EPA with limited discretion to modify those requirements.

As further outlined in Section 202(a) of the Clean Air Act Amendments, the Administrator shall determine whether emissions of any unregulated pollutant present an unreasonable risk to the public health, welfare, or safety. One approach used by the EPA to estimate the likelihood that pollutants will produce adverse health effects is risk assessment. This mathematical approach characterizes and quantifies potential detrimental effects that may result from exposures to harmful agents in the environment. One phase of the risk assessment process is exposure assessment, which estimates the magnitude, frequency, duration, and route of exposure to a pollutant. Exposure assessment is critical for appropriately interpreting an association between an individual's exposure to a pollutant and the amount of the pollutant potentially reaching its target site in the body and producing a biological effect. Studies that lead to improvements in exposure assessment methodology, such as the two discussed in this Commentary, can improve estimates provided by risk assessment procedures.

SCIENTIFIC BACKGROUND

Humans are exposed to a wide variety of pollutants in the air they breathe each day. An important health question stemming from these exposures is whether chronic inhalation of outdoor and indoor air pollutants is associated with an increased risk of adverse health effects and lung disease (American Thoracic Society 1985; Crapo et al. 1992; U.S. Congress 1992). The federal government has determined that a basic criterion for an air pollutant's regulation under the Clean Air Act (U.S. Congress 1991) is its ability to cause adverse health effects. Because air pollutants enter the body primarily through the respiratory tract, ventilation, defined as the volume of air entering or exiting the lungs, is an important determinant of the potentially detrimental effects of inhaled air pollutants. Ventilation governs the amounts, or doses, of air pollutants that reach the respiratory tract, and ultimately, the effects of these pollutants both on healthy individuals and susceptible individuals in the population.

THE RISK ASSESSMENT PROCESS

Recognition of the complexity of evaluating health effects of everyday exposures to air pollutants and other potentially toxic substances has led to the development of methods to assess the potential risks associated with these exposures. Risk assessment is a quantitative approach that has evolved during the last ten years as a process for estimating the likelihood that a given pollutant will produce human health effects (National Research Council 1991). The four basic components that constitute the risk assessment process are hazard identification, dose-response assessment, exposure assessment, and risk characterization.

The first two phases of risk assessment, hazard identification and dose-response assessment, are grouped under the term "health effects assessment." Hazard identification establishes that a substance produces an adverse effect, and dose-response assessment quantifies the health response as a function of the dose to the organism. Dose-response data are based on results from in vitro assays, in vivo animal tests, and, if available, data from studies with humans. Improvements in the technology for measuring small amounts of toxicants in the environment and innovations in techniques to assess changes in biological systems, particularly through molecular biology, have refined the health effects assessment process. However, these advances have not been matched by comparable advances in exposure assessment, the next step in the risk assessment process.

Exposure assessment estimates the exposures to a given pollutant that an individual may experience; these estimates are based on the magnitude, frequency, and duration of exposure to a substance (National Research Council 1991; U.S. Congress 1992). The term "exposure" can be defined simply as contact between a human and a toxic substance for a specific period of time. Estimates of exposure are calculated from several parameters, including toxicant concentration, exposure conditions, and route of exposure (e.g., inhalation, ingestion, or absorption through the skin). For the purposes of this discussion, exposure differs distinctly from dose. Exposure refers to the conditions and frequency under which an individual is exposed to a pollutant, whereas dose is the amount of a pollutant that is absorbed by or deposited in the tissues of an exposed individual during a specific time interval (National Research Council 1991).

Improvements in exposure assessment methodology are essential for reducing the uncertainty in the calculations used in the risk assessment process (National Research Council 1991). To date, numerous environmental regulations have been based on eight-hour work-place exposures and 24-hour outdoor exposures for a standard 70-kg man (National Research Council 1991). Similarly, epidemiological studies have often used exposure as a surrogate for dose because of the complexity of determining actual doses for large study populations. Such estimates of exposure cannot, however, accurately reflect the wide variability in doses experienced by the general population because individuals vary in age, gender, activity level, and health status.

For inhalation scenarios, the focus of the present discussion, exposure has often been calculated as the product of the concentration of a given pollutant in the air multiplied by the time interval, or duration, of an exposure. For repeated exposures, exposure frequency can be included in these calculations. This approach yields a simple, but often inaccurate, measure of dose for many individuals because it does not consider the variability in breathing patterns for the exposed subjects. For example, during the course of a day, people experience a variety of breathing patterns, or ventilation levels, due to various activities. These levels and the different environments in which they occur directly influence an individual's dose of an inhaled pollutant. For example, personal exposure scenarios are affected by the nature of the work place, modes of commuting, recreational choices, and the ambient atmosphere within the home. This range of individual life-styles presents complicating factors for determining how people breathe during their daily lives and the doses of pollutants they inhale.

VENTILATION MEASUREMENTS IN EXPOSURE ASSESSMENT

Increasing the validity and reliability of information that describes an individual's ventilatory patterns during exposures to air pollutants is a critical factor for improving dose estimates (Morgan and Frank 1977; Brain and Valberg 1979). It is clear that increases in frequency and depth of breathing can increase the volume of pollutant-laden air that enters the respiratory tract. Such increases can also result in a greater fraction of inhaled particles and gases reaching and being retained in the airspaces of the lungs. Inhaled pollutants are removed more slowly from this gas exchange region because it lacks the mucous layer and the action of ciliated cells that work together in the airways to trap and clear pollutants from the lungs. Experiments with human subjects have demonstrated that exercise and its accompanying rise in ventilation increase the total lung burden, or retained amount, of an inhaled aerosol (Bennett et al. 1985).

Whether an individual breathes through the nose or mouth (the route of breathing) affects both dose and deposition sites. If one breathes through the mouth and bypasses the air-scrubbing action of the nose and pharynx, more inhaled material will reach the lower airways and the gas exchange region in the lungs (Morgan and Frank 1977). Increasing ventilation rates, for example, through exercise, naturally shifts the route of breathing from nose to mouth and similarly increases the dose of an inhaled pollutant.

Ventilation measurements that characterize an individual's breathing patterns can be determined easily in the laboratory using specialized equipment, such as spirometers or pneumotachographs. A subject usually wears a nose clip and breathes via a mouthpiece or face mask into this equipment while specific data about breathing are recorded. One standard measure of ventilation is minute volume ($\dot{V}_E$), the total amount of air exhaled in one minute. Minute ventilation values can range from 10 to 20 L/min while at rest, and up to 100 L/min during heavy exercise. Other ventilatory parameters include various measures of lung capacity and time intervals for inspiration and expiration. Ventilation data also can be combined with estimates for the fraction of a pollutant that remains in the respiratory tract after inhalation. If both the volume of inhaled air and the percentage of the pollutant retained from that inhaled air are known, more accurate estimates of pollutant dose can be calculated.

One recognized artifact of ventilation measurements obtained with a spirometer or pneumotachograph is that the presence of the mouthpiece or face mask alters the ventilation measurements. Gilbert and colleagues (1972) were the first to report that the presence of a mouthpiece increased tidal volume and inspiratory time and decreased breath frequency in their study subjects. These artifacts create uncertainty about the relevance of laboratory ventilatory measurements to the way people breathe in everyday life. The use of noninvasive methods for measuring ventilation, that is, methods not requiring a mouthpiece, face mask, or nose clip, could eliminate such effects. By the very definition of the term noninvasive, such methods can only indirectly measure the actual volumes of air that a study subject breathes. Therefore, a critical step toward the future use of noninvasive methods for measuring ventilation is determining the accuracy and feasibility of this approach compared to conventional approaches.

NONINVASIVE MEASUREMENTS OF VENTILATION

Investigators have been interested in methods for measuring ventilation noninvasively for many years. In early studies (Konno and Mead 1967; Stagg et al. 1978; Robertson et al. 1980), special devices that can measure magnetic field intensity, called magnetometers, were affixed to the front and back of a subject's rib cage and abdomen to detect changes in body dimensions during breathing maneuvers. Changes in distance between specific sets of excitation and receiver magnetometers were correlated with changes in lung volume, as measured by spirometry. One limitation of this approach was that it provided accurate correlations with spirometry data only if the subjects remained upright.

Sackner and colleagues (1980) subsequently described techniques for measuring ventilation noninvasively using respiratory inductive plethysmography. (A commercially available product is called Respitrace.) With this technique, a subject wears bands of elastic material containing coils of Teflon-coated wire around the chest and abdomen. Movements during breathing change the cross-sectional areas of the rib cage and abdomen and alter the electrical signals sent from the coils. Changes in these signals then can be calibrated with ventilatory volumes measured simultaneously with a spirometer. One advantage of this method is that it detects changes in cross-sectional areas of the rib cage and abdominal compartments rather than linear changes in distance measured by magnetometers. As a result, respiratory inductive plethysmography can measure changes in lung volumes more accurately over a wider range of body positions than magnetometers alone.

HEART RATE AS A NONINVASIVE ESTIMATOR OF VENTILATION

An alternative approach for measuring ventilation noninvasively uses heart rate as a predictor of ventilation. The physiological basis for this relationship is that ventilation and heart rate vary together to match the oxygen needs of the body. As activity and ventilation levels fluctuate, the heart rate changes so that appropriate amounts of oxygen-carrying blood are delivered to the cells (Weibel 1984; Wasserman et al. 1987). Establishing a mathematical relation between heart rate and ventilation could be very useful for determining ventilation noninvasively because heart rate is much easier to measure than ventilation. As discussed previously, ventilation measurements usually require relatively cumbersome clinical equipment, such as a spirometer or pneumotachograph that entails the use of an invasive mouthpiece. In comparison, heart rate, or pulse rate, can be measured readily with a variety of commercially available athletic watches designed for this purpose.

That ventilation would match heart rate in a rather simple, linear fashion seems intuitively obvious; as muscles demand more oxygen, the heart delivers more blood. However, several physiological phenomena interfere with such a direct correlation (Vander et al. 1985). For example, elevations in heart rate often lag behind increases in activity level. Although the mechanism for this response is not yet understood, increases in ventilatory rate precede increases in heart rate during the first few minutes of increased activity. Another factor is the voluntary control one can exert over breathing, as exemplified by sighing or panting. Such voluntary increases in ventilation can occur without producing corresponding increases in heart rate. Drugs, such as beta-blockers, can also interfere with any direct relationship between heart rate and ventilation by modulating changes in heart rate. Fitness level is an additional factor that can directly influence heart rate because regular exercise can lower basal heart rate.

Several investigator groups have recently explored the use of the relationship between heart rate and ventilation data to refine estimates of pollutant dose. For example, Raizenne and Spengler (1988) used this approach to calculate inhaled pollutant doses from time-activity data and estimates of ventilation based on heart rates in girls attending summer camp. They estimated the doses of ozone and acid aerosols inhaled from the ambient air with four parameters: a retention factor that predicted the amounts of the air pollutants retained in the lungs, minute ventilation, time or duration of the exposure, and the mean pollutant concentration during the exposure interval. They then correlated their dose calculations with measured changes in lung function. In another study, Shamoo and colleagues (1990) tested two indirect methods for estimating ventilation. They trained subjects to estimate their own ventilation levels with a portable device and also measured the subjects' heart rates with athletic watches. They concluded that ventilation estimates based on heart rate provided a viable approach for improving dose estimates.

Despite both the physiological caveats and the advantages of using heart rate as a predictor of ventilation, the utility of this approach for large-scale epidemiological studies has not yet been tested rigorously. Much remains to be learned about the variability of this relationship among individuals and the influence of factors such as age, gender, fitness, and health status, and the normal variability in ventilation and heart rates that ordinary individuals experience each day.

Methods for noninvasive monitoring of ventilation and breathing patterns in subjects while they move about freely in their work and home environments can lead to improved estimates of dose for epidemiological studies. Such refinements in exposure assessment technology could provide more realistic data for the amounts of toxic substances that are inhaled by humans during everyday life. The resulting progress in exposure assessment methodology can ultimately reduce uncertainty in the estimates provided by the risk assessment process.

JUSTIFICATION FOR THE MCCOOL AND SAMET STUDIES

In 1988, HEI recognized the uncertain relationship between outdoor air pollutant levels measured by fixed monitors and personal exposures to these pollutants. Because variations in breathing patterns, environments, and activity levels throughout the day affect an individual's exposures to pollutants, measurements of outdoor air pollutant levels alone are poor predictors of dose.

In its RFA 88-1, HEI requested proposals for research to improve estimates of personal exposure and dose for two specific pollutants, ozone and carbon monoxide, in populations that move about freely. Specific objectives included: (1) developing and testing personal exposure monitors for ozone; (2) developing and testing methods for estimating individual ventilation, breathing patterns, and route of breathing (oral versus nasal breathing as a predictor of pollution dose to the lungs); and (3) characterizing the carbon monoxide dose in human populations.

Drs. Dennis McCool and Jonathan Samet each submitted a proposal in response to the second aim of this RFA. Both investigators were interested in exploring approaches for measuring ventilation in subjects while they moved about freely in their environments. They both also wanted to develop and test methods for measuring ventilation without the use of any invasive apparatus, such as a mouthpiece.

Dr. McCool proposed to evaluate a model for measuring ventilation noninvasively in unrestrained subjects that was based on an earlier model developed by Konno and Mead (1967). He stated that flexion and extension of the spine, which occurs normally during many physical activities, had not been factored into the Konno and Mead model. That model had measured ventilation with magnetometers by detecting changes in the dimensions of rib cage and abdomen only. Dr. McCool proposed a body surface displacement (BSD)* model that combined respiratory inductive plethysmography belts with magnetometers. The belts measured rib cage and abdominal changes, while magnetometers at the base of the breastbone and at the navel of a study subject detected changes in the distance between them and indicated alterations in spinal flexion. To support the feasibility of this new approach, Dr. McCool described results from preliminary studies in which he and his colleagues at the Harvard School of Public Health had estimated ventilation using this three-parameter BSD model.

Dr. Samet and his colleagues proposed to develop and validate methods for estimating ventilation from heart rate by recording data with a special athletic watch, called the Heartwatch. The goal of their research was to evaluate the feasibility of this approach for estimating ventilation in large numbers of human subjects who participate in epidemiologic studies. Their estimates of ventilation would be correlated with time, place, and activity data recorded throughout the day in diaries maintained by the subjects. Their overall objective was to combine ventilatory estimates with the diary information and obtain more accurate projec-

* A list of abbreviations appears at the end of the Investigators' Reports.

tions for the doses of air pollutants that subjects inhale during the course of their daily lives.

As part of their study, Dr. Samet and his colleagues planned to evaluate several important factors that could affect doses of inhaled air pollutants. These factors included ventilatory rates during different work tasks and identification of nasal or oral breathing patterns at varied levels of work task intensity. In his proposal, Dr. Samet, a physician and epidemiologist from the University of New Mexico, also described his own extensive experience, as well as that of his collaborators, for evaluating the health effects of indoor and outdoor air pollutants.

The Research Committee noted that both teams of investigators were well-qualified to conduct studies on noninvasive methods of ventilation. The Committee suggested minor modifications for both proposals and recommended that the investigators discuss their respective proposals with each other. As requested by the Committee, Dr. McCool included in his revised proposal experiments to evaluate the relationship between heart rate and ventilation so that these data could serve as a cross-validation for data from similar experiments proposed by Dr. Samet and colleagues. The Research Committee subsequently recommended funding for revised proposals from both investigator groups.

SPECIFIC AIMS AND STUDY DESIGN FOR THE MCCOOL STUDY

Dr. McCool and his colleague, Dr. Paek, proposed four specific aims; the first three encompassed laboratory studies and the fourth was a pilot field study to test their laboratory methods. The first specific aim was to compare ventilatory data obtained from their BSD model with data obtained by spirometry, a conventional approach for measuring pulmonary function parameters. The investigators recorded changes in dimensions of the chest wall and abdomen with respiratory inductive plethysmography belts (Respitrace system) and magnetometers. This BSD equipment was worn by 10 laboratory subjects (eight males and two females) while they simultaneously breathed into a spirometer to record corroborative data about lung volume measurements and breath times.

The investigators recorded spirometric data for tidal volume, breath frequency, inspiration time, and total breath time. As a first step, they plotted the spirometric volume measurements against the body surface measurements recorded with the BSD equipment. Using regression analysis methods, they then determined the slopes of these plots and used these slope values as volume-motion coefficients in their equations for calculating breathing parameters from the BSD model. Subsequent BSD data collected while subjects breathed with or without the mouthpiece were then input directly into these equations to obtain values for $\dot{V}_E$. Other motion data recorded by the BSD equipment were used to determine values for breath frequency as well as inspiratory and total breath times. These BSD data were also compared with spirometry data collected during the same recording period to compare the levels of precision for the two recording techniques.

For their second specific aim, the investigators evaluated how breathing pattern parameters were affected by different work tasks and by the presence of a mouthpiece. Under a range of work loads, the subjects performed different tasks, including the upper body activities of lifting, pulling, and arm cranking (in which the subject performs a pedaling motion with the arms), and the lower body activity of cycling. Following a four- to six-minute warm-up period, the subjects performed each work task for four minutes. During this four-minute work period, the investigators also evaluated the influence of a mouthpiece on the ventilation measurements. During the first two minutes, the subjects breathed through a mouthpiece into a spirometer while wearing the BSD equipment; during the latter two minutes, they breathed without the mouthpiece while data was recorded by BSD only. Following a rest period, the work task session was repeated with the mouthpiece sequence reversed.

The investigators' third specific aim was to evaluate the validity of using heart rate as an index of $\dot{V}_E$. The data for this specific aim were collected in the laboratory from the nine subjects who also participated in the field study, the fourth specific aim. During this laboratory session, the subjects rested for five minutes and then cycled with progressively increasing work loads for 15 minutes while wearing the BSD equipment. Heart rate was measured with electrodes attached to the chest. The subjects performed this test twice, once while breathing through a mouthpiece into the spirometer and once without the mouthpiece. The investigators then analyzed the effects of different work tasks and the mouthpiece on the relationship between ventilation and heart rate.

The fourth specific aim was to conduct a field study to evaluate the feasibility of using the BSD model to measure ventilation outside of the laboratory. The subjects were nine vocational high-school students who had not participated in the earlier laboratory phase for their specific aim. The BSD and heart rate data were collected over a two-hour period that included a sedentary classroom period and an auto body repair workshop period in which the subjects painted, sanded, and welded. During this period, subjects were continuously attached to all of the equipment needed for recording BSD and heart rate; the equipment was placed on a cart to ease subject mobility during the classroom and workshop sessions. The BSD measurements were calibrated during the classroom sessions with data from a pneumotachograph, a portable device for measuring lung volumes. Baseline data for BSD and heart rate were obtained in a laboratory session that followed the workshop session described above. To evaluate different methods for estimating $\dot{V}_E$ from heart rate, the investigators used both linear ($\dot{V}_E$-HR[full]) and exponential ($\dot{V}_E$-HR[exp]) approaches to derive calibration curves over a full range of exertion extending from rest to heavy exercise; they used a linear model ($\dot{V}_E$-HR[low]) to derive a calibration curve over a low range of exertion.

TECHNICAL EVALUATION OF THE MCCOOL STUDY

ATTAINMENT OF STUDY OBJECTIVES

Drs. McCool and Paek accomplished the four specific aims of their project. Their laboratory studies established the validity of the BSD model for measuring ventilation and breathing patterns accurately in freely mobile individuals. Using data obtained for a variety of work tasks and work loads, they evaluated differences in breathing patterns induced by upper and lower body work tasks. They also evaluated the relationship between heart rate and ventilation for a range of exercise intensity levels and the effects of a mouthpiece upon ventilation. Finally, their field study demonstrated the utility of their BSD model for measuring ventilation outside of the laboratory.

STUDY DESIGN AND METHODS

The study design was described clearly, and the methods for obtaining data from the BSD equipment and for calculating the results from the model's equations were reported in appropriate detail, as described previously (McCool et al. 1986; Paek et al. 1990).

The investigators had divided their study into two major phases: laboratory studies and a field study. Two strong aspects of the laboratory studies were the selection of a variety of work tasks involving both upper and lower body and the inclusion of a range of work loads. Through comparisons with spirometric data, they successfully validated the BSD model as a method for measuring various ventilatory parameters. The investigators then applied their BSD methodology to assess the effects of the presence of a mouthpiece on ventilatory measurements. All work tasks were repeated with the order of the mouthpiece presence and absence reversed to reduce bias that might have been introduced by the experimental sequence order. The investigators then compared the BSD data obtained from the on- and off-mouthpiece periods to evaluate the impact of the presence of the mouthpiece on ventilatory measurements.

The field study was well designed for evaluating the feasibility of using the BSD model for studies outside of the laboratory. Completion of this field study required integration of several complex components, including the design of a system that facilitated equipment mobility, careful measurement of ventilation data during the classroom and workshop sessions, and subsequent collection of baseline data in the laboratory.

STATISTICAL METHODS

This study utilized several different protocols to collect and analyze data for the laboratory and field studies. The investigators combined the data from all exercise periods in which the mouthpiece was present and used these data to develop the motion-volume coefficients necessary for Equation 2 described in their report. This equation was then used to estimate tidal volume based on the data recorded from the BSD equipment during the periods of breathing with and without a mouthpiece.

Although the study utilized a somewhat complex experimental design, the data analysis did not require complicated statistical methods. The ventilation data for the various measurement periods were summarized for each subject as means, regression coefficients, or measures of correlation, such as the coefficient of determination (R^2). These values were then analyzed by paired t tests, analyses of variance, and other relatively straightforward statistical methods.

The investigators calculated R^2 values for the data from the BSD model and from the spirometric measurements (Table 2). This coefficient indicates the strength of the correlation between these two types of data. The investigators also presented the mean percentage differences between the spirometric values and the BSD measurements (Table 3). Their results indicated good agreement between the values measured by spirometry and those estimated from the BSD model data.

RESULTS AND INTERPRETATIONS

The central finding from this study was that reliable estimates of ventilation and other breathing parameters could be obtained with the BSD model while subjects performed a wide variety of work tasks. The investigators clearly demonstrated the feasibility and accuracy of their BSD model both in the laboratory and in the field.

Previous investigators had estimated ventilation noninvasively with models that used only two degrees of freedom (Konno and Mead 1967; Robertson et al. 1980). Drs. McCool and Paek improved the accuracy of these approaches by adding a third component that accounted for spinal flexion. They demonstrated that $\dot{V}_E$ estimated from their BSD model incorporating three degrees of freedom correlated well with standard spirometric measurements of ventilation. Their results also illustrate the importance of careful calibration between these two ventilation measurement techniques for producing accurate BSD data.

The investigators observed that the BSD measurements and spirometric data were most comparable during the monotonous tidal volume patterns induced by the lower body work of cycling. The more erratic breathing patterns associated with the upper body work, particularly lifting, produced the greatest discrepancies between the data recorded from two methods (Figure 3). They noted that, although mean values of $\dot{V}_E$ were comparable for different tasks, the type of work task notably affected the degree to which individual measurements varied from the mean value (Figure 4).

Effects of a Mouthpiece on Ventilation Measurements

The investigators reported that the mean values of the measured respiratory parameters were significantly affected by the presence of a mouthpiece (Table 4). The effects of the mouthpiece were less at higher levels of ventilation, such as those induced by vigorous exercise, than at lower ventilatory levels. The effect of the mouthpiece on $\dot{V}_E$ was independent of task. However, other aspects of breathing were found to be influenced more subtly by a mouthpiece. With a mouthpiece, breathing patterns were more erratic during the upper body work tasks than during the lower body task of cycling.

The investigators used standard statistical methods, including paired *t* tests and analysis of variance, to show that the presence of a mouthpiece induced significant differences in the measured variables for all tasks (Tables 4 and 5). Tasks performed with a mouthpiece were associated with higher levels of ventilation than those observed without a mouthpiece. However, whether or not the mouthpiece was present, the absolute increase in $\dot{V}_E$ was nearly equal at low and high levels of exertion.

The investigator's observations substantiate previous findings that the presence of a mouthpiece increases tidal volume and inspiratory time and decreases respiratory frequency (Askanazi et al. 1980; Hirsch and Bishop 1982; Perez and Tobin 1985). The mechanism for these effects is unknown. Drs. McCool and Paek state that it is unlikely that the mouthpiece and tubing of the spirometric apparatus added extra work to breathing because the large tubing of the equipment contributed little flow resistance or dead space volume.

An important consequence of this finding is that investigators who have used a mouthpiece for determining ventilation during certain activities or under specific physiological conditions may have overestimated ventilatory levels. Such overestimates may be particularly relevant for low levels of activity because most daily activities of the majority of the population occur at rather low levels of ventilation. The potential for such errors should be noted by investigators planning future field studies in which ventilation and doses of inhaled air pollutants will be estimated.

Evaluation of Heart Rate as a Predictor of Ventilation

Evaluating the validity of using heart rate as a predictor of $\dot{V}_E$ was an important facet of this study because it served as an important cross-validation for similar data obtained by Dr. Samet and his colleagues. The fidelity of the relationship between heart rate and $\dot{V}_E$ and the subsequent correlation of this data with time-activity data recorded in subject diaries were the core of the future epidemiological studies proposed by Samet and his colleagues.

Heart rate and $\dot{V}_E$ data were collected from all of the field study subjects during a progressive work load cycling exercise and were used to evaluate the validity of the relationship between these two variables. A significant finding from these experiments was that subject-to-subject variability existed in the heart rate–ventilation relationship. This finding indicated the importance of determining a calibration curve for this relationship for each study participant; one average curve could not be applied to all subjects. The investigators used the calibration curve from the low-range linear model from each subject to estimate $\dot{V}_E$ from the heart rate data obtained during the field study. These estimates and $\dot{V}_E$ data obtained with the BSD equipment during the calibration period of the field study were compared with pneumotachograph data. The investigators reported that, although the $\dot{V}_E$ estimates based on BSD data correlated well with the values of $\dot{V}_E$ measured with the pneumotachograph, the estimates of $\dot{V}_E$ based on heart rate did not correlate well (Table 7).

The investigators provided a critical evaluation of the heart rate–ventilation relationship by using three different mathematical approaches to analyze their data from the field study. The results from the $\dot{V}_E$-HR(full), $\dot{V}_E$-HR(low), and $\dot{V}_E$-HR(exp) analytical approaches provided a useful framework for comparing the impact of using these different methods to fit the data and for comparing the outcomes with the $\dot{V}_E$ data derived from the BSD model. Two examples demonstrate the usefulness of such comparisons. First, results from all three analytical approaches indicated that the mouthpiece produced significant increases in ventilation at both the high- and low-range heart rates; however, this effect was more prominent at the low heart range with the $\dot{V}_E$-HR(low) approach. Second, Figure 8 clearly demonstrates that the $\dot{V}_E$ values derived from the BSD model matched well with the heart rate data collected during the same time interval. In comparison, the $\dot{V}_E$ values calculated from both the $\dot{V}_E$-HR(low) and $\dot{V}_E$-HR(exp) approaches did not parallel these same heart rate data as well. To further demonstrate this point, the $\dot{V}_E$ values derived from the $\dot{V}_E$-HR(full) approach were only 45% of those values determined from the BSD model; in some cases, the full-range approach yielded negative values of ventilation.

The investigators anticipated that the $\dot{V}_E$-HR(low) model would provide the best fit for the ventilation calculations derived from the heart rate data. This prediction was based on the observation that the subjects' activity levels during the field study were lower than those imposed on them during the exercise sessions in the laboratory. From the results of the field study, the investigators found that the $\dot{V}_E$ data estimated from the BSD model correlated well with $\dot{V}_E$ values obtained directly from the pneumotachograph during the classroom session (R^2 = 0.87) (Table 7). Contrary to their expectations, the investigators found that the $\dot{V}_E$-HR(low) model yielded a poor correlation with the pneumotachograph data (R^2 = 0.42). The investigators attributed this discrepancy to deep breaths requested from the subjects during the classroom session for calibrations between the BSD equipment and the pneumotachograph. The effect of these deep breaths was to increase ventilation, but not heart rate, thus uncoupling the relationship between heart rate and ventilation for the low-range calibration curve.

The investigators also noted that a distinct advantage of the BSD model was that brief transients in breathing patterns detected by the BSD equipment were not detected by heart rate monitoring. Even at higher levels of work task activity, delays occurred between increases in ventilation measured by the BSD equipment and corresponding increases in heart rate. The investigators noted that such changes in breathing patterns and ventilation rates would be missed if calculation of $\dot{V}_E$ were based on heart rate data alone.

SPECIFIC AIMS AND STUDY DESIGN FOR THE SAMET STUDY

Dr. Samet and colleagues described three specific aims that focused on developing methods to estimate the levels and patterns of ventilation in subjects participating in epidemiological studies. Their study population included 31 healthy adults and 12 children drawn from the general population, 5 individuals with asthma, and 10 hospital patients with heart and lung diseases. As clearly noted in the report, subject selection was not directed toward gathering a group of individuals that would be representative of the general population.

The first specific aim was to evaluate the relationship between heart rate and ventilation for each of the 58 subjects in the study. The investigators determined the subject-to-subject variability for this relationship, as well as the influence of upper- versus lower-body exercise and personal characteristics, such as age, gender, fitness level, and health status, on this relationship. Heart rate and standard ventilatory measurements, such as $\dot{V}_E$, tidal volume, frequency, and oxygen uptake, were monitored with conventional equipment, including spirometers and electrocardiographs. Calibration studies were conducted in the laboratory to compare heart rate data recorded with a Heartwatch during cycling exercise with conventional electrocardiogram data recorded simultaneously. The Heartwatch is a device that combines a small transmitter worn on the chest with a wristwatch-style receiver and can calculate mean minute heart rate data. In an additional validation study, 10 ambulatory hospital patients wore the Heartwatch while their heart rates were also recorded by Holter monitoring; data from both recording devices were then compared.

The second specific aim was to characterize the route of breathing, that is, whether a subject breathed through the nose or mouth, during increasing levels of exercise intensity. Information about this partitioning of breathing at various ventilatory levels is important because breathing through the mouth bypasses the air cleansing action provided by the nose and pharynx region of the respiratory tract. As a result, oral breathing can increase an individual's dose of an air pollutant by permitting a greater fraction of inhaled particles and gases to reach the lower airways and lungs (Bennett et al. 1985). To accomplish this aim, the investigators redesigned a continuous positive airway pressure face mask so that they could sample inspiratory flow rates either through the nose or mouth while subjects cycled. The nasal airway resistance of subjects was also measured and categorized according to age, gender, and breathing difficulties associated with allergies, infections, and congestion.

The final specific aim was to test the feasibility of using data from the Heartwatch to estimate ventilation in subjects participating in a pilot field study. The investigators wanted to calculate ventilation by combining heart rate data recorded by Heartwatch with information about activity levels listed in time-activity diaries maintained by the study subjects on two separate days. Using the heart rate–ventilation relationship derived for each subject in the laboratory studies, the investigators estimated the subjects' ventilatory levels during the two-day monitoring period. The $\dot{V}_E$ data were categorized according to activity, age, gender, and health status of the subjects.

TECHNICAL EVALUATION OF THE SAMET STUDY

ATTAINMENT OF STUDY OBJECTIVES

This study attained its primary goal, which was to develop and validate methods for monitoring ventilation that could be used in large-scale epidemiologic investigations. The Heartwatch successfully recorded heart rates in field study subjects while they moved about freely for extended periods. These values were used to estimate ventilation based on the heart rate–ventilation relationship previously established for each study subject in the laboratory during exercise testing procedures. A notable strength of this study was an evaluation of the heart rate–ventilation relationships for different age and gender groups and for groups with compromised heart and lung function. An important finding was that the investigators observed substantial interindividual variability in this relationship.

The investigators also assessed the partitioning of breathing route during different work tasks. Dr. Samet and his colleagues carefully evaluated air flow rates through the oral and nasal routes in subjects while they increased their ventilation levels during exercise testing. The investigators also successfully evaluated the effects of age, nasal obstruction, and cardiopulmonary diseases on ventilatory levels.

STUDY DESIGN AND METHODS

The protocol for this study was well described and carefully executed. The investigators used conventional equipment for measuring ventilation, heart rate, and nasal airway resistance in their laboratory studies. They also specifically modified a face mask to evaluate the partitioning of ventilation between nose and mouth.

The investigators tested the variability in the heart rate–ventilation relationship among different work tasks. Several exercises were selected for the laboratory studies to simulate more closely activities that occur in the field. For lower body exercise, the subjects cycled; for upper body exercise, they pushed a vacuum cleaner and lifted a progressively weighted bag. The protocols describing these activities are well detailed.

As part of the field study, the subjects used standardized forms to record their activities during two 16-hour periods on successive days. In general, the subjects were conscientious about completing their time-activity diaries. Activities and locations recorded in the diaries were subsequently coded by two investigators using a formal coding system.

STATISTICAL METHODS

For the first specific aim, Dr. Samet and his colleagues applied linear regression to the data obtained from 58 subjects during specific work tasks to regress the logarithm of ventilation on heart rate. The slopes and R^2 values for the plotted data were compared for different subjects for the different work tasks using standard statistical methods, such as paired and unpaired t tests.

In the second protocol, subjects cycled under progressively increasing work loads while wearing a mask designed to measure nasal and oral breathing separately. Nasal airway resistance was also measured for each individual using linear regression analysis to estimate the coefficients of Rohrer's equation, which calculates pressure from air flow values. Estimated resistance at a selected flow of 0.25 L/sec was used for comparative analyses of the effects of subject characteristics on nasal airway resistance.

In the field study, ventilation was estimated from the heart rate data recorded by the Heartwatch and from entries in the time-activity diaries logged by the subjects during the two 16-hour periods of monitoring. Values for $\dot{V}_E$, $\dot{V}_E$ normalized to maximum ventilatory volume, and percentile ranks for these values were presented as distributions according to subject activity and location.

RESULTS AND INTERPRETATIONS

Dr. Samet and his colleagues evaluated the relationship between heart rate and ventilation in the laboratory during a variety of work tasks and exercise intensity levels to determine whether heart rate could serve as a surrogate for ventilation. They reported that $\dot{V}_E$ increased faster than heart rate when a subject performed an upper body exercise than when a subject performed a lower body exercise. They also observed considerable variation in the heart rate–ventilation relationship among individuals. The investigators concluded that future studies will require individual calibration of this relationship for each subject. This finding eliminated the possibility of using a single heart rate–ventilation ratio for a particular gender or age group to predict ventilation for large-scale epidemiologic studies.

The laboratory results obtained by Dr. Samet and his colleagues also demonstrated that considerable variability existed among individuals with respect to partitioning of breathing between the nose and the mouth. Predictably, the investigators reported that ventilation shifted increasingly toward the oral route as exercise intensity increased. Dr. Samet and associates suggested that the use of a face mask for these studies may have affected breathing patterns and, as a result, influenced the partitioning of ventilation between the nose and mouth. They noted that the effects of a mouthpiece on ventilatory patterns have been reported previously (Askanazi et al. 1980; Hirsch and Bishop 1982; Perez and Tobin 1985).

The investigators observed that the nasal route represented a relatively large fraction of the ventilatory route, even at relatively high levels of ventilation. For example, nasal ventilation represented more than 50% of total ventilation even when total ventilation had reached 25% of maximum voluntary ventilation. From their studies of nasal airway resistance, the investigators reported that, contrary to their expectations, resistance values were not associated with a past or current history of nasal symptoms. In addition, although the number of subjects was too few to draw clear conclusions, the investigators also observed that oral breathing was more evident among the four subjects with asthma and among those subjects with the highest values for nasal airway resistance than in subjects without these health conditions.

The field study conducted by the investigators demonstrated the feasibility of estimating ventilation from a combination of accurate heart rate monitoring and detailed activity-diary data. The investigators expended considerable effort to insure accurate collection of the heart rate data and information about the subjects' daily activities during the field study. This attention to detail yielded data that permitted calculating the time allotted by the subjects to various daily activities.

The field study yielded heart rate data and $\dot{V}_E$ estimates carefully categorized according to subject age, gender, health status, and activity. The investigators reported that children generally exhibited higher heart rates than adults, but heart rates were similar between healthy adults and those with heart and lung disease. Heart rates were often higher when subjects were outdoors, which accounted for approximately 10% of the subjects' days. Predictably, the range of $\dot{V}_E$ estimates mirrored the heart rate patterns across the various categories of activities and locations for the subjects.

In their report, the investigators discuss several physiological factors that could contribute to decreased accuracy in the ventilation estimates derived from heart rate data collected with the Heartwatch. One factor is that the ventilatory estimates were based on the heart rate–ventilation regression equation derived under conditions of maximal cycling effort. In contrast, the data from the field study indicated that heart rates of most individuals rose only modestly above basal rates during the course of the day. This result suggests that the calibration curves developed from the heart rate and ventilation data collected during laboratory testing may not be appropriate for estimating ventilatory values in the field. Future calibration curves should be based on activity levels that match those anticipated during an epidemiological study.

A second complicating factor was that the relationship between heart rate and ventilation was influenced by the type of physical activity. The investigators noted that ventilatory rates increased more rapidly than heart rates when subjects performed upper body exercise than when subjects performed lower body exercise. Based on the maximal exercise tests, Dr. Samet and colleagues reported that R^2 values exceeded 0.94 for linear regressions of the natural log of ventilation on heart rate for all groups (Table 4). In contrast, R^2 values ranged from 0.36 to 0.82 for the lifting and vacuuming tasks. In their report, Drs. McCool and Paek also noted that weaker correlations existed between ventilation and heart rate during upper body exercise than during lower body exercise (Table 2). A third complicating factor was that estimates of ventilation based on heart rate alone can miss transient changes in breathing patterns, such as occasional deep breaths or sighing. Drs. McCool and Paek observed this phenomenon during their study with the BSD model. They postulated that such missed excursions could result in underestimates of $\dot{V}_E$ based on heart rate.

Factors Contributing to Differences in the Coefficient of Determination Values for the Two Studies

Differences in the predictive accuracy of the R^2 values reported in these two studies can be attributed to three factors. First, R^2 tends to increase when variables are studied over a wide range. Thus, maximal cycling, which produced the widest range and highest values for heart rate, produced higher values of R^2 than other work tasks, particularly vacuuming. Second, both Dr. Samet and associates and Drs. McCool and Paek reported that heart rate generally remained in a low range during normal daily activities. Drs. McCool and Paek reported a mean R^2 value of 0.42 for comparisons between $\dot{V}_E$ values obtained with a pneumotachograph and those estimated using the low-range linear model (Table 7). These data were obtained during the calibration phase of the field study, while subjects sat in the classroom and did not exhibit a wide range of heart rates. The third factor was the periodic deep breaths that Drs. McCool and Paek instructed their subjects to take as part of this calibration procedure. These deep breaths reduced the predictive value of R^2 in a manner that would not occur during ordinary activities. Thus, the mean R^2 value of 0.42 obtained during the field study conducted by Drs. McCool and Paek is comparable to that obtained in the Samet field study, but may underestimate minute ventilation values that could be expected during ordinary activities.

REMAINING UNCERTAINTIES AND IMPLICATIONS FOR FUTURE RESEARCH

The findings of these two studies have yielded new insights into methods for measuring ventilation noninvasively. A clear strength of both studies was that the investigators applied their laboratory-tested methods in actual field studies. As a result, problems were discovered that might have gone unrecognized if testing had been limited to laboratory studies alone. Recognition of these problem areas provides new perspectives for improvements in the methodology that can ultimately benefit the design of future studies.

One issue highlighted by the field studies was the importance of establishing a heart rate–ventilation relationship within a range of heart rates that is appropriate for the range anticipated in subsequent field studies. Both investigator groups reported that the extended range of exercise intensity used for the calibration series in these studies did not reflect activity levels and heart rates experienced by most subjects in the field studies. As noted in the Samet report, heart rates ranged from 60 to 100 beats per minute for the majority of subjects during the field study, rather than the rates of 140 beats per minute and higher observed during cycling exercises in the laboratory. Drs. McCool and Paek observed that ventilation estimates derived from the low-range linear model of the heart rate–ventilation relationship better fit the data obtained with the BSD model. These results further support the concept that calibration studies incorporate heart rates and activity levels that are comparable to the levels expected during a proposed field study.

As noted in both reports, many investigators use cycling as a standard exercise protocol for measuring ventilatory parameters. However, many daily tasks involve a variety of maneuvers that are unlike the rhythmic motion of cycling. The type of task and whether it involves the upper or lower body influences breathing characteristics. The findings from both of these studies suggest that future studies should incorporate a broader range of work tasks if they are to reflect the diversity in breathing patterns experienced by individuals during their daily lives.

One problem area particularly highlighted by the study from Drs. McCool and Paek was that breathing through a mouthpiece or face mask produces artifacts in ventilatory measurements. Previous investigators (Gilbert et al. 1972; Askanazi et al. 1980) have reported that the presence of this apparatus can increase tidal volume and time of inspiration and decrease breath frequency. These artifacts in ventilatory measurements became apparent through comparisons between BSD ventilation measurements, which did not require a mouthpiece, and the spirometric measurements. Such artifacts complicate extrapolation of ventilatory measurements obtained with a mouthpiece or face mask in the laboratory setting to ventilation measurements obtained in community settings in which people move about freely. An additional issue that will require future evaluation is how a face mask alters the partitioning of breathing between the nose and the mouth at different ventilatory levels and how this may affect estimations of pollutant doses in the respiratory tract.

The complexity of the technical equipment required by both of these studies may pose some limitations for large-scale epidemiologic studies. Although the Heartwatch that Dr. Samet and colleagues used to monitor heart rate in their study is less cumbersome than the BSD equipment used by Drs. McCool and Paek, some malfunctioning of the watches occurred due to subject error. Simplifying the operation of these watches could reduce future subject error. One advantage of the BSD equipment is that, although it is more complicated to set up and monitor than the Heartwatch, it provides direct continuous measurements of ventilation and can detect brief transients in ventilation potentially missed by the Heartwatch. Further refinements and miniaturization of the BSD equipment could facilitate the use of this methodology for larger occupational or epidemiological studies.

One recommendation for future studies using the BSD model would be to include obese subjects in the study pool so that investigators could identify potential signal interference problems caused by adipose tissue. Another recommendation for the study population would be to include subjects from a broader age range as well as more females, so that differences in results associated with age or gender could be evaluated. In addition, investigators could determine whether breast tissue interferes with signal recordings on the Respitrace in females.

Finally, future studies could improve exposure assessment information by combining these noninvasive measurements of ventilation with data from personal monitors. Personal monitors are designed to measure the amounts of specific pollutants to which individuals have been exposed. Data from the monitors could be combined with time-activity data to calculate better dose data (Sexton and Ryan 1988). One current obstacle to this type of assessment is the limited capability of personal monitors to measure pollutant concentrations reliably and accurately. To address this issue, HEI has recently supported three studies to evaluate the sensitivity of several different personal ozone monitors. The results of those studies will be presented in a future volume of this Research Report series.

CONCLUSIONS

Accurate estimation of inhaled air pollutant dose requires information about ambient pollutant concentration, exposure duration, and ventilation. However, estimates of inhaled pollutant dose most often have been based on ambient air levels and exposure duration alone because ventilatory data for exposed individuals generally have been unavailable. Through their studies, Drs. McCool and Samet developed and tested methods for measuring ventilation noninvasively so that ventilatory data could be obtained. Dr. McCool estimated ventilation by measuring dimensional changes in the chest wall and abdomen, whereas Dr. Samet estimated ventilation from heart rate.

A strength of both studies was that ventilation was measured under a variety of work tasks and work loads. Both groups of investigators reported that different work tasks, such as cycling, lifting, pulling, and vacuuming, produced variable ventilatory patterns. Drs. McCool and Paek observed better statistical correlations between ventilation and heart rate when subjects performed lower body exercise than when subjects performed upper body exercise. Dr. Samet and his colleagues noted that, compared with lower body exercise, ventilation increased more quickly than heart rate when subjects performed upper body exercise. Drs. McCool and Paek also corroborated previous reports that the presence of a mouthpiece alters ventilation.

The two investigator groups drew different conclusions regarding the reliability of using heart rate as an estimator for ventilation. Drs. McCool and Paek noted that extrapolations based on heart rate alone miss brief aberrations in ventilation that occur during low ranges of activity. These missed aberrations could lead to underestimates in $\dot{V}_E$. Drs. McCool and Paek concluded that their BSD approach could detect these minor aberrations and improve estimates of ventilation. In comparison, Dr. Samet and colleagues concluded that heart rate measured with a Heartwatch could serve as an appropriate estimator for ventilation, particularly during the low levels of activity that constitute the majority of an average day. They did not address the question of missed ventilatory aberrations posed by Drs. McCool and Paek. Dr. Samet and colleagues did note that the BSD equipment would require miniaturization and reductions in per-subject cost before it could be used in epidemiological studies involving large numbers of subjects.

Both groups of investigators acknowledged the necessity of establishing a ventilation–heart rate calibration curve for each subject in future studies. Due to the substantial intersubject variability in this relationship observed by both groups, the investigators agreed that calibration curves based on data averaged from one group of subjects could not be applied reliably to other studies using different subjects. This conclusion implies that future epidemiological studies using the Heartwatch will, as a first step, require laboratory testing of each study subject to establish an appropriate calibration curve. Both groups also recommended that future studies should base these calibration curves on levels of activity that are commensurate with those anticipated for the study subjects.

In summary, these investigators tested methods for measuring ventilation noninvasively; their focus was to improve the accuracy of estimating doses of air pollutants delivered to the respiratory tract. Because dose can influence response, such dose estimates are essential for correctly interpreting lung function alterations caused by inhaled pollutants and for predicting the risk of lung disease in healthy individuals and more susceptible individuals in the population. Both investigator groups provided important new data for ventilation and heart rates in subjects while they moved about freely in their work and home environments. Whereas much of the existing data for these end points has been gathered during laboratory studies, these reports provide data collected under actual occupational and community settings. Although the BSD equipment imposes some restrictions on mobility, Dr. McCool's approach may find ready application in small occupational studies. In comparison, the range of mobility provided by the Heartwatch indicates that Dr. Samet's approach could be suitable for epidemiologic studies if a $\dot{V}_E$-HR calibration curve is established for each study subject. Application of these methods can improve estimates of inhaled pollutant dose and advance risk assessment methodology.

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