Dynamic Ventilation and Power Output of Urban Bicyclists

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Bicyclists' intake of air pollutants is linked to physical exertion levels, ventilation rates, and exposure concentrations. Whereas exposure concentrations have been widely studied in transportation environments, there has been relatively scant research linking on-road ventilation with travel conditions and exertion levels. This paper investigates relationships between power output, heart rate, and ventilation rate for urban bicyclists. Heart rate and ventilation rate were measured on-road and combined with power output estimates from a bicycle power model. Dynamic ventilation rates increased by 0.4% to 0.8% per watt of power output, with a mean lag of 0.8 min. The use of physiology (ventilation) monitoring straps and heart rate proxies for dynamic on-road ventilation measurements is discussed. This paper provides for a clearer and more quantitative understanding of bicyclists' ventilation and power output, which is useful for studies of pollutant inhalation risks, energy expenditure, and physical activity.

Active travelers experience conflicting health effects from physical activity on urban streets. Increased regular physical activity leads to well-established health benefits. At the same time, greater physical exertion leads to increased ventilation, and, in turn, greater inhalation of traffic-related air pollution (*1*). Although high ventilation rates for bicyclists are documented in the literature, existing studies of pollutant inhalation analyzed and reported on ventilation rates by mode or trip (*2*). Little is known about how bicyclists' ventilation varies with travel conditions and over the course of a trip.

The pollutant inhalation rate I is the product of the exposure concentration (C) and ventilation rate ($\dot{V}_E$). $\dot{V}_E$ (also called minute ventilation) is the product of the breathing frequency (f_b) and tidal volume (V_T). Hence, inhalation rate (in mass per unit time) is calculated with the following:

$$I = C \cdot \dot{V}_E = C \cdot f_b \cdot V_T$$

where C is in mass per volume of air, $\dot{V}_E$ is in volume of air per unit time, f_b is in breaths per unit of time, and V_T is in volume of air per breath.

Beyond inhalation rate, particle deposition and location of gas absorption in the respiratory tract are affected by the relative values of f_b and V_T, in addition to other factors, such as fraction oral breathing (*2*).

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Energy expenditure or power output is a key factor determining respiration and ventilation. Low to moderate levels of energy expenditure utilize aerobic respiration, which requires inhalation of oxygen. Up to the anaerobic threshold, the ventilation rate $\dot{V}_E$ is closely related to the volume rate of oxygen inhalation ($\dot{V}_{O_2}$). $\dot{V}_E$ increases primarily by an increase in V_T at lower levels of exertion, then, increasingly, by f_b. At 70% to 80% of peak exercise level, f_b becomes the dominant factor, although professional bicyclists can achieve a greater effect through V_T (*3, 4*).

One previous study directly measured dynamic on-road ventilation rates while bicycling for the purpose of pollutant dose estimation, although analysis of ventilation was not provided (*5*). That study used a face mask system to measure ventilation—a method also used in other on-road (*6*) and laboratory (*1*) study settings. Another approach has been to estimate dynamic on-road ventilation rate ($\dot{V}_E$) from measured heart rate (HR), on the basis of laboratory-derived $\dot{V}_E$ ~ HR relationships for individual subjects (*7, 8*). Laboratory $\dot{V}_E$ measurements typically use a bicycle ergometer (stationary bicycle) and a face mask.

Figure 1 illustrates the connection between bicyclist ventilation and travel conditions. A rider's energy expenditure affects heart and ventilation rates, mediated by an individual subject's physiology (and, to a lesser degree, other variables such as air density). At the same time, the energy expenditure above baseline or resting metabolic rate leads to a commensurate energy transfer to the bicycle, mediated by bicycle attributes and the style of riding (pedaling cadence, upper body control, and so on). The energy transferred to the bicycle produces a certain travel speed, depending on bicycle, roadway, and travel attributes that determine energy state changes and losses.

The focus of this study is variation in bicyclist ventilation during riding. Hence, subject-specific variables are assumed to be constant over the course of a trip and are grouped into a "Subject" factor. Then the connection between ventilation and travel conditions can be made in two steps: first, estimate energy transferred to the bicycle on the basis of travel and roadway conditions, and, second, model ventilation as a function of energy transferred to the bicycle, mediated by the subject. The objectives of this paper are to

1. Describe and validate a new approach to measure on-road ventilation rate using an unobtrusive and economical chest strap and
2. Analyze bicyclists' dynamic ventilatory response to power output while bicycling, as determined by roadway and travel conditions.

The goal of this research is to provide a clearer and more quantitative understanding of on-road ventilation and power output for urban bicyclists. Quantifying the relationship between on-road ventilation and travel conditions (road grade, speed, acceleration, and so on) will be useful for future studies of pollutant inhalation by bicyclists as well as studies of energy expenditure and physical activity.

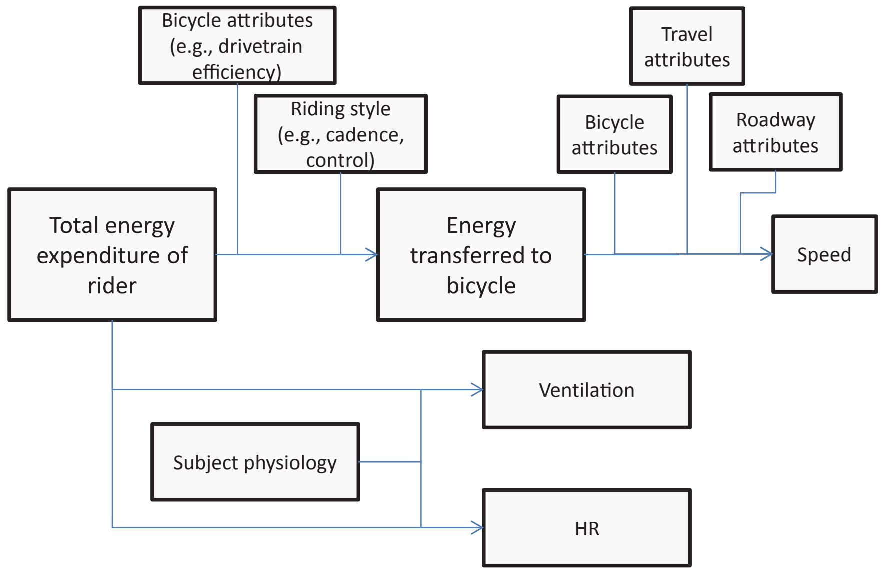

FIGURE 1 Conceptual diagram of the connection between bicyclist ventilation and travel conditions.

METHOD

Data Collection

On-road data were collected in Portland, Oregon, on nine days between October 2012 and September 2013. Approximately 55 person-h of data were collected, with each subject riding 2 to 4 h per day on days on which he or she participated. All data were collected near the morning peak period (7 to 10 a.m.). A variety of roadway facilities were included in prescribed routes, including off-street bicycle and pedestrian paths and mixed-use roadways that ranged from local roads to major arterials. The study subjects were volunteers instructed to adhere to safe riding practices, to follow traffic laws, and to ride at a pace and exertion level typical for utilitarian travel (i.e., commuting).

Three subjects participated in the data collection; this was considered adequate because the primary focus of the study involved travel covariates rather than intersubject covariates. The subjects were recruited from Portland State University's student body. Approval for the research was obtained from Portland State University's Human Subjects Research Review Committee. All subjects were nonsmokers who reported moderate regular physical activity and good respiratory health [on the basis of the American Thoracic Society respiratory disease questionnaire (*9*)]. The characteristics of Subjects A, B, and C were as follows (respectively): male, male, and female; aged 34, 28, and 45; average bicycle weight (including all gear), 25, 22, and 23 kg; and average post-ride body weight, 80, 70, and 75 kg.

GPS receivers recorded 1 Hz location data with time stamps. Redundant GPS devices and simultaneous on-bicycle video were used to cross-check the location data for reliability. Meteorological variables were also measured for context. Temperature and humidity were measured on-road with a HOBO U12 data logger attached to the bicycle. Wind data were retrieved from an Oregon Department of Environmental Quality monitoring station in the data collection area (Station SEL 10139).

To calculate grade, elevation was extracted from archived data (1-m digital elevation maps based on lidar) and differentiated in two dimensions. Grade of travel (G) was calculated as $G = \Delta\text{elevation}/\text{distance}\ 100\%$, using 1 Hz elevation and location data. Grades of more than or less than 25% were removed (0.3% of grade data), and a smoothing algorithm was applied (5-s moving average).

Physiology Monitoring

Heart rate and breathing rate were measured by a physiology (ventilation and heart rate) monitoring strap (BioHarness 3) worn around the bicyclist's chest. The BioHarness 3 is a relatively new commercial device for mobile physiological monitoring. Data were logged at 1 Hz, and a custom Android application was written to log the BioHarness data stream with simultaneous GPS data on a smartphone.

The BioHarness band stretches around the chest and contains a conductive elastic fabric. Expansion of the chest is monitored by measuring the resistance in the conductive fabric. The breathing rate (f_b) is assessed by detecting inflections in the resistance waveform. The BioHarness also reports a raw breath amplitude (B_A) value in volts that is "indicative." Because the measured resistance changes with the expansion of the chest, there should be a relationship between breath amplitude, B_A, and V_T. However, the relationship between B_A and V_T will likely depend on the location and tightness of the strap. By calibrating B_A to V_T each time the BioHarness was used, session-specific $B_A \sim V_T$ relationships were estimated and used to calculate dynamic V_E from on-road measured f_b and B_A. The BioHarness data fields used in this research were

1. HR (from electrocardiograph sensors),
2. Heart rate confidence (in percentage),
3. f_b, and
4. B_A.

Tidal Volume Calibration

A tidal volume calibration was conducted by each subject at the beginning and end of each data collection period. The tidal volume calibration consisted of 30 to 60 s of steady ventilation at prescribed tidal volumes of 500, 1,000, 1,500, and 2,000 mL. An incentive spirometer (DHD222500, Medline) was provided to the subjects to monitor tidal volume. The first 10 s of B_A readings at each tidal volume were discarded and the remaining B_A values averaged for each tidal volume. A curve was fit to each set of calibration data with the use of the equation $V_T = a + b \cdot B_A$. Calibration periods with missing data or a statistical fit of $R^2 < .75$ were discarded (four calibration periods with poorly fitted straps or inconsistent tidal volumes). Median coefficients for the calibration curves were $a = -0.5702$ and $b = 16.454$ (V_T in liters and B_A in millivolts).

On-road V_T was estimated from B_A measurements by applying the calibration curve $V_T = a + b \cdot B_A$ with calibration parameters a and b interpolated between the before and after calibration periods for each data collection. Data collections without calibration data at one end (before or after) used a single set of calibration parameters. Minute ventilation was then calculated ($\dot{V}_E = V_T f_b$). Observations were filtered with the following constraints:

- BioHarness reported HR confidence value of ≥80%,
- B_A values within the range of calibration data,
- $1 < f_B < 100$, and
- $20 < \mathrm{HR} < 200$.

Some 50,241 observations (23%) did not meet these constraints or were missing data. The processed physiological data set included 165,473 1-s data points (46 h).

Ergometer Testing

Physiological attributes of the subjects were assessed with a standard bicycle ergometer exercise test (*4*). Tests were conducted on bicycle ergometers (New Bike Exc 700, Technogym) on September 12, 2013. The protocol was a 3-min incremental power output of 50 W from 0 W to volitional exhaustion—which was 350, 250, and 200 W for Subjects A, B, and C, respectively. Self-selected cadences were around 70 rpm.

Physical Model of Bicyclist Power Output

A first-principles physical model was used to estimate bicyclist power output from measured roadway and travel characteristics. Olds provides a review of bicycle energy and power models (*10*). Beyond accounting for changes in energy state caused by speed and elevation, almost all power demand models include aerodynamic drag and rolling resistance terms. Some models include other factors in varying level of detail, such as angular momentum of the wheels and the rider's limbs, spoke drag, turbulence around the pedals, rolling resistance sensitivity to grade, and varying air density (*11–16*).

The energy state of a bicycle + rider system is the sum of its potential energy (PE) and kinetic energy (KE). The energy flux balance for the bicycle + rider system is

$$W_M - W_L - W_B = \Delta\mathrm{KE} + \Delta\mathrm{PE} \tag{1}$$

where

W_M = mechanical work input from the bicyclist;
W_B = energy dissipated through braking (as heat);
W_L = other energy lost through drag, rolling resistance, friction;
ΔKE = change in kinetic energy; and
ΔPE = change in potential energy.

W_M is related to the total bicyclist-generated external work by an efficiency of power transfer to the bicycle (including losses in the drivetrain and energy used for upper body control). W_M and W_B are difficult to measure directly and unavailable in the study data set. KE and PE can be estimated from speed, weight, and elevation data, and W_L can be estimated from the literature with the assumption of certain parameters.

The net work on the bicycle + rider system is defined as $W_N = W_M - W_B$. The following assumptions—

1. $W_B \geq 0$ (i.e., brakes only remove energy from the system),
2. $W_M \geq 0$ (i.e., the bicyclist can only input energy to the system), and
3. $W_M = 0 | W_B = 0$ (i.e., the bicyclist is never pedaling and braking at the same time)

—then lead to

$$W_M = \begin{Bmatrix} W_N & W_N > 0 \\ 0 & W_N \leq 0 \end{Bmatrix} \tag{2}$$

Additionally, $W_B = W_N$ when $W_N \leq 0$ and $W_B = 0$ otherwise. With work in units of energy, the time rates of work and energy transfer are in units of power (e.g., watts). From the bicycle energy literature (*13*), neglecting spoke drag, rotational inertia of the wheels, and bearing losses, and assuming relatively low wind speeds and grades, energy transfer rates per unit time (t) are

$$\frac{\Delta\mathrm{KE}}{\Delta t} = \frac{m_T}{2}\frac{\Delta v_b^2}{\Delta t}$$

$$\frac{\Delta\mathrm{PE}}{\Delta t} = v_b m_T g G$$

$$\frac{W_L}{\Delta t} = \frac{1}{2}\rho C_D A_F v_b^3 + v_b C_R m_T g$$

where

m_T = total mass of the bicycle + rider system,
v_b = ground speed of the bicyclist,
g = acceleration caused by gravity,
G = grade of travel,
ρ = air density,
C_D = drag coefficient,
A_F = frontal area of the bicyclist (assuming 0 yaw angle), and
C_R = coefficient of rolling resistance.

A modified drag coefficient is defined as $C_D' = 1/2\, \rho C_D A_F$, leading to a rate of net work ($\dot{W}_N$) of

$$\dot{W}_N = \frac{W_N}{\Delta t} = \frac{\Delta\mathrm{KE} + \Delta\mathrm{PE} + W_L}{\Delta t}$$

$$\dot{W}_N = \frac{m_T}{2}\frac{\Delta v_b^2}{\Delta t} + v_b m_T g G + C_D' v_b^3 + v_b C_R m_T g \tag{3}$$

All of the parameters needed to calculate $\dot{W}_N$ are measured in the study data set except C_D' and C_R, for which there is information in the literature.

TABLE 1 Parameters Used in Calculating Bicyclist Power

Parameter	Subject A	Subject B	Subject C	Source
m_r (kg)	80	70	75	Measured; mass of the rider
m_T (kg)	105	91	97	Measured; includes rider and bicycle
Height, H (cm)	189	175	163	Measured; standing
Surface area of rider, A_S (m^2)	2.32	2.07	2.02	Olds et al. (*14*); $A_S = H^{0.725}\, m_T^{0.425}\, 0.007184$
Frontal area of rider, A_{Fr} (m^2)	0.59	0.51	0.49	Olds et al. (*14*); $A_{Fr} = 0.3176A_S - 0.1478$
Frontal area of bicycle, A_{Fb} (m^2)	0.12	0.12	0.12	Olds et al. (*14*)
Frontal area inflation factor, F	1.2	1.1	1.1	Assumed; loose clothing, upright position, panniers, and equipment
Total frontal area, A_F (m^2)	0.85	0.69	0.67	$A_F = F(A_{Fr} + A_{Fb})$
C_D	1.1	1.0	1.0	Wilson (*16*)
ρ (kg/m^3)	1.23	1.23	1.23	Assumed; sea level, 15°C
C'_D	0.6	0.4	0.4	$C'_D = \frac{1}{2}\rho C_D A_F$
C_R	0.004	0.004	0.004	Wilson (*16*)
Maximum power output (W)	300	250	200	Ergometer testing; <3 min

Table 1 shows power output parameters applied for the three study subjects, including measured values and estimates informed by the literature. All three subjects had 700c commuter-style (semislick) tires, 25 mm to 28 mm. Subjects A and B rode touring bicycles; Subject C rode a more upright, "city" bicycle. All three subjects rode with rear panniers, although Subject A also had a large trunk box holding sample bags and air sampling equipment mounted in a front basket. These additions increased both the frontal area and the drag coefficient for Subject A. All three subjects rode in touring or upright positions. The values in the following table for the unmeasured parameters are estimates from several sources in the literature, especially Olds et al. (*14*) and Wilson (*16*).

Power output or rate of work estimates ($\dot{W}_M = W_M/\Delta t$) were made for each subject with the use of Equations 2 and 3 with on-road speed and grade data and the parameters in Table 1. $\dot{W}_M$ was constrained to the maximum power output from ergometer testing in Table 1. Power output was also calculated in units of MET, which is a standardized unit of metabolic energy expenditure that is normalized to body mass and resting metabolic rate. Resting activities are at a MET of 1. Standard MET values are calculated with respect to a resting metabolic rate of 3.5 mL O_2/minute/kg of body mass. The American College of Sports Medicine's equation for oxygen consumption during bicycling (in milliliters of oxygen per kilogram per minute) is

$$\dot{V}_{O_2} = 10.8\frac{\dot{W}_M}{m_r} + 7$$

with $\dot{W}_M$ in watts and m_r (body mass) in kilograms (*17*). Standard MET can then be calculated as

$$\text{MET} = \frac{\dot{V}_{O_2}}{3.5} = 3.09\frac{\dot{W}_M}{m_r} + 2$$

RESULTS

Summary statistics for physiology and power output data are shown in Table 2 for 5-s aggregated data. Ventilation volumes are presented at ambient temperature and pressure, which allows direct application

TABLE 2 Summary Statistics for Physiology and Power Output Data

Variable	Units	Minimum	1st Quartile	Median	Mean	3rd Quartile	Maximum	N
HR	min^{-1}	20	69	81	84	96	200	39,508
f_b	min^{-1}	2	16	22	22	28	51	39,508
B_A	mV	24	61	85	92	116	280	38,675
V_T	mL	0	600	889	1,002	1,275	7,238	32,471
V_E	l min^{-1}	0.0	10.3	18.0	22.4	29.7	165.6	32,471
$\dot{W}_M$								
Pooled	W	0	0	114	126	235	300	21,963
Subject A	W	0	0	126	135	265	300	16,950
Subject B	W	0	0	73	101	207	250	2,555
Subject C	W	0	0	74	90	200	200	2,458
MET								
Pooled	MET	2.0	2.0	6.5	7.0	11.2	13.6	21,963
Subject A	MET	2.0	2.0	6.8	7.2	12.2	13.6	16,950
Subject B	MET	2.0	2.0	5.2	6.4	11.1	13.0	2,555
Subject C	MET	2.0	2.0	5.0	5.7	10.2	10.2	2,458

for inhalation rate estimates. A mean ventilation rate of 22.4 L/min is in good agreement with past studies of bicyclist inhalation (*2*). The average sampling conditions were 17 km/h travel speed (without stops), 19°C temperature (range: 11°C to 25°C), 75% relative humidity (range: 57% to 91%), and wind speed 1.8 m/s (range: 0.6 to 3.6 m/s).

The calculated MET values agree well with published research. The *Compendium of Physical Activity* lists 16 different types of bicycling as activities with assumed static energy expenditures that range from 3.5 MET for "leisure" bicycling (5.5 mph) to 16 MET for competitive mountain bicycle racing (*18*). "General" bicycling is at a MET of 7.5 and bicycling "to/from work, self selected pace" is at a MET of 6.8 in the *Compendium.* Other research has reported typical nonracing bicyclist METs of 5 to 7 (*15, 19, 20*).

Ventilation and Heart Rate

The lagged covariance between ventilation and heart rate was calculated with the use of 1-s data. The covariance peaks at 20 s, indicating that heart rate changes lead ventilation changes by around 20 s. This lag is relevant to research designs that use on-road measured HR to predict dynamic ventilation rates.

The relationship between ventilation and heart rate was modeled as

$$\ln\left(\dot{V}_E\right)_i = \alpha + \beta \cdot \mathrm{HR}_{i-4}$$

with 5-s data, where HR_{i-4} is heart rate lagged by four periods (4 lags = 20 s) and α and β are fit parameters. Pooled and subject-segmented ordinary least squares models were estimated with Newey-West heteroscedasticity and autocorrelation consistent (HAC) robust standard error estimates. The estimated model results by subject and pooled are shown in Table 3. All coefficients are significant at $p < .01$. Because of serial correlation, using unlagged heart rate (HR_i) as the independent variable generates similar models but with higher standard errors.

The estimated β coefficients in Table 3 are in line with the literature, which suggests central values of 0.016 to 0.023 for bicyclist $\ln\dot{V}_E \sim \mathrm{HR}$ slope coefficients, heterogeneous to individuals (*1, 19, 21, 22*). Mermier et al. report slopes ranging from 0.016 to 0.029 for 15 healthy men who performed maximum exercise tests on ergometers (*8*). The ventilation–heart rate models provide validation support for the BioHarness-based estimation of on-road ventilation. The model fits (R^2) in Table 3 are lower than past reported values from laboratory ergometer studies (*1, 8*), which is attributable to greater measurement error in the indirect field measurements of ventilation rate (using the BioHarness chest strap with spirometer calibrations) than the direct laboratory measurements of ventilation rate (using face masks and pneumotachometers).

TABLE 3 Model Parameters Relating Ventilation to Heart Rate

Parameter	Subject A	Subject B	Subject C	Pooled
α	.406	.159	1.487	.782
β	0.0298	0.0271	0.0156	0.0244
N	23,127	5,053	4,291	32,471
R^2	0.371	0.239	0.151	0.290

Power Analysis

The application of the power equations allows the power demands on the bicyclists to be broken down by terms. The net energy attributable to each power term follows:

- Kinetic energy flux (ΔKE) = 0 kW,
- Potential energy flux (ΔPE) = −155 kW (net elevation loss),
- Aerodynamic drag loss = 1,792 kW, and
- Rolling resistance loss = 403 kW.

Cumulative wattage by power equation term was also calculated for observations with complete power data (some observations were missing grade data, so the ΔPE term was not available). Of the 39,508 5-s periods in the data set, 21,963 had complete power data, with total energy expenditure of the riders of 3,908 kW. This energy (plus the input of 155 kW of PE) was dissipated as 43.5% aerodynamic drag, 9.7% rolling resistance, and 46.8% braking.

The bicyclists were performing pedaling work ($W_N > 0$) for 14,978 (68%) of the complete observations (20.8 h). Isolating those periods when the riders were pedaling, the individual sums of energy for the other terms of the power equation were 54.5% kinetic energy, 2.2% potential energy, 35.7% aerodynamic drag, and 7.7% rolling resistance. In other words, when a bicyclist was pedaling, 43% of the energy input was immediately dissipated as drag and rolling losses (maintaining speed) and the other 57% went to usable, recoverable energy (primarily as speed, but also as elevation).

Ventilation and Power Output

Lagged covariance between $\dot{W}_M$ and HR and between $\dot{W}_M$ and $\dot{V}_E$ was calculated with 5-s aggregated data (a 5-s moving average was used to estimate grades). Covariance between $\dot{W}_M$ and HR peaks at one lag (5 s), and covariance between $\dot{W}_M$ and $\dot{V}_E$ peaks at six lags (30 s). Thus, the physiological response to increased power output is fast in heart rate and slower in ventilation. Again, this is relevant for study designs where ventilation is not measured directly but estimated from heart rate or power output.

An unconstrained distributed lag model of ventilation on power output was specified out to 30 lags (2.5 min):

$$\ln\left(\dot{V}_E\right)_t = \alpha + \sum_{i=0}^{30} \beta_i \dot{W}_{M,t-i} + \varepsilon_t$$

with $\dot{V}_E$ in liters per minute, $\dot{W}_M$ in watts, and ε_t an independently and identically distributed error term. Longer lags were explored but found to be not significant. The model was estimated separately for each subject, with Newey-West HAC robust standard error estimates. The cumulative effect of $\dot{W}_M$ on $\dot{V}_E$ is represented by $\beta_T = \sum_{i=0}^{30} \beta_i$.

Estimated subject-specific and pooled model results are shown in Table 4. As in Table 3, low R^2 values are attributable to measurement error in the indirect field measurements of ventilation rate, in addition to estimated energy transfer rates. The left plot in Figure 2 shows the marginal impact of $\dot{W}_M$ on $\dot{V}_E$ as $\beta_i \cdot 100\%$ (versus lag in seconds, $5i$). The right plot in Figure 2 shows the cumulative lagged impact of $\dot{W}_M$ on $\dot{V}_E$, calculated at lag L as $\sum_{i=0}^{L} \beta_i/\beta_T \cdot 100\%$.

The plots in Figure 2 show that the majority of the effect of power output on ventilation is realized within the first minute. The mean lag (the time period at which half of the effect of $\dot{W}_M$ on $\dot{V}_E$ is achieved,

TABLE 4 Distributed Lag Models of On-Road Ventilation as a Function of Power Output

Parameter	Subject A	Subject B	Subject C	Pooled
α	2.185	2.674	2.318	2.348
β_T	0.00744	0.00417	0.00761	0.00645
Number of significant lags ($p < .05$)	28	10	11	26
N	13,044	2,248	2,156	17,448
Adjusted R^2	.154	.024	.111	.140
F-statistic	77.36	2.76	9.72	92.36

computed as $\Sigma_{i=0}^{30} i * \beta_i/\beta_T$) was 0.56 to 0.85 min for individual subjects and 0.78 min in the pooled model. The median lag (the lag at which $\Sigma_{i=0}^{L} \beta_i/\beta_T \approx 0.5$) was 0.58 to 0.83 min for individual subjects and 0.75 min in the pooled model. The lag values compare well with previous studies that found around 50% adaptation of ventilation to exercise after the first minute, with some intersubject variability (*23, 24*).

Figure 3 illustrates the sensitivity of the $\dot{V}_E \sim \dot{W}_M$ relationship to the energy equation parameters C_D' and C_R. The three plots in Figure 3 show modeled β_T as shadings over a wide range of values for C_D' and C_R, for each subject. Note the different color scales in each figure, centered near the β_T estimate in Table 4. The selected ranges for C_D' and C_R rely on the literature used in Table 1 (*13, 16*). The $\dot{V}_E \sim \dot{W}_M$ relationship is more sensitive to C_D' than C_R. Higher values of these power equation parameters increase estimates of on-road $\dot{W}_M$ and so reduce the size of β_T. The modeled β_T is within 0.001 of the initial estimate over a wide range of parameter values.

Comparison with Theory

The β_T values in Table 4 are consistent with expectations from physiology. Oxygen demand ($\dot{V}_{O_2}$) increases with power output at around 10 to 12 mL O_2/min/W (*4, 10, 25–27*). Zoladz et al. (*27*) found that $\dot{V}_{O_2}$ increases nonlinearly at power output over 250 W. This slope reflects a unit conversion of 1W = 2.86 ml O_2/min and a human mechanical cycling efficiency (the amount of energy derived from atmospheric oxygen that is translated to external work) of ~25% (*3, 16, 28*).

The relationship between V_E and $\dot{V}_{O_2}$ has been modeled as both linear and exponential, with better fits over a wide range of $\dot{V}_{O_2}$ using exponential forms. The exponential form, ln $\dot{V}_E \sim \dot{V}_{O_2}$, has been estimated with a slope of around 1.2 (*29–31*). Models using the oxygen uptake efficiency slope (OUES), defined as $\dot{V}_{O_2} = \text{OUES} \cdot \log_{10} \dot{V}_E + \mu$, can be converted to a ln $\dot{V}_E \sim \dot{V}_{O_2}$ slope coefficient by calculating ln 10/OUES. Typical OUES values are around 1.8 to 2, increasing

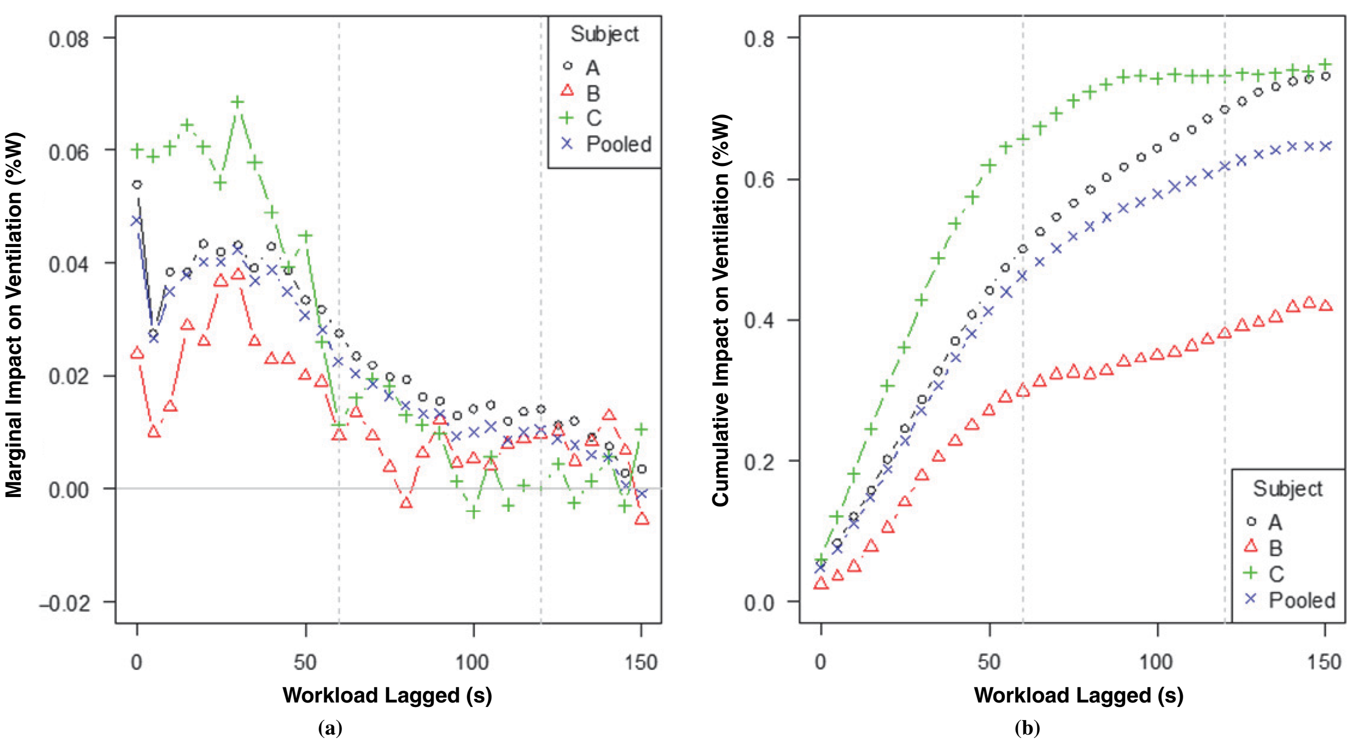

FIGURE 2 Power output on ventilation: (*a*) marginal impact and (*b*) cumulative impact.

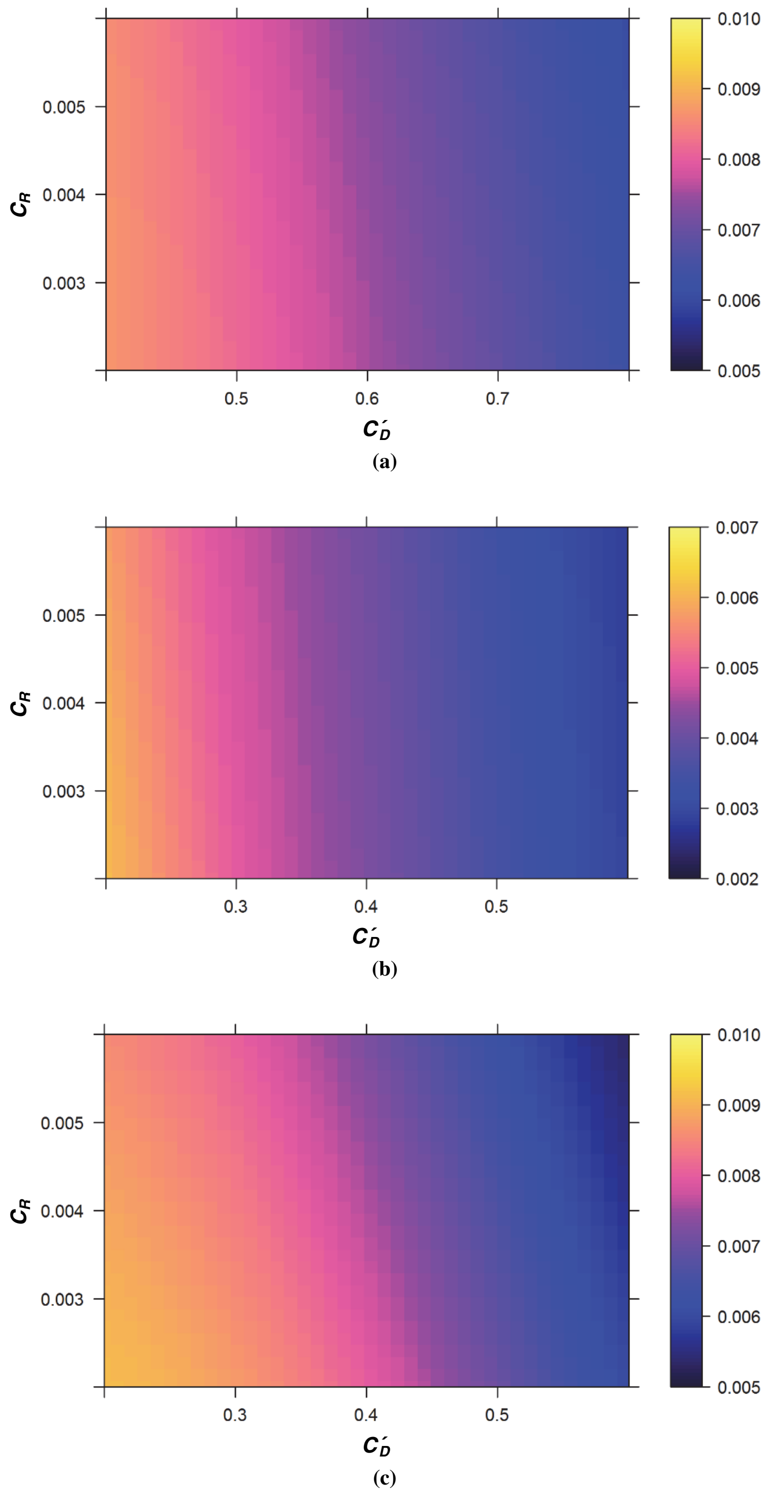

FIGURE 3 Sensitivity of modeled β_T to power equation parameters C_D' and C_R for (*a*) Subject A, (*b*) Subject B, and (*c*) Subject C.

with cardiac fitness. In linear form, the ventilatory equivalent for oxygen ($\dot{V}_E/\dot{V}_{O_2}$) during moderate exercise is around 20 to 30 (*32–34*). Assuming a linear ventilatory equivalent of 25 (*35*), at ventilation rates of 20 to 50 L/min during exercise, the semielasticity of $\dot{V}_E$ to $\dot{V}_{O_2}$ (i.e., the slope of ln $\dot{V}_E \sim \dot{V}_{O_2}$) would be expected to be around 0.5 to 1.3.

The slope of ln $\dot{V}_E \sim \dot{V}_{O_2}$ can be converted to ln $\dot{V}_E \sim \dot{W}_M$ using the factor 0.01 (LO_2/min/W), resulting in expected ln $\dot{V}_E \sim \dot{W}_M$ slopes of roughly 0.005 to 0.013. Thus, the modeled values of β_T in Table 4 and the sensitivity ranges in Figure 3 are in the range of expected ventilation response to power output. The theoretical values are based on steady state relationships and ergometer testing protocols used in physiology studies. Low-range values of the ln $\dot{V}_E \sim \dot{W}_M$ slope in these data could be attributed to a muted ventilatory response to dynamic power output.

For a body mass of 75 kg, standard MET increases at 0.04 per watt $\dot{W}_M$. Thus, the expected ln $\dot{V}_E \sim \dot{W}_M$ slopes can be converted to expected ln $\dot{V}_E \sim$ MET slopes of 0.1 to 0.3. In linear form, ventilatory equivalents for oxygen ($\dot{V}_E/\dot{V}_{O_2}$) of 20 to 30 can be converted to expected $\dot{V}_E \sim$ MET slopes of 5.7 to 8.6. Ventilation versus MET relationships were estimated with 60-min aggregated data ($N = 47$). A regression of ln$\dot{V}_E$ on MET generates a slope coefficient of 0.22 ($p < .01$, $R^2 = .16$), and a regression of $\dot{V}_E$ on MET generates a slope coefficient of 6.5 ($p < .01$, $R^2 = .27$)—both well in line with expectations.

The ventilation versus power relationships are expected to vary some with personal characteristics. The ventilatory equivalent for oxygen $\dot{V}_E/\dot{V}_{O_2}$ (and in turn the slope of $\dot{V}_E \sim \dot{W}_M$) tends to increase with pulmonary or cardiovascular diseases (*4*), be higher in children and adolescents than in adults, and decrease with aerobic training (*4, 35, 36*). Hence, a broader population of bicyclists that included children and adults with respiratory diseases could have higher $\dot{V}_E \sim \dot{W}_M$ slope coefficients than estimated in this study. But power output would also likely vary, and a populationwide analysis of bicyclist ventilation would have to consider both aspects jointly.

Comparison with Ergometer Testing and Direct Power Measurements

The ln $\dot{V}_E \sim \dot{W}_M$ relationship was estimated for the same subjects with the use of ergometer test data. A model [$\ln(\dot{V}_E) = \gamma + \lambda \dot{W}_M$] was specified for each subject, with $\dot{V}_E$ in liters per minute, $\dot{W}_M$ in watts, and parameters γ and λ. Subject-specific and pooled models were estimated with the use of ordinary least squares with Newey-West HAC standard errors for data aggregated at each power output level from the ergometer test. Model estimation results are shown in Table 5. All coefficients were significant at $p < .01$.

The parameter estimates in Table 5 are also in range of expectation from theory and compare reasonably well with the slope parameters from on-road data shown in Table 4. The pooled model is nearly the same. In both the on-road and ergometer models, Subject B has higher baseline ventilation but less ventilatory response to power output than the other subjects. Subject C has the highest ventilatory response to power output. Subjects B and C both showed stronger ventilatory responses to power output in ergometer testing than on-road; the opposite occurred for Subject A. Differences between ergometer and on-road testing could be caused by static versus dynamic power output, errors in assumed bicycle power equation parameters, or both (Figure 3).

The bicycle for Subject A was equipped with a PowerTap G3 Hub capable of measuring power transfer to the rear wheel. The ventilation versus power relationship was estimated with the use of this smaller set of directly measured power data with the distributed lag model specification, yielding coefficient estimates of $\alpha = 2.564$ and $\beta_T = 0.00662$, with a mean lag of 0.75 min ($N = 7{,}626$, adjusted R^2 of .25). The consistency of these parameters with the previous results using modeled power output provides additional validation of the study findings.

TABLE 5 Model Parameters Relating Ventilation to Power Output from Ergometer Testing

Parameter	Subject A	Subject B	Subject C	Pooled
γ	2.512	2.550	1.815	2.328
λ	0.00628	0.00561	0.01197	0.00728
R^2	.60	.72	.71	.65

CONCLUSIONS

Physiology monitoring straps provide an unobtrusive way to measure ventilation rates for bicyclists. Monitoring straps that measure breathing can be purchased for a small fraction of the cost of a portable face mask system, are less cumbersome for participants, and enable concurrent measurement of ventilation and uptake doses. Indeed, this study is part of a larger research project that simultaneously measures ventilation and breath biomarkers of volatile organic compound uptake for urban bicyclists. Ventilation rate measurements were validated by heart rate versus ventilation rate relationships. Future work should further validate this method by direct comparison with portable face mask systems.

Average ventilation rate and power output in this study were 22 L/min and 126 W (MET of 7.0), in agreement with past studies of commuting bicyclists. The on-road ventilatory response to dynamic power output was 0.4% to 0.8% per watt, slightly lower than from ergometer testing for the same subjects and at the low end of expected values from the physiology literature. This quantification allows ventilation to be estimated directly from travel conditions (road grade, speed, and so on) and a few key bicyclist parameters (mass and coefficients of rolling resistance and aerodynamic drag), or from power output measurements generated by power meters in the rear hub, crank, or pedals.

On-road ventilation lagged heart rate by 20 s and lagged power output by 50 s. The ventilation lag of heart rate is important to consider for study designs using only heart rate monitors to estimate dynamic on-road ventilation. The ventilation lag of power output implies that ventilatory responses are not coincident with locations of energy expenditure but spread out over 1 to 2 min. Assuming bicycling speeds around 15 km/h, a lag of 50 s is equivalent to a spatial difference of 200 m. This spatial lag in the ventilatory response is a potentially important consideration for pollutant inhalation "hot spots." Exposure concentrations are expected to be elevated near intersections; power output, too, is high during an acceleration from a stop at an intersection—but the ventilatory response is spread out over several blocks. Conversely, when bicyclists are stopped at an intersection with a power output of 0 W, they are breathing with the residual influence of the past 2 min of exertion.

In this study, 47% of on-road energy loss was caused by braking and 44% was caused by aerodynamic drag. A more naturalistic bicycle travel data set would be needed to estimate a more representative distribution of power demands for urban bicycling. Future work will explore the influence of travel attributes on power output and ventilation in more detail, including the relative effects of stops, grades, and travel speeds, and trade-offs between power and speed for total ventilation per unit distance or per trip. This paper is an important step toward quantifying the impact travel characteristics have on bicyclists' pollutant inhalation risks.

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