Dew-point hygrometry system for measurement of evaporative water loss in infants

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1Department of Pediatrics, Stanford University School of Medicine, Stanford, California 94305-5119; 2Program in Human Biology, Stanford University, Stanford, California 94305-2160; 3Brain Research Institute, University of Los Angeles, California 90095; 4National Aeronautics and Space Administration Ames Research Center, Moffett Field, California 94035-1000; and 5Department of Mechanical Engineering, Stanford University, Stanford, California 94305-3030

Ariagno, Ronald L., Steven F. Glotzbach, Roger B. Baldwin, David M. Rector, Susan M. Bowley, and Robert J. Moffat. Dew-point hygrometry system for measurement of evaporative water loss in infants. J. Appl. Physiol. 82(3): 1008–1017, 1997.—Evaporation of water from the skin is an important mechanism in thermal homeostasis. Resistance hygrometry, in which the water vapor pressure gradient above the skin surface is calculated, has been the measurement method of choice in the majority of pediatric investigations. However, resistance hygrometry is influenced by changes in ambient conditions such as relative humidity, surface temperature, and convection currents. We have developed a ventilated capsule method that minimized these potential sources of measurement error and that allowed second-by-second, long-term, continuous measurements of evaporative water loss in sleeping infants. Air with a controlled reference humidity (dew-point temperature = 0°C) is delivered to a small, lightweight skin capsule and mixed with the vapor on the surface of the skin. The dew point of the resulting mixture is measured by using a chilled mirror dew-point hygrometer. The system indicates leaks, is mobile, and is accurate within 2%, as determined by gravimetric calibration. Examples from a recording of a 13-wk-old full-term infant obtained by using the system give evaporative water loss rates of ~0.02 mg H2O·cm−2·min−1 for normothermic baseline conditions and values up to 0.4 mg H2O·cm−2·min−1 when the subject was being warmed. The system is effective for clinical investigations that require dynamic measurements of water loss.

evaporation; water loss; sweating; sleep; infant

EVAPORATIVE WATER LOSS (EWL) refers to the total water loss through the skin, including both “sensible” (i.e., sweating) and “insensible” (i.e., vapor diffusion through the epidermis) water loss. Transepidermal water loss (TEWL) is the portion of EWL that moves through the skin by diffusion, as contrasted with liquid water ejected from the sweat glands.

EWL is an important component of heat exchange in the newborn infant because of the permeability of the stratum corneum and the infant’s relatively large surface area-to-volume ratio. EWL is a major cause of water loss and energy loss in premature infants during the first week after birth (6, 12, 14, 23, 26, 31). In older patients, accurate measurement of EWL is important to studies of burn damage and is also useful in the assessment of skin barrier function in a variety of dermatological applications (27, 29). Some studies in adult humans (15, 16, 21) have reported decreased sweat rates in rapid-eye-movement (REM) sleep, which supports the contention that inhibition of thermoregulatory mechanisms occurs during REM compared with non-REM (NREM) sleep or awake states. However, Amoros et al. (1) reported that cessation of sweating during REM was not supported in their study. Recent studies have reported that local sweating rate may be similar during REM and NREM states in premature infants during warming (4) and that evaporative water loss in term infants, aged between 2 days and 3 mo, is not different in REM and NREM states (3). Excessive sweating has been reported in siblings of sudden infant death syndrome (SIDS), and some SIDS victims had an elevated temperature at death (2, 8, 9, 13, 17, 24, 28). The higher proportion of REM in infants, compared with adults, makes the effect of sleep state on thermoregulatory capabilities important in these investigations. Studies of thermoregulation and EWL in the infant may be important to investigate abnormalities associated with SIDS.

The relationships between body temperature, sleep, and insensible loss of weight have been the subject of scientific investigations for over 50 years (7). Determinations of EWL in infants have been made by a number of techniques that have been discussed in several publications (10, 20, 25).

Our research required a system less subject to environmentally induced artifacts, one which could provide accurate, time-resolved, local measurements of EWL, with a time constant sufficiently low that transient events with a period of <1 min could be detected. Such an instrument would be advantageous for studies examining dynamic change in EWL. The ventilated-capsule system developed in our laboratory addresses these requirements. The purpose of this communication is to
describe the system and to present bench and clinical performance data.

METHODS

Figure 1 is a schematic depicting the key system components: 1) air preconditioner (chiller bath) to establish a consistent, known input condition; 2) flowmeter upstream of the vapor pickup site as one of two components necessary for leak detection; 3) pickup capsule over the measurement site; 4) dew-point temperature ($T_{dp}$) sensor; 5) flowmeter downstream of the sensor as the second component of leak detection; 6) controlled suction; and 7) hydrophobic tubing between the preconditioning and $T_{dp}$ sensor sites. Dew-point, environmental, and infant temperatures, along with airflow rate values, are captured and stored by data-acquisition/computer.

An ice-slurry, chiller bath apparatus (Fig. 2) preconditions the air to 0°C, providing a stable reference $T_{dp}$ in the incoming air. Copper tubing (~4.25 m × 1/4-in. ID) was rolled into a coil (10 cm high × 14 cm in diameter), with the low end bent to a

Inlet-stabilizing bath. The need for an inlet-stabilizing bath was clearly demonstrated by a series of trials in which significant scatter was observed due to variations in inlet dew point (as much as 1°C) during a measurement period. Chemical dryers (anhydrous calcium sulfate chambers) were tried, but these dried the air too much. Dual $T_{dp}$ sensors were tested, but the passive chiller bath was more easily maintained.

Fig. 1. Flow schematic of key system components. From left to right (as air flows through system) are input of room air to chiller bath; upstream flowmeter; pickup capsule; measurement chamber with dew-point temperature ($T_{dp}$) sensor; downstream flowmeter; and controlled constant flow suction to pull conditioned air through the system. Hydrophobic tubing (bold lines between components) is used between preconditioning and $T_{dp}$ sensor sites. Dew-point, environmental, and infant temperatures, along with airflow rate values, are captured and stored by data-acquisition/computer.

Fig. 2. Diagram of chiller bath. Room air is pulled into a copper coil at top of its spiral (a) and passes through a coil chilled by an ice slurry (b) and down to a petri dish (c), which has a pool of water at its bottom. After passing over the pool, conditioned air is pulled up through Teflon tubing (d) and to an upstream flowmeter.
gentle right angle to pass through one hole of a two-hole stopper. The stopper covers a petri dish (15 × 55 mm), which has a layer of water at the bottom. The second hole accepts a piece of thick-wall Teflon tubing (1/4 in. diameter × 55 mm length), which mates to corrugated Teflon tubing. This apparatus is placed into an ice-water slurry [some ice is held in position below the dish by an aluminum-wire mesh (1.5 × 1.5 mm lattice)]. Air is drawn through the chilled coil, over the water in the dish and exits to a flowmeter. This arrangement precipitates moisture when the ambient Tdp is >0°C and adds moisture if ambient Tdp is <0°C.

Interconnecting tubing. Teflon tubing was used between the stabilizing bath and the dew-point sensor to avoid absorption or desorption of water in the walls of the tubes. Repeatability tests and hysteresis tests conducted to investigate the system's performance under clinical conditions showed that the Tygon tubing, originally used, was a source of error. Absorption and desorption greatly increased stabilizing time (due to the time required to "load" or "unload" the tubing), and results were sensitive to changes in room temperature or thermal irradiation of the tubing. As long as the room air temperature is above the measured Tdp of the sample, neither insulation nor heating of the tubing is necessary.

Upstream flow measurement. It is essential to eliminate leaks in advance in the system before the sample reaches the Tdp-measuring chamber. The system operates subatmospherically, so leakage would bring room air into the system. Leakage can be detected by noting a discrepancy between the upstream and the downstream flowmeters. A digital flowmeter measures flow upstream from the capsule, and the resultant value is sampled by a data-acquisition system at 0.1 Hz and stored to hard disk. Core body (rectal), skin, and other temperatures are also sampled at 0.1 Hz and stored. The resultant data arrays are downloaded to a desktop computer for statistical and graphic analysis. The flowmeter has a special calibration showing accuracy of ±0.4% in the flow range usually used (0–100 ml/min).

Pickup capsule. The flowmeter is connected to a pickup capsule (Fig. 3) with convoluted and corrugated Teflon fluorinated ethylene propylene (FEP) tubing with connectors of thick-wall Teflon tubing. Plastic luer fittings on the tubing mate to the manifolds of the pickup capsule or to a 30-cm Teflon coil, which is used periodically to check baseline chiller bath conditioning. The pickup capsule was milled from clear acrylic, is held together with screws, and is affixed to the skin by double-sided tape and medical-grade adhesive dressing. No force or pressure was therefore necessary. To prevent leaks, care needed to be taken not to distort the skin of the subject when affixing and maintaining the capsule. Air enters the capsule through four holes (0.1-cm diameter) on one side of the collecting chamber, passes over the skin site, then exits through four holes (0.1-cm diameter) in the facing side. This arrangement, designed for turbulence, produces a well-mixed sample. The pickup capsule weighs 9.3 g; has external dimensions of 3.5 × 2.8 × 1.2 cm; covers an exposed skin area 1.97 × 0.97 cm², yielding a collection area of 1.91 cm²; and has a chamber volume of 1.55 cm³. There is sufficient wall thickness (0.4 cm) on the collecting chamber to allow it to be comfortably attached to the subject. With a flow of 100 ml/min, the average air velocity across the skin inside the chamber is 0.04 m/s.

Tdp sensor. The Tdp sensor (General Eastern, model D-2) is 11.4 × 11.4 × 10.2 cm in external dimensions. It contains only the Tdp-measuring chamber and is generally placed at the foot of the patient's bed, with sufficient space to allow cooling of the instrument. The flow from the pickup capsule is brought to the measuring chamber through Teflon tubing and flows over a thermoelectrically chilled mirror. An optical system detects the formation of dew on the mirror, and a Tdp monitor (General Eastern, model M-1) regulates the thermoelectric cooler to maintain incipient condensation. The average temperature for incipient condensation is the Tdp, which, along with the barometric pressure, specifies the moisture content of the mixture. Data from the Tdp monitor are captured at 1.0 Hz and stored. Final analysis on the desktop computer yields EWL capsule (EWLc) evaporation rate (mg·cm⁻²·min⁻¹) determinations for each second of the recording. Based on the data provided by the manufacturer, the temperature resolution and repeatability of the sensor is ±0.05°C and the absolute accuracy is ±0.25°C. An on-line system-accuracy check is obtained by measuring the Tdp of the air from the stabilizing chiller bath at the beginning and end of each run. With a loop of Teflon in place of the pickup capsule, Tdp should equal 0.0°C. In operation, as a routine calibration check of the system, the first Tdp observed is that of that of the conditioned air from the chiller bath.

Downstream flow measurement and flow control. A second flowmeter is downstream from the Tdp sensor. The associated data-acquisition and control software calculates the mean flow difference between the two flowmeters in a moving 50-s window. If the two readings differ by >3 ml/min, an indication of possible leak, a notification is presented to the operator. Final determinations of data quality are made from the stored data.

The flow rate was chosen after two criteria were considered. First, the flow must be low enough that the change in Tdp due to the EWLc can be measured within ±2%. Second, the pickup capsule Tdp should be within the normal range of nursery ambient conditions as well as a mass transfer coefficient inside the capsule, which is representative of a subject exposed to ambient air. Both of these criteria are affected by flow rate. By using a sample collecting area of ~2 cm², a standard flow rate of 95–100 ml/min was selected.

Approximately 1.4 m of Silox tubing connect the downstream flowmeter to the control valve of a variable-area flowmeter, and then by 0.4 m of tubing to a 1-liter plenum
with a bleed valve (to ensure smooth flow), and finally to a suction pump.

In vivo recordings. During the development and implementation of the system, recordings were made in full-term infants. The pickup capsule was sited on the infant's chest or back, a few centimeters lateral of the sternum or spine. These recordings included sweat measurements; environmental, skin, and rectal temperature measurements; and polygraph recordings. Infants were monitored continuously during afternoon recording sessions in a study examining the effects of warming on temperature regulation.

The manufacturers of components were as follows. Corrugated and convoluted tubing: Teflon FEP convoluted, Cole-Parmer; upstream flowmeter: TOP-TRAK, model 821S1-L-1, Sierra Instruments, Monterey, CA; skin adhesive dressing: OpSite, Smith and Nephew Medical, Hull, UK; dew-point sensor and monitor: models D2 and M1, General Eastern Instruments, Woburn, MA; downstream flow: Fleisch pneumotachograph model 0000, OEM Medical, Richmond, VA, with models DP-103–12–871 and CD15, Validyne Engineering, Northridge, CA; flow control: model RMA 12 TMV, Dwyer Instruments, Michigan City, IN; data-acquisition system: Keithley DAS, series 500, Boston, MA; data-acquisition computer: Apple IIe, Apple Computer, Cupertino, CA; data-acquisition hard disk: Sider II, 20 MB, First Class Peripherals, Carson City, NV; desktop computer: Quadra 950; Apple Computer, Cupertino, CA; analysis software: StatView 4.02, Abacus Concepts, Berkeley, CA; electronic analytical balance, model 1712MP8, Sartorius Instruments, McGaw Park, IL.

RESULTS

System characteristics were quantified by a number of tests and analyses that investigated the steady-state calibration accuracy and response to an abrupt change in input.

Calibration accuracy. A gravimetric calibration procedure was used to determine the accuracy of the system. A device with a constant air-water interface ("wet block") was constructed to assess system performance. The wet block design is shown schematically in Fig. 4. The pickup capsule is held in place on an aluminum plate stage over a water reservoir in a clear acrylic block. Stage is secured to the block, with a rubber gasket intervening, by screws. A wick of cotton is loosely fitted into a hole in the stage and has its other end immersed in water reservoir.

![Fig. 4. Cutaway diagram of "wet block." Pickup capsule is held in place on an aluminum plate stage (a) over a water reservoir in a clear acrylic block. Stage is secured to the block, with a rubber gasket intervening (b), by screws. A wick of cotton is loosely fitted into a hole in (c).](http://jap.physiology.org/)

The wet block was weighed on an electronic analytic balance just before and just after test periods. Typical weight loss was ~0.065 g. The wet block was at room temperature (~20°C), yielding EWLc rates of ~0.15 mgH₂O·cm⁻²·min⁻¹, which is within the expected range of rates for measurements in patients (0.017–1 mgH₂O·cm⁻²·min⁻¹). Data were recorded over test periods of ~210 min, downloaded to a desktop computer, and integration of the second-to-second EWLc rates yielded "system EWLc," i.e., total calculated EWLc system water loss. These values were compared with the "weighed EWL"; i.e., the weight loss of the wet block during the measurement period.

Representative results. The wet-block test data in the following Table 1 are those achieved with the final design of the EWL measurement system. Note the repeatability over long intertest intervals.

Based on these data sets, the measurement uncertainty of the system is within the theoretical estimation (see APPENDIX).

Response characteristics. The transient response of the system depends on the void volume within the system and the thermal time constant of the sensor. The void volume depends on the volumes of the pickup capsule and Tdp measuring chamber and on the length and diameter of the connecting tubing. With 1 m of

<table>
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<th>%Weighed</th>
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<td>27 Jan ‘94</td>
<td>0.0645</td>
<td>0.0643</td>
<td>99.69</td>
</tr>
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</table>

Accuracy, expressed as %weight change (n = 5, mean ± SD) was 99.09 ± 0.70%. EWL, evaporative water loss; EWLc, capsule EWL.
tubing, the system's flow time constant (volume divided by flow rate) is \(\sim 20\) s. The humidity sensor's characteristic time is \(\sim 15\) s. Overall, the system requires \(\sim 80\) s to complete its response to a step change (Fig. 5A). Figure 5B is a y-axis magnification of the baseline test condition. Figure 5, C-E, shows the results of bench testing to elucidate the response characteristics of the system. Multiple sequential inflation and deflation cycles of a water droplet under the pickup capsule were done at different cycle lengths (20-, 40-, and 60-s durations). The detection of periodicities appears to be valid for cycles on the order of tens of seconds.

In vivo examples. Figures 6 through 8 show examples from a recording from a 13-wk-old healthy male full-term infant in a study evaluating the relationship between sleep state and EWLc. Figure 6 is from a representative 3.5-h recording showing the 0°C Tdp condition at the start and the end of the session. Figures 7 and 8 show additional time-related clinical examples of EWLc rates measured with this system. Figure 7 shows a rapid increase in EWLc after the subject was exposed to 20 min of warming. Figure 8 shows cyclical patterns in the same subject. Further analysis and interpretation of these clinical phenomena will follow in a future publication. We believe the variations in EWLc are real and that the system allows the examination of dynamic changes in EWLc rate.

DISCUSSION

The rate of TEWL from skin depends on four primary factors: 1) the subcutaneous skin temperature; 2) the permeability of the skin to diffusion of water vapor; 3) the mass transfer coefficient between the skin and the ambient air; and 4) the concentration of water vapor in the ambient air. Any measurement system that affects these parameters will likely alter TEWL and, hence, alter EWL. Ideally, the measured EWL should repre-
sent the undisturbed value of EWL. The subcutaneous temperature and the skin permeability are patient and situation specific and may, in fact, be the objective of a measurement series. The mass transfer coefficient depends on the mean velocity and turbulence intensity in the airflow around the skin. The ambient humidity is part of the patient-care environment. The clinical examples provided in this report are estimates of EWLc from chest and back sites; however, estimates for the total body could be obtained by sampling from multiple sites (11). Estimates of the boundary layer characteristics in infant and adult patients suggest that the TEWL of healthy full-term infants and of adults with healthy skin is only slightly affected by air velocity. The overall resistance to water loss is the sum of two resistances: that of the stratum corneum and that of the boundary layer above the skin (inversely proportional to the mass transfer coefficient). For healthy full-term infants and adults with healthy skin, substantially all of the overall resistance to TEWL lies in the stratum corneum, with the boundary layer resistance contributing only 5–10% of the total. The mass transfer coefficient varies approximately with the square root of the air velocity. A change of 50% in the air velocity would decrease the mass transfer coefficient by only 25%. Thus an increase in the air velocity by 50% would change the TEWL by <2.5%. The other component of EWL, active sweating, represents water loss by direct evaporation from liquid water on the surface of the skin. Whereas the evaporation rate of an existing droplet is directly proportional to the mass transfer coefficient, the rate at which the droplets are produced is set by the thermoregulatory mechanisms of the patient.

In this communication, we have described a system that can measure EWL with an accuracy of ±2%, based on gravimetric calibration, and has the capability of detecting periodicities on the order of tens of seconds associated with sweating. The system is not affected by environmental changes (e.g., temperature, humidity, and air currents) and has the advantage of measuring

![Fig. 6. Representative 2.75-h recording showing the 0°C condition at start and at end of session. Three traces are shown: measured Tdp, measured flow rate, and calculated EWLc. Values for Tdp and measured flow rate have been divided by 1,000 for display compatibility. On the Tdp trace, note initial and final confirmations of 0°C value (the chiller bath temperature). This measurement provides a check on system accuracy at beginning and end of each data run. Note excursions in measured Tdp and corresponding excursions in EWLc. Correlation coefficient between Tdp and EWLc rate is 0.98 and between Tdp and flow is 0.34. These variations in EWLc appear to be physiological, as is supported by response time of the instrument in in vitro conditions.](http://jap.physiology.org/10.1152/jap.00223.2016)
EWL during steady-state and dynamic changes in evaporative water loss rate with minimal disruption to the subject. The modifications described are the result of attempting to use existing techniques and finding significant limitations for the study of steady-state and dynamic changes in EWL during multihour studies of infants. In pediatrics, many clinical investigations have utilized resistance hygrometry (Evaporimeter) (20, 22, 25, 29). The Evaporimeter uses the water vapor concentration gradient within a hollow cylindrical placed perpendicular to the skin surface to determine evaporative water loss. This measurement condition changes the local evaporation rate by introducing extra resistance due to the height of the cylinder. This technique was investigated by Wheldon and Monteith (30), who concluded that the Evaporimeter would underestimate "true" water loss by 10% when indicating 20 gH$_2$O·m$^{-2}$·h$^{-1}$ and by 60% or more at rates of $\approx$80 gH$_2$O·m$^{-2}$·h$^{-1}$. Blichmann and Serup (5) quote the manufacturer's accuracy statement as "±15% or 2 gH$_2$O·m$^{-2}$·h$^{-1}$, whichever is larger" and report tests showing the "technical accuracy" (in vitro calibration under carefully controlled conditions) to be ±3.3%.

Their in vivo results showed reproducibility, same location on the same subject, between ±6.3 and ±14.5%, depending on the time between successive measurements. This in vivo experience was corroborated by the studies of Smit et al. (27), who reported 13.5% (same day) and 15.1% (consecutive days) intraindividual coefficients of variation.

A recent technique developed by Graichen et al. (10) uses a Peltier module device (Bi-Tronics Dewpoint Sensor, model BI-102). This technique held the expectation for rapid and accurate measurements, but after using it for some months we found that it fell short of our needs. In this method, the sample capsule receives ambient air from the environment adjacent to the capsule, resulting in sampling site vapor mixed with vapor of unknown humidity composition. Nilsson (20) described the problems that must be solved if a ventilated chamber system was to work: 1) precise measurement of the humidity of the inlet and exit streams; 2) thoughtful selection of the flow rate, to match the normal mass transfer coefficient; 3) susceptibility to leakage; and 4) possible physiological interferences if too much force was applied to prevent leaks. The problem of obtaining a precise measurement of the humidity of the BI-102 inlet air was addressed by instituting a 2-m piece of tubing that allowed us to pull presample vapor from a more removed site. The instru-

![Figure 7](https://via.placeholder.com/150)
ment was unusable in conditions in which the dew point is \(<2^{\circ}\text{C}\), an ambient condition that can occur in our environment. It also had no capability for detection of leaks in the system, and the cable connecting the Peltier module to the main console had substantial torque, which made maintenance of an adequate leak-free measurement condition difficult.

In our system, two flowmeters are used to detect leaks, allowing corrective action and documentation of nonoptimal measurement conditions. The air inlet to the pickup capsule is a set of small jets, which increases mixing inside the pickup capsule to simulate a "well-stirred reactor." This ensures that mass transfer from the skin can be evaluated between the surface concentration at the skin and the mixed mean concentration in the capsule. With the assumption of a well-stirred reactor, the mass transfer is related to the difference between the mixed mean concentration in the chamber and the surface concentration. The mixed mean concentration in the pickup capsule is the value that is measured by the $T_{\text{dp}}$ sensor. With the above problems addressed, significant improvements in precision and accuracy in the measurement of TEWL were possible.

Advantages of our system include 1) continuous long-term recording capability; 2) excellent accuracy and repeatability in dynamic testing; 3) excellent precision; 4) minimal susceptibility to environmental perturbation compared with other devices, e.g., Evaporimeter and Dew-Point Hygrometer; 5) calibration checks at beginning and end of data acquisition; 6) leak detection; 7) maintenance of integrity of sample quasi-bolus; and 8) standardization of measurement conditions. Some limitations of our system are as follows: 1) additional control systems would be necessary to run the system in high EWL rate conditions occurring in a cool testing situation (i.e., over $\sim1.167\text{mgH}_2\text{O}\cdot\text{cm}^2\cdot\text{min}^{-1}$ at $\sim75^{\circ}\text{F}$); 2) tubing to patient could be restrictive (or could be dislodged) in an ambulatory or vigorously active patient; 3) slight delay ($\sim80\text{s}$ for a step change) in determinations; and 4) at present, nonambient measurement conditions.

In summary, the accuracy of this system, based on gravimetric calibration, is $\pm2\%$ and stable to environmental disturbance (e.g., changes in temperature, humidity, and air currents). The system has the advantage of examining steady-state and dynamic changes in evaporative water loss rate with minimal disturbance of the subject. Therefore, it is very suitable for studies on the development of the thermoregulatory effector re-

Fig. 8. Cyclical patterns in EWL$_c$ rate, which may differ depending on sleep state. Scoring of sleep state with standard polysomnography established a transition from REM to NREM sleep at elapsed time of 58.5 min.
sponses and the effect of sleep development or sleep state on those responses.

**APPENDIX**

Equations for Calculation of EWL (m$_{s2}$)

The general form of the equation used to calculate the rate of water loss is

$$\dot{m}_{s2} = \frac{\dot{Q}}{A} (1.204.7)(SH_2 - SH_1)$$

where $\dot{m}_{s2}$ is the mass flux of water from the skin surface (in mgH$_2$O·cm$^{-2}$·min$^{-1}$); $\dot{Q}$ is volume flow rate of mixture at the upstream flow meter (in l/min); $A$ is skin area exposed inside the pickup capsule (in cm$^2$); $SH_2$ is specific humidity down-stream of the pickup capsule (in mgH$_2$O/mgair); $SH_1$ is specific humidity upstream of the pickup capsule (in mgH$_2$O/mgair); and 1204.7 is density of dry air at srp (in mgair/l).

The value of $SH_1$ is set by the equation of state of water and the design of the inlet conditioning tank. Its value is known a priori and will be considered to have negligible uncertainty.

The equations used treat the mixture density at the flow meter location as a constant, equal to that of air at standard conditions of temperature and pressure. Because flow meters are essentially in equilibrium with the room air temperature, this seems a reasonable assumption.

Uncertainty

The uncertainty in measured values of $\dot{m}_{s2}$ has been estimated following the method of Kline and McClintock (18) as expanded by Moffat (19). The following form is used, which neglects the uncertainty and level of $SH_1$ as expanded by Moffat (19). The following form is used, which

$$\frac{\delta \dot{m}_{s2}}{\dot{m}_{s2}} = \sqrt{\left(\frac{\delta \dot{Q}}{\dot{Q}}\right)^2 + \left(\frac{\delta A}{A}\right)^2 + \left(\frac{\delta SH_2}{SH_2}\right)^2}$$

We estimate $\delta \dot{Q}/\dot{Q}$ to be ±0.02 based on manufacturer's specifications. The uncertainty in area arises not so much from the geometric problem of measuring the area but from the question of deformation of the skin inside the capsule in response to the applied force holding the capsule in place and the slight suction force exerted by the negative pressure in the cavity. We estimate $\delta A/A$ to be ±0.02.

The remaining factor is $\delta SH_2/SH_2$. We estimate this from the uncertainty in the measured $T_{dp}$ temperature (±0.02°C), by using a polynomial curve fit to the relationship between $SH$ and $T_{dp}$. Based on that curve fit

$$\frac{\delta SH}{\delta T_{dp}} = 
\begin{align*}
0.0003 & \text{ at } 0°C \\
0.0005 & \text{ at } 10°C \\
0.0008 & \text{ at } 20°C
\end{align*}$$

with an uncertainty of ±0.2°C in the measured $T_{dp}$ (a conservative estimate).

Then

<table>
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<th>$T_{dp}$</th>
<th>$\delta SH$</th>
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<td>0</td>
<td>$6 \times 10^{-6}$</td>
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<tr>
<td>20</td>
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</tr>
</tbody>
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The overall uncertainty in $\dot{m}_{s2}$ is, therefore, a slight function of the operating $T_{dp}$.

These estimates reflect uncertainty in the measurement but do not assess whether the act of measurement altered the patient's behavior. The uncertainty reflects the effect on the calculated $m_{s2}$ associated with the measurements of $T_{dp}$, flow rate, and area.

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