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Abstract

A small and wireless device that can capture the temporal pattern of perspiration by a novel structure of water vapor collection combined with reusable desiccant has been developed. The novel device consists of a small cylindrical case with a temperature/relative humidity sensor, battery-driven data logger, and silica gel (desiccant). Water vapor of perspiration was detected by the change in relative humidity and then adsorbed by silica gel, allowing continuous recording of perspiration within a closed and wireless chamber, which has not been previously achieved. By comparative experiments using the commercially-available perspiration monitoring device, the developed device could measure perspiration as efficiently as the conventional one, with a normalized cross coefficient of 0.738 with 6 s delay and the interclass correlation coefficient [ICC(2, 1)] of 0.84. These results imply a good agreement between the conventional and developed devices, and thus suggest the applicability of the developed device for perspiration monitoring.

Key words: perspiration monitoring, emotional sweating, sympathetic activity

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1. Introduction

Perspiration, or sweating, is one of the most fundamental phenomena in human physiological events. The main result of systemic sweating is cooling effect for thermoregulation by sweat evaporation [1], which is called “thermal sweating”. There is another type of sweating called “emotional (or mental) sweating.” Emotional stresses (e.g., rising tension, upset, and concentration) trigger sweating, particularly on the face, palm, and sole via sympathetic nervous tone [2-4]. To date, a number of diseases have been reported to be associated with sweating abnormalities such as thyroid diseases [5], dysautonomia [6], menopause [7], and social anxiety disorder [8]. In addition, the perspiration monitoring can be utilized for prediction or diagnosis of nervous disorders such as brachial plexus avulsion (BPA) [9] and reflex sympathetic dystrophy [10]. Especially, the monitoring of sympathetic activity such as perspiration might be important for the early diagnosis of obstetric BPA [11], because it is often difficult for neonates to express their symptoms verbally. In light of the possible applications of ubiquitous perspiration monitoring such as a prediction or diagnosis of perspiration-related disorders, a small, convenient, and sensitive device for perspiration monitoring has been desired.

A skin conductance meter was used as an indirect method for estimation of sweating [12], and a simple humidity meter was employed to measure water evaporation from the skin [13,14]. Recently, wearable, adhesive, and tattoo-like sweat monitoring device have been proposed [15-17], although they are more intended for prediction of sweat electrolytes rather than perspiration amount, or they give an indirect index of perspiration. At this time the latest perspiration measurement device involves colorimetric detection by a digital camera, which requires special setup [18]. As a more direct measurement of water exchange, the vapor pressure diffusion method and ventilated chamber method were developed [19,20]. The vapor pressure method utilizes the theory that the amount of water exchange (F) in natural flow is calculated as $F = D(\partial p/\partial x)$, where D is the temperature- and atmospheric pressure-dependent diffusion coefficient, p is the water vapor pressure, and x is distance from the surface [19]. Because p can

be calculated from relative humidity and temperature, at least four sensors [(humidity + temperature) × two points] are required. In addition, this method relies on the assumption that the state of the outer atmosphere is unchanging, which is unlikely in daily perspiration monitoring. To address this, a closed chamber system with water vapor condenser was proposed [21,22], although the coolant (Peltier device) is required and thus power consumption would be measurable. The theory of the ventilated chamber method is similar to that of the vapor pressure method, except this method uses forced and constant airflow. The constant airflow is injected into a small chamber adjacent to the skin, and the air with evaporated water vapor is collected in an outlet chamber. The amount of water exchange is then calculated with the airflow rate and difference of humidity between inlet and outlet air [19,20,23]. However, it is difficult to contain air ventilator and chambers in one small package such that it can be of practical use in daily life. Therefore, it was considered beneficial to develop a small device that could monitor perspiration and allow prediction of emotional and physiological status.

The aim of this study was to develop a small device for perspiration monitoring and compare its performance with a conventional sweat meter and stress analyzing method under conditions of mental stress.

2. Materials and Methods

2.1. Developed device

Fig. 1 shows the exterior (Fig. 1A, B) and structure (Fig. 1C) of the developed device. A custom-made data logger circuit with battery, dry silica-gel (Wako Pure Chemical Industries, Ltd., Osaka, Japan), and a small temperature/relative humidity (T/RH) sensor (SHT-21, Sensirion AG, Zürich, Switzerland; accuracy of temperature is ±0.3°C, accuracy of relative humidity is ±2%, calibrated at Industrial Research Institute of Ishikawa, Japan) with a sampling rate of 1 Hz was encapsulated in this order in a small plastic chamber toward measuring windows facing the skin (Fig. 1C).

90

91 *2.2. Perspiration rate calculation*

92 The theory of this equipment is based on the vapor pressure method [19,21] with modifications.
93 Fig. 2 illustrates the method of perspiration measurement in the developed equipment.
94 According to the Fick's law of diffusion, the flux of water vapor J (g m^{-2} s^{-1}) between two
95 points can be calculated as Eq. (1):

$$J = -D\frac{\mathrm{d}H}{\mathrm{d}x} \qquad (1)$$

96 where D (m^2 s^{-1}) is a diffusion coefficient of water vapor in the air, dH (g m^{-3}) is a difference of
97 concentration of water vapor, and dx (m) is a distance between two points. In the developed
98 equipment, two different fluxes of water vapor: from the skin surface to the T/RH sensor (Fig.
99 2A, green arrow; **s–w**), and from the sensor to the dry chamber (Fig. 2A, blue arrow; **w–d**), can
100 be theorized. The flux difference between (**s–w**) and (**w–d**) could be detected as a change of
101 humidity in T/RH sensor. The water exchange between the skin and silica gel via T/RH sensor
102 should satisfy Eq. (2):

$$\begin{aligned} V\frac{\Delta H_x(t)}{\Delta t} &= A_1J_1 - A_2J_2 \\ &= A_1D\frac{H_1(t) - H_x(t)}{L_1} - A_2D\frac{H_x(t) - H_2}{L_2} \end{aligned} \qquad (2)$$

103 where V (m^3) is a volume of wet chamber in which the T/RH sensor exists; $H_1(t)$, $H_x(t)$, and H_2
104 (g m^{-3}) are the concentrations (i.e., absolute humidity) of water vapor at the skin surface, T/RH
105 sensor, and dry chamber, respectively; A_1 and A_2 (m^2) are the areas of windows at (**s–w**) and
106 (**w–d**) junctions, respectively; J_1 and J_2 are the fluxes of (**s–w**) and (**w–d**), respectively; L_1 and
107 L_2 (m) are the distances of (**s–w**) and (**w–d**), respectively (Fig. 2A). H_2 is assumed to be
108 constant due to a buffering effect of desiccant (preliminary experiment is shown in Fig. S1).
109 Because J_1 in Eq. (2) simply represents the flux of total water vapor from the skin surface [i.e.,
110 perspiration and constant transepidermal water loss (TEWL)], the Eq. (2) can be solved for J_1 as
111 following Eq. (3):

$$W(t) = J_1 = \frac{V}{A_1}\frac{\Delta H_x(t)}{\Delta t} + \frac{A_2}{A_1}\frac{D(t)}{L_2}(H_x(t) - H_2) \qquad (3)$$

where the rate of water vapor diffusion from the skin $W(t)$ (g m^{-2} s^{-1}) can be calculated only by measuring $H_x(t)$ with T/RH sensor, as V, A_1, A_2, L_2, and H_2 are all considered fixed, and $D(t)$ can be calculated by using following Eq. (4) under normal atmospheric pressure [24]:

$$D(t) = 1.87 \times 10^{-10} \times T(t)^{2.072} \qquad (4)$$

where $T(t)$ (K) is the temperature at the time t.

Here, the fixed values were set to: $V = 6.3 \times 10^{-7}$ m^3, $A_1 = 1.3 \times 10^{-5}$ m^2, $A_2 = 5.6 \times 10^{-5}$ m^2, $L_2 = 5.0 \times 10^{-3}$ m according to the equipment design, and H_2 was estimated to be 1.7 g m^{-3} (Fig. S1). Finally, the perspiration $Per(t)$ (mg cm^{-2} min^{-1}) [20,23] can be calculated from $W(t)$ by a simple conversion Eq. (5):

$$Per(t) = 6W(t) \qquad (5)$$

because 1 g m^{-2} s^{-1} is equal to 6 mg cm^{-2} min^{-1}.

The conversion from relative humidity h (%) to absolute humidity H (g m^{-3}) was as following Eq. (6) based on the ideal gas law:

$$H = \frac{M_w P_s(T)}{RT}\frac{h}{100} \qquad (6)$$

where M_w is the molecular weight of water (= 18.02 g mol^{-1}), $P_s(t)$ is the saturated water vapor pressure (kPa) at the temperature T (K), and R is the gas constant ($R = 8.314 \times 10^{-3}$ kPa m^3 K^{-1} mol^{-1}), according to the American Society of Heating, Refrigerating and Air-Conditioning Engineers guidelines [25].

Because the change in water vapor flux includes constant water loss [26] and perspiration, a baseline subtraction has been employed. As shown in Fig. 2B, it is theorized that the baseline (i.e., lower envelope) and wave crests reflect the constant water loss and perspiration, respectively. To extract the perspiration pattern after the recording of water vapor flux, the difference between water vapor flux and the baseline was calculated by the embedded program in Origin software (version 2015; OriginLab Corp., MA, USA).

2.3. Verification of developed equipment

The performance test, in which the direct water vapor was applied to the developed device, was first performed (for details see Fig. S2). For verification of the developed perspiration monitor and calculation method described above, the perspiration pattern obtained from the developed device and commercially available conventional sweat meter was compared as follows. First, five individuals for the test were employed after obtaining written informed consent. The two devices, both newly developed and conventional devices (SKD-1000, Skinos Co., Ltd., Nagano, Japan; nominal uncertainty is ±10% of measured value), were attached side-by-side to the palm of each individual, followed by simultaneous perspiration recording in a sitting position for 30 min. After recording, the temporal changes of perspiration were compared using a normalized cross-correlation function (nCCF). The corresponding peak-to-peak values of perspiration patterns were analyzed by general Deming regression [27], absolute interclass correlation coefficient of two variables [ICC(2, 1)] [28,29], and Bland–Altman plot [30,31] to estimate the agreement of both devices.

2.4. Measurement of perspiration evoked by sympathetic activity

To determine if the developed equipment could detect perspiration under stress conditions, an experiment was conducted utilizing sympathetic activity. First, the developed device and skin potential sensor (NE-114A; Nihon Kohden Corp., Tokyo, Japan) were attached to their palm (Fig. 4A). Furthermore, they were requested to perform the following tasks: (1) take a deep inspiration 5 times at intervals of 1 min and (2) do a mental calculation (e.g., the subjects were orally requested to continuously subtract 7 from 100) for 5 min to evoke sympathetic activity that involves perspiration on the palm [32,33]. During the test, the perspiration and sympathetic skin response (SSR) on the palm were recorded [33]. These data were used to confirm if the developed device could detect perspiration by mental stress.

2.5. Ethical approval

These protocols involving human subjects were approved by the Medical Ethics Committee of Kanazawa University (#553).

3. Results

3.1. Perspiration recording by the developed and conventional devices

As a result of performance test, the developed device was confirmed to be able to measure the water exchange with an uncertainty of <±5% and a long-term stability of >4 h (Fig. S2). The representative temporal changes in perspiration recorded by the developed and conventional devices are shown in Fig. 3A, B (the baseline data were shown in Fig. S3). The comparable time profiles of perspiration in a steady state were observed (Fig. 3A, B). Perspiration pattern determined by the developed device was in good agreement with that of the conventional device [Fig. 3C; peak correlation coefficient of 0.738 at −6 s, and Fig. 3D; ICC(2, 1) of the corresponding peak-to-peak amplitudes was 0.84 with the 95% confidence interval (CI) of 0.76–0.90]. The Bland–Altman plot revealed a fair agreement between both devices [Fig. 3E; bias = −0.0042 mg cm^{-2} min^{-1} (95% CI: −0.014–0.0056) with the limits of agreement −0.087 to 0.079]. These results imply that the developed device could capture perspiration as efficiently as the conventional one. The other subjects showed similar results (data not shown).

3.2. Detection of palmar perspiration evoked by the sympathetic activity using the developed device

Furthermore, whether the developed device could capture the onset of perspiration was tested. To test this, sympathetic activation that is related to palmar sweating [32,33] was utilized. For measurement of sympathetic activity, the palmar SSR was recorded at the same time that perspiration was measured (Fig. 4A). With these sensors, subjects were requested to perform two tasks (a deep inspiration and mental calculation) to evoke sympathetic activity. During the

stress test, the palmar SSR suggested substantial reactions according to the stressor (Fig. 4B, SSR). In the same manner, palmar perspiration recorded by the developed device showed a stress-induced pattern with good agreement with the SSR (Fig. 4B, perspiration). From these results, it is plausible that the developed device can indeed capture the onset of perspiration. The other subjects showed similar results (data not shown).

4. Discussion

The aim of this research was to develop a small device for perspiration monitoring. To achieve this, a small, stand-alone temperature and relative humidity sensor used to calculate absolute humidity was designed, allowing wireless monitoring of water exchange with a small exterior. In addition, a novel closed-chamber system with silica gel allowing constant measurement independent from the ambient condition was introduced. The developed sensor for perspiration monitoring was validated in human subjects by a comparison of the conventional and developed devices and by concurrent monitoring of sympathetic activity-related perspiration.

In this study, a modification of the vapor pressure method [19] was utilized. In the conventional vapor pressure method, water exchange on the skin (i.e., constant water loss and perspiration) can be detected as the natural flow of water vapor. However, the flow from the skin to ambient air is dependent on outer air conditions such as temperature and humidity, i.e., the previous method would be deeply affected by the nature of outer atmosphere. Thus, a combination of a closed chamber with enforced ventilation has been developed [20,23,34]. These combined methods use dehumidified nitrogen or ventilation pumps, which could hamper daily monitoring of perspiration patterns. To address these limitations, a closed-chamber filled with silica gel above the T/RH sensor was developed (Figs. 1 and 2). In this system, the adsorption of water vapor into silica gel generates a natural but constant flow of water vapor. Under such a constant flow, the T/RH sensor below the silica gel can constantly measure the water exchange without the interference of ambient air in a small and wireless exterior (Figs. 1

212 and 2).

213 In principle, the developed device is relying on the consistency of humidity in a
214 desiccant-filled chamber [H_2 in Fig. 2A and Eq. (3)]. According to the preliminary study, the
215 variability of relative humidity in a silica gel-filled chamber was small (Fig. S1; about 1–2 g
216 m^{-3}) and can be considered static when compared to the change of humidity in a wet chamber
217 facing to the skin (about 15–30 g m^{-3}). The change of H_2, therefore, could be negligible. That
218 said, a dual T/RH sensor system (one sensor is in a wet chamber, the other is in a
219 desiccant-filled one) would increase an accuracy of perspiration measurement by eliminating
220 the small fluidity of H_2, although the power consumption would be doubled and thus,
221 measurable time would be halved.

222 Water evaporation from the skin includes constant water loss [26] and perspiration.
223 Among them, perspiration was particularly focused because a number of diseases are associated
224 with perspiration abnormalities [5-8,35-41]; therefore, monitoring of perspiration would be
225 considered beneficial for health. Because the water exchange measured by the developed device
226 includes both constant water loss and perspiration, a baseline subtraction (Fig. 2B) was adopted
227 to extract perspiration. In this method, it was theorized that the baseline indicates constant water
228 evaporation, while the “crests” reflect perspiration. As a result of developed methods,
229 comparable perspiration profiles were obtained between conventional and developed devices
230 (Fig. 3). The high cross correlation (nCCR = 0.738) and interclass correlation coefficient of the
231 peak-to-peak values [ICC(2, 1) = 0.84] indicate the considerable agreement between the
232 conventional and developed equipment (Fig. 3). In addition, the developed device could detect
233 perspiration evoked by mental stress (Fig. 4) like as the conventional device [32,33], which
234 further adds to the applicability of this system for perspiration monitoring.

235 It should be noted, however, that there are some limitations with respect to the
236 developed device.

237 First, the developed device cannot detect constant water loss (e.g., TEWL) that the

238 conventional device can measure (note the baseline shift of perspiration profiles in Fig. 3A).
239 This is because of the baseline subtraction introduced in this study (Fig. 2B). In principle, the
240 information about constant water loss including TEWL was eliminated by baseline subtraction.
241 The data of baseline themselves might have information on TEWL, although further analysis
242 about the baseline data would be required.

243 Second, the perspiration amount (i.e., the peak-to-peak amplitude of perspiration
244 profiles) did not always correspond between conventional and developed devices, especially at
245 higher values (Fig. 3D, E). The slight difference could be explained by the uncertainty of each
246 variable in Eq. (3) used for perspiration calculation. Although we have determined the overall
247 variability as being <5% (Fig. S2), the uncertainty of each variable such as $H_x(t)$ and H_2, which
248 has not been estimated in this study, is also the seed of error. A more accurate measurement
249 could be achieved by incorporating such errors, although the calibration of the T/RH sensor and
250 the performance test should be enough for general perspiration monitoring. The difference could
251 also be explained by the responding speed of the T/RH sensor. The sensor used in this study
252 (SHT-21) has a time constant of 8 s, corresponding to the delay of approximately 8 s in the
253 output. It is also possible that there is a speed limit of water vapor adsorption in the silica gel. It
254 is plausible, therefore, that the large and sharp spike of perspiration might not be appropriately
255 detected by the T/RH sensor because of the sensing delay and/or adsorption speed limit. Indeed,
256 the perspiration pattern recorded by the developed device reported slightly delayed (6 s) data
257 compared to the conventional one (Fig. 3C), the waveform of the developed device was blunter
258 than that of the conventional device, and the response against the decrease of perspiration is
259 slower than that of the perspiration onset (Fig. 3A, B). Third, the saturation of desiccant would
260 be a problem. The silica gel used in the device (about 4 g) can capture up to about 1.2 g of water
261 vapor (i.e., about 30% of its own weight) at body temperature [42]. This capacity could be
262 sufficient to capture water vapor for >900 minutes if the constant 10 mg cm^{-2} min^{-1} water vapor
263 evaporation is simulated, which is beyond the observed perspiration plus water loss (Figs. 3A

and S3). In addition, the performance test has proved that at least 4 h continuous water vapor adsorption did not affect the readout value of the device (Fig. S2). Therefore, the saturation of desiccant in the developed device would be considered negligible.

Despite these drawbacks, the developed device could measure perspiration profiles in an easy and convenient way, which may be suitable for daily monitoring of perspiration. The next goal should be to confirm the more precise estimation of perspiration amount for diagnostic purpose, to analyze the baseline data which may contain the information about TEWL, and to explore the applicability of the device at various positions on the body; nonetheless the detection of perspiration at anterior chest has been already confirmed (data not shown).

In conclusion, a small and wireless device was developed to capture the temporal pattern of perspiration using a novel method of water vapor collection combined with a reusable desiccant. With further refinement, this system could be applicable for daily perspiration monitoring, and could predict the onset of the diseases related to perspiration abnormalities.

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Ethical Approval

This study was approved by the Medical Ethics Committee of Kanazawa University (#553-1).

Conflict of interest

FM is the president of the Rousette Strategy Inc. where the developed device was assembled.

No financial support was received from either FM or the Rousette Strategy Inc.

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393

394

395 **Figure legends**

396 **Figure 1** Outline of the developed device. (A, B) Exterior of the device. A small plastic cylinder
397 (A) contains a temperature/relative humidity (T/RH) sensor, electric boards, and silica gel (B).
398 (C) Schematic of the device.

399

400 **Figure 2** Principles of perspiration monitoring. (A) In this system, two different water vapor
401 fluxes were theorized: from the skin surface to the wet chamber (green arrow; **s–w**), and from
402 the wet chamber to the dry chamber (blue arrow; **w–d**). The perspiration with constant water
403 loss can be obtained by the calculation of water vapor flux from the skin surface (green arrow).
404 (B) After obtaining the temporal data of water vapor flux, the baseline subtraction was
405 introduced to separate perspiration and constant water loss.

406

407 **Figure 3** Comparison of the temporal patterns of perspiration. (A) Perspiration patterns
408 obtained by the conventional (upper) and developed (lower) devices are shown. (B) These
409 devices showed similar patterns of perspiration. (C) The normalized cross correlation function
410 (nCCF) of these devices. Note a good correlation (0.738) between the conventional and
411 developed devices with a short delay (−6 s). (D, E) Agreement of peak-to-peak values by the
412 two devices. (D) Scatter plot of conventional (*x*-axis) and developed (*y*-axis) devices with linear
413 Deming regression (red line) and 95% prediction interval (PI). These devices showed a good
414 agreement, with a high absolute interclass correlation coefficient [ICC(2, 1) = 0.84]. A blue line
415 indicates complete agreement. (E) Bland–Altman plot with the limits of agreement (bias =
416 −0.0042 mg cm^{-2} min^{-1} with the limits of agreement −0.087 to 0.079).

417

418 **Figure 4** Detection of sympathetic palmar perspiration. (A) Experimental condition: the subject
419 was requested to attach the developed device and sensors of sympathetic skin response (SSR)

side-by-side followed by maintaining a rest position. (B) Representative data of the SSR and palmar perspiration. SSR activity was observed by deep inspiration and mental calculation. The developed device could successfully record palmar perspiration in response to the SSR.

Figure S1 The humidity change in wet and dry (desiccant) chambers. Although these data were separately obtained in the same condition, a very small change of absolute humidity in a dry chamber (B) compared to the wet one (A) should be noted.

Figure S2 The performance test of the developed device. In the test, the developed device was placed over the cylindrical water tank with a polytetrafluoroethylene (PTFE) sealant (A) in the temperature-controlled room. The constant and massive water vapor was created by heating the tank at 35°C, and the recording of water vapor flux was performed. As a result, about 4 h continuous recordings could be achieved (B). The uncertainty of the calculated water vapor rate was less than ±5% for ~4 h (average = 14.89 g m^{-2} min^{-1}, min–max = 14.54–15.55; corresponding to −2.3–+4.4% error against the average). After the experiment, the weight change of the silica gel (i.e., the amount of water transfer) was 108 mg, whereas the integral of calculated water vapor flux was 103.47 mg for 4 h; the error was −4.2% of the actual value (B).

Figure S3 The baseline data used in Fig. 3A.

Figure1

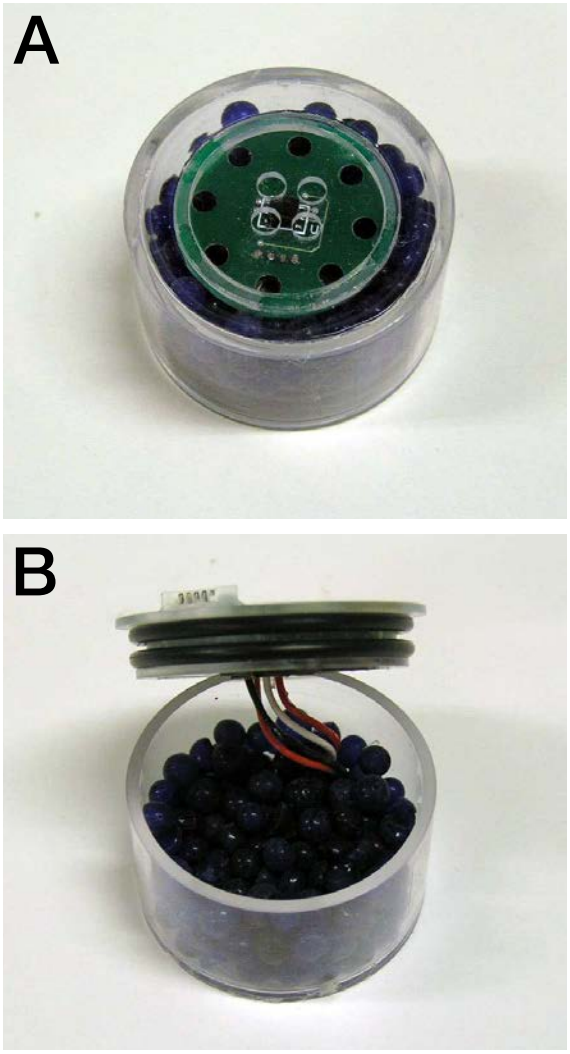
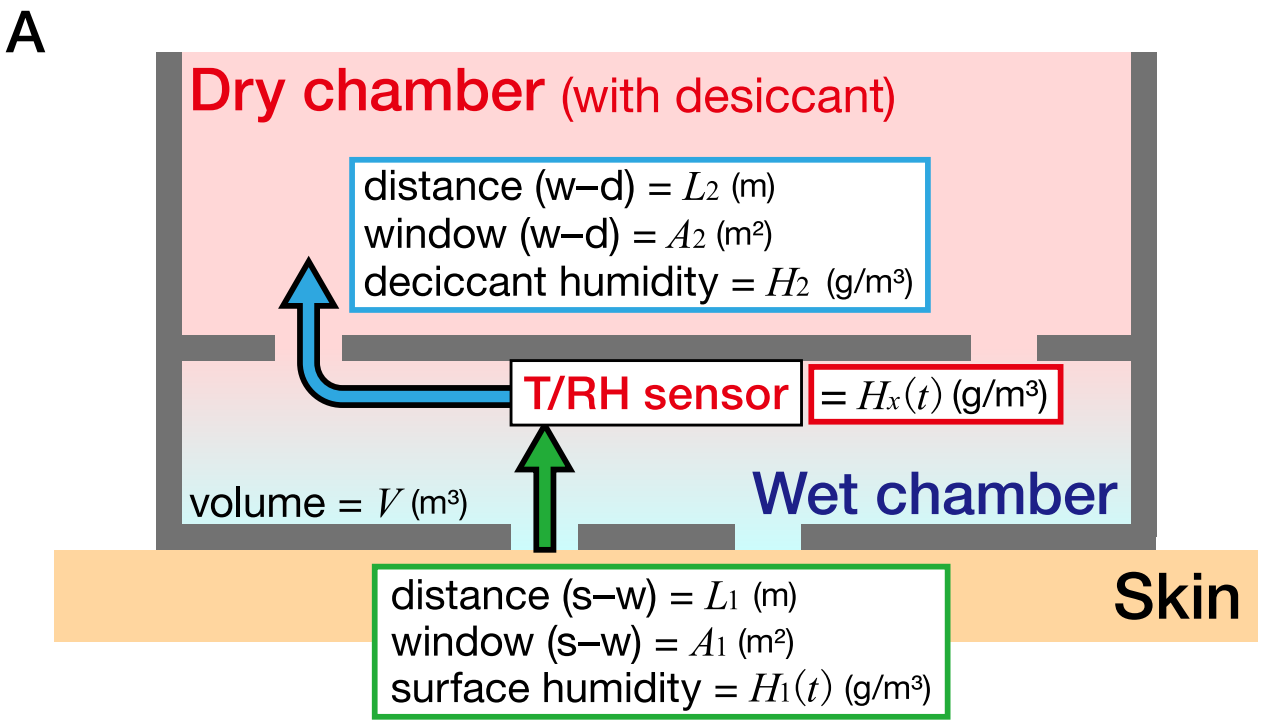

A
B
C
28 mm
Holes
Sensor board
T/RH sensor
Data logger
20 mm
Silica gel
T/RH sensor
Water vapor from
the skin surface

Figure2


A
Dry chamber (with desiccant)
distance (w–d) = L_2 (m)
window (w–d) = A_2 (m²)
deciccant humidity = H_2 (g/m³)
T/RH sensor
= $H_x(t)$ (g/m³)
volume = V (m³)
Wet chamber
Skin
distance (s–w) = L_1 (m)
window (s–w) = A_1 (m²)
surface humidity = $H_1(t)$ (g/m³)

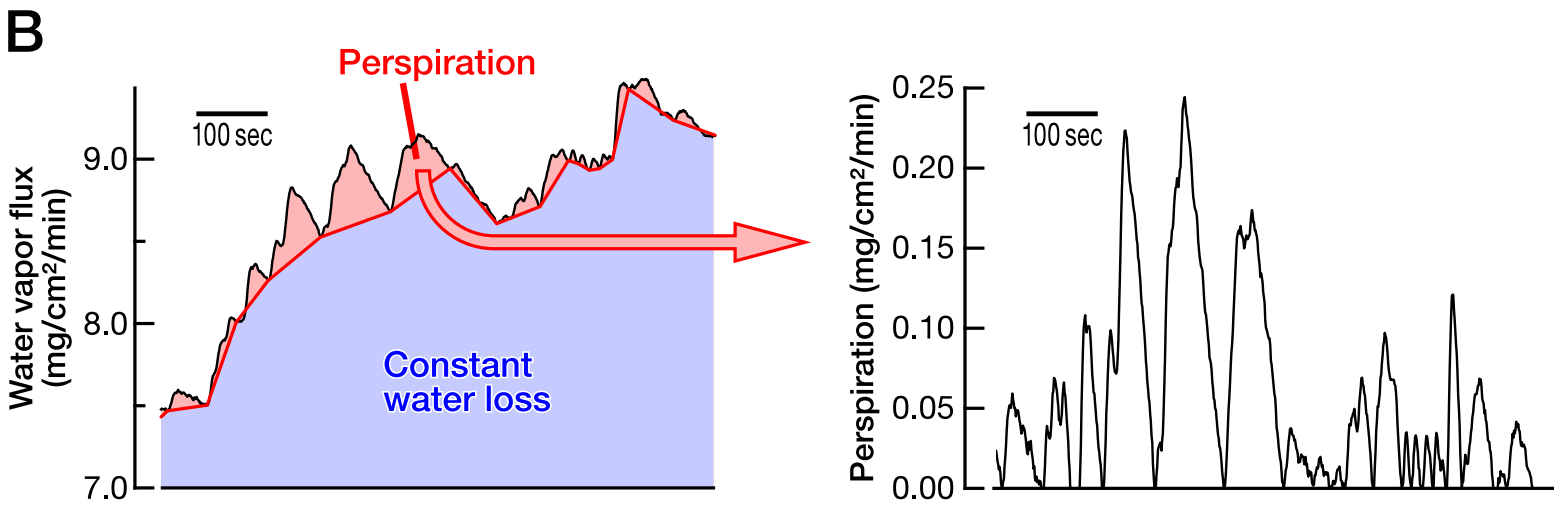

B
Perspiration
100 sec
Water vapor flux (mg/cm²/min)
9.0
8.0
7.0
Constant water loss
Perspiration (mg/cm²/min)
0.25
0.20
0.15
0.10
0.05
0.00
100 sec

Figure3

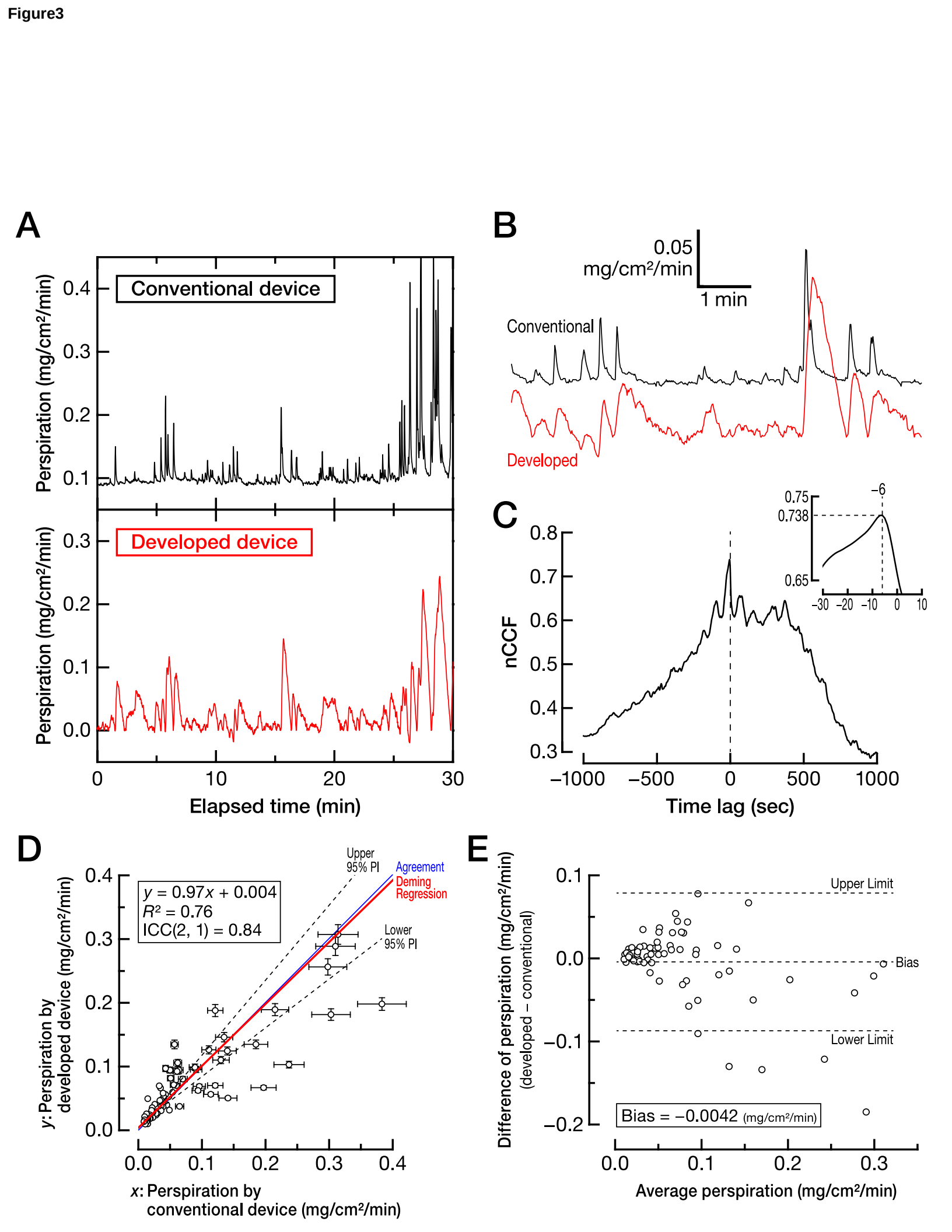
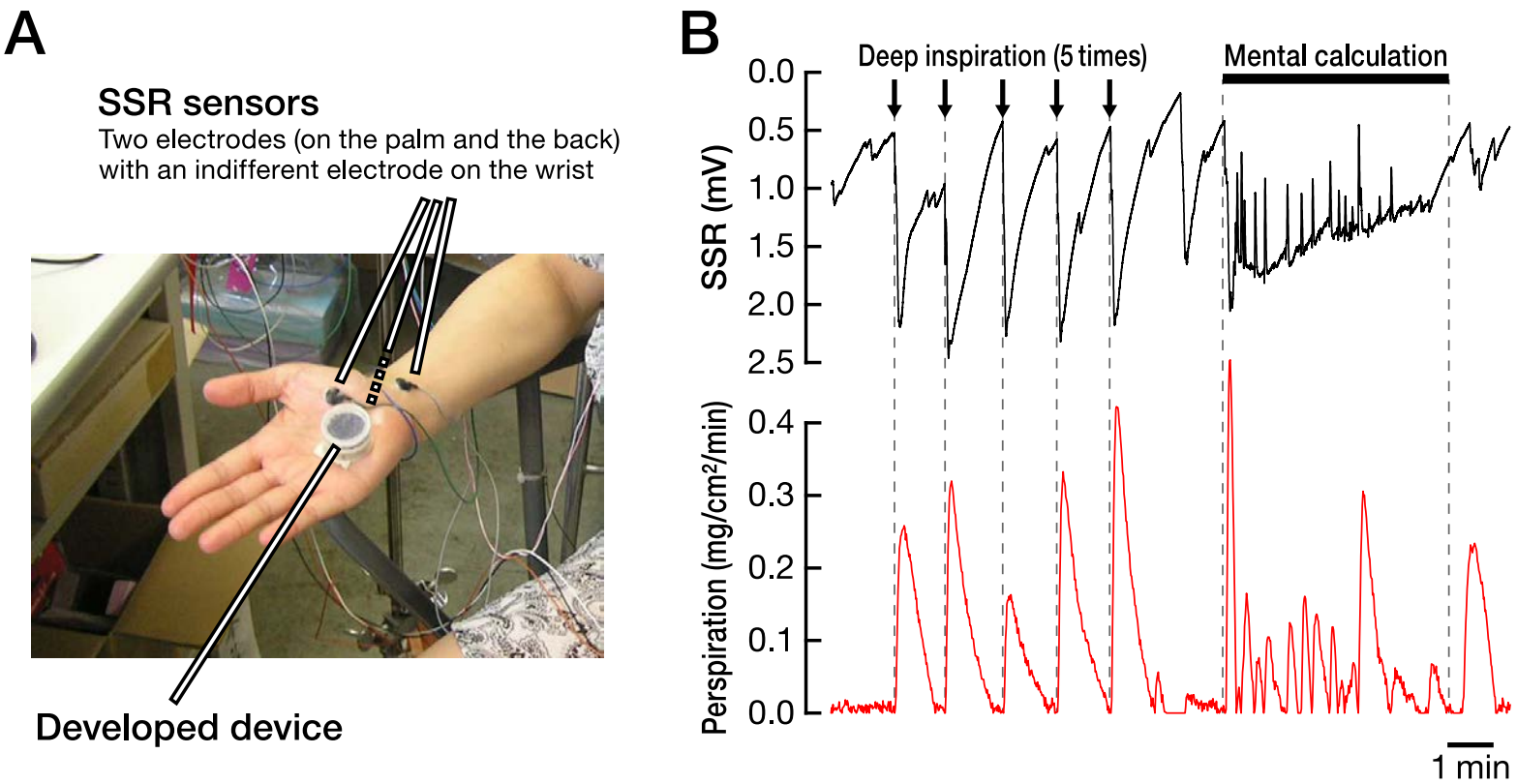

A
Conventional device
Developed device
Perspiration (mg/cm²/min)
Elapsed time (min)
B
0.05
mg/cm²/min
1 min
Conventional
Developed
C
nCCF
Time lag (sec)
D
y = 0.97x + 0.004
R² = 0.76
ICC(2, 1) = 0.84
Upper 95% PI
Agreement
Deming Regression
Lower 95% PI
y: Perspiration by developed device (mg/cm²/min)
x: Perspiration by conventional device (mg/cm²/min)
E
Upper Limit
Bias
Lower Limit
Bias = −0.0042 (mg/cm²/min)
Difference of perspiration (mg/cm²/min) (developed − conventional)
Average perspiration (mg/cm²/min)

Figure4


A
SSR sensors
Two electrodes (on the palm and the back)
with an indifferent electrode on the wrist
Developed device
B
Deep inspiration (5 times)
Mental calculation
SSR (mV)
0.0
0.5
1.0
1.5
2.0
2.5
Perspiration (mg/cm²/min)
0.4
0.3
0.2
0.1
0.0
1 min