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著者	Ogai Kazuhiro, Fukuoka Masakazu, Kitamura Kei-ichiro, Uchide Kiyoshi, Nemoto Tetsu
journal or publication title	Medical Engineering and Physics
volume	38
number	4
page range	391-397
year	2016-04-01
URL	http://hdl.handle.net/2297/44851

doi: 10.1016/j.medengphy.2015.12.009

1 **Development of a small wireless device for perspiration monitoring**

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3 Kazuhiro Ogai^{a,*}, Masakazu Fukuoka^b, Kei-ichiro Kitamura^c, Kiyoshi Uchide^d, Tetsu Nemoto^e

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5 ^a Wellness Promotion Science Center, Institute of Medical, Pharmaceutical and Health Sciences,
6 Kanazawa University, 5-11-80 Kodatsuno, Kanazawa, Ishikawa 920-0942, Japan

7 ^b Advanced Research Center for Human Sciences, Waseda University, 2-579-15 Mikajima,
8 Tokorozawa, Saitama 359-1192, Japan

9 ^c Department of Clinical Laboratory Science, Graduate School of Medical Science, Kanazawa
10 University, Kanazawa, Ishikawa 920-0942, Japan

11 ^d Asanogawa General Hospital, Naka 83, Kosaka-machi, Kanazawa, Ishikawa 920-8621, Japan

12

13 *All correspondence should be addressed to Kazuhiro Ogai

14 Wellness Promotion Science Center, Institute of Medical, Pharmaceutical and Health Sciences,
15 Kanazawa University, 5-11-80 Kodatsuno, Kanazawa, Ishikawa 920-0942, Japan

16 Tel./fax.: +81-76-265-2590

17 E-mail: kazuhiro@staff.kanazawa-u.ac.jp

18 **Abstract**

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

31

32 **Key words:** perspiration monitoring, emotional sweating, sympathetic activity

33

34 **Word count in the body text** (“Introduction” to “Discussion”): 3,092

35 **The number of figures:** 4

36 **The number of supplemental figures:** 3

37 **The number of tables:** 0

38 **1. Introduction**

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

53 A skin conductance meter was used as an indirect method for estimation of sweating
54 [12], and a simple humidity meter was employed to measure water evaporation from the skin
55 [13,14]. Recently, wearable, adhesive, and tattoo-like sweat monitoring device have been
56 proposed [15-17], although they are more intended for prediction of sweat electrolytes rather
57 than perspiration amount, or they give an indirect index of perspiration. At this time the latest
58 perspiration measurement device involves colorimetric detection by a digital camera, which
59 requires special setup [18]. As a more direct measurement of water exchange, the vapor pressure
60 diffusion method and ventilated chamber method were developed [19,20]. The vapor pressure
61 method utilizes the theory that the amount of water exchange (F) in natural flow is calculated as
62 $F = D(\partial p/\partial x)$, where D is the temperature- and atmospheric pressure-dependent diffusion
63 coefficient, p is the water vapor pressure, and x is distance from the surface [19]. Because p can

64 be calculated from relative humidity and temperature, at least four sensors [(humidity +
65 temperature) × two points] are required. In addition, this method relies on the assumption that
66 the state of the outer atmosphere is unchanging, which is unlikely in daily perspiration
67 monitoring. To address this, a closed chamber system with water vapor condenser was proposed
68 [21,22], although the coolant (Peltier device) is required and thus power consumption would be
69 measurable. The theory of the ventilated chamber method is similar to that of the vapor pressure
70 method, except this method uses forced and constant airflow. The constant airflow is injected
71 into a small chamber adjacent to the skin, and the air with evaporated water vapor is collected in
72 an outlet chamber. The amount of water exchange is then calculated with the airflow rate and
73 difference of humidity between inlet and outlet air [19,20,23]. However, it is difficult to contain
74 air ventilator and chambers in one small package such that it can be of practical use in daily life.
75 Therefore, it was considered beneficial to develop a small device that could monitor
76 perspiration and allow prediction of emotional and physiological status.

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

80

81 **2. Materials and Methods**

82 *2.1. Developed device*

83 Fig. 1 shows the exterior (Fig. 1A, B) and structure (Fig. 1C) of the developed device. A
84 custom-made data logger circuit with battery, dry silica-gel (Wako Pure Chemical Industries,
85 Ltd., Osaka, Japan), and a small temperature/relative humidity (T/RH) sensor (SHT-21,
86 Sensirion AG, Zürich, Switzerland; accuracy of temperature is $\pm 0.3^{\circ}\text{C}$, accuracy of relative
87 humidity is $\pm 2\%$, calibrated at Industrial Research Institute of Ishikawa, Japan) with a sampling
88 rate of 1 Hz was encapsulated in this order in a small plastic chamber toward measuring
89 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 ($\text{g m}^{-2} \text{s}^{-1}$) between two
95 points can be calculated as Eq. (1):

$$J = -D \frac{dH}{dx} \quad (1)$$

96 where D ($\text{m}^2 \text{s}^{-1}$) 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):

$$\begin{aligned} V \frac{\Delta H_x(t)}{\Delta t} &= A_1 J_1 - A_2 J_2 \\ &= A_1 D \frac{H_1(t) - H_x(t)}{L_1} - A_2 D \frac{H_x(t) - H_2}{L_2} \end{aligned} \quad (2)$$

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):

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

112 where the rate of water vapor diffusion from the skin $W(t)$ ($\text{g m}^{-2} \text{s}^{-1}$) can be calculated only by
 113 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
 114 be calculated by using following Eq. (4) under normal atmospheric pressure [24]:

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

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

116 Here, the fixed values were set to: $V = 6.3 \times 10^{-7} \text{ m}^3$, $A_1 = 1.3 \times 10^{-5} \text{ m}^2$, $A_2 = 5.6 \times 10^{-5} \text{ m}^2$, L_2
 117 $= 5.0 \times 10^{-3} \text{ m}$ according to the equipment design, and H_2 was estimated to be 1.7 g m^{-3} (Fig.
 118 S1). Finally, the perspiration $Per(t)$ ($\text{mg cm}^{-2} \text{ min}^{-1}$) [20,23] can be calculated from $W(t)$ by a
 119 simple conversion Eq. (5):

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

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

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

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

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

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

133

134 *2.3. Verification of developed equipment*

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

148

149 *2.4. Measurement of perspiration evoked by sympathetic activity*

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

159

160 2.5. *Ethical approval*

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

163

164 **3. Results**

165 *3.1. Perspiration recording by the developed and conventional devices*

166 As a result of performance test, the developed device was confirmed to be able to measure the
167 water exchange with an uncertainty of $<\pm 5\%$ and a long-term stability of >4 h (Fig. S2). The
168 representative temporal changes in perspiration recorded by the developed and conventional
169 devices are shown in Fig. 3A, B (the baseline data were shown in Fig. S3). The comparable time
170 profiles of perspiration in a steady state were observed (Fig. 3A, B). Perspiration pattern
171 determined by the developed device was in good agreement with that of the conventional device
172 [Fig. 3C; peak correlation coefficient of 0.738 at -6 s, and Fig. 3D; ICC(2, 1) of the
173 corresponding peak-to-peak amplitudes was 0.84 with the 95% confidence interval (CI) of
174 0.76–0.90]. The Bland–Altman plot revealed a fair agreement between both devices [Fig. 3E;
175 bias = -0.0042 mg cm⁻² min⁻¹ (95% CI: -0.014 – 0.0056) with the limits of agreement -0.087 to
176 0.079]. These results imply that the developed device could capture perspiration as efficiently as
177 the conventional one. The other subjects showed similar results (data not shown).

178

179 *3.2. Detection of palmar perspiration evoked by the sympathetic activity using the developed*
180 *device*

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

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

191

192 **4. Discussion**

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

200 In this study, a modification of the vapor pressure method [19] was utilized. In the
201 conventional vapor pressure method, water exchange on the skin (i.e., constant water loss and
202 perspiration) can be detected as the natural flow of water vapor. However, the flow from the
203 skin to ambient air is dependent on outer air conditions such as temperature and humidity, i.e.,
204 the previous method would be deeply affected by the nature of outer atmosphere. Thus, a
205 combination of a closed chamber with enforced ventilation has been developed [20,23,34].
206 These combined methods use dehumidified nitrogen or ventilation pumps, which could hamper
207 daily monitoring of perspiration patterns. To address these limitations, a closed-chamber filled
208 with silica gel above the T/RH sensor was developed (Figs. 1 and 2). In this system, the
209 adsorption of water vapor into silica gel generates a natural but constant flow of water vapor.
210 Under such a constant flow, the T/RH sensor below the silica gel can constantly measure the
211 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 $10 \text{ mg cm}^{-2} \text{ min}^{-1}$ water vapor
263 evaporation is simulated, which is beyond the observed perspiration plus water loss (Figs. 3A

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

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

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

278

279 **Acknowledgments**

280 We would like to thank Mr. Ryohei Suganuma and Ms. Sakie Tachibana, who helped perform
281 the experiments. This study was supported in part by JSPS KAKENHI Grant Numbers
282 15K20664 (to KO), 24500848 (to KK), and 21500405 (to TN). This study was also funded in
283 part by the MEXT/JST Tenure Track Promotion Program (to KO). A part of this study was
284 based on the Japanese Unexamined Patent Application Publication Nos. 2011-169881 and
285 2012-085983.

286

287 **Ethical Approval**

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

289

290 **Conflict of interest**

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

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

293

294 **References**

295 [1] Hardy JD. Physiology of temperature regulation. *Physiol Rev.* 1961;41:221.

296 [2] Drummond PD, Lance JW. Facial flushing and sweating mediated by the sympathetic
297 nervous system. *Brain.* 1987;110:793-803.

298 [3] van Dooren M, De Vries J, Janssen JH. Emotional sweating across the body: Comparing 16
299 different skin conductance measurement locations. *Physiol Behav.* 2012;106:298-304.

300 [4] Kamei T, Tsuda T, Kitagawa S, Naitoh K, Nakashima K, Ohhashi T. Physical stimuli and
301 emotional stress-induced sweat secretions in the human palm and forehead. *Anal Chim Acta.*
302 1998;365:319-26.

303 [5] Niepomnische H, Amad RH. Skin disorders and thyroid diseases. *J Endocrinol Invest.*
304 2001;24:628-38.

305 [6] Leung AK, Chan PY, Choi MC. Hyperhidrosis. *Int J Dermatol.* 1999;38:561-7.

306 [7] Kronenberg F, Cote LJ, Linkie DM, Dyrenfurth I, Downey JA. Menopausal hot flashes:
307 thermoregulatory, cardiovascular, and circulating catecholamine and LH changes. *Maturitas.*
308 1984;6:31-43.

309 [8] Davidson JR, Foa EB, Connor KM, Churchill LE. Hyperhidrosis in social anxiety disorder.
310 *Prog Neuropsychopharmacol Biol Psychiatry.* 2002;26:1327-31.

311 [9] Brunelli GA, Brunelli GR. Preoperative assessment of the adult plexus patient. *Microsurgery.*
312 1995;16:17-21.

313 [10] Chelimsky TC, Low PA, Naessens JM, Wilson PR, Amadio PC, O'Brien PC. Value of
314 autonomic testing in reflex sympathetic dystrophy. *Mayo Clin Proc.* 1995;70:1029-40.

315 [11] Anand P, Birch R. Restoration of sensory function and lack of long-term chronic pain

316 syndromes after brachial plexus injury in human neonates. *Brain*. 2002;125:113-22.

317 [12] Pickup JC. Preliminary evaluation of a skin conductance meter for detecting hypoglycemia
318 in diabetic patients. *Diabetes Care*. 1982;5:326-9.

319 [13] Stenstrom SJ. A study on skin humidity in leprosy patients using a new type of humidity
320 meter. *Int J Lepr Other Mycobact Dis*. 1984;52:10-8.

321 [14] Chang BW, Yeh SJ, Tsai PP, Chang HC. Monitoring perspiration from palms of
322 hypohidrosis patients with a stopped-flow conductometric mini-system. *Clin Chim Acta*.
323 2004;348:107-11.

324 [15] Jia W, Bandodkar AJ, Valdes-Ramirez G, Windmiller JR, Yang Z, Ramirez J, et al.
325 Electrochemical tattoo biosensors for real-time noninvasive lactate monitoring in human
326 perspiration. *Anal Chem*. 2013;85:6553-60.

327 [16] Rose D, Ratterman M, Griffin D, Hou L, Kelley-Loughnane N, Naik R, et al. Adhesive
328 RFID Sensor Patch for Monitoring of Sweat Electrolytes. *IEEE Trans Biomed Eng*.
329 2015;62:1457-65.

330 [17] Huang X, Liu Y, Chen K, Shin WJ, Lu CJ, Kong GW, et al. Stretchable, wireless sensors
331 and functional substrates for epidermal characterization of sweat. *Small*. 2014;10:3083-90.

332 [18] Matzeu G, Fay C, Vaillant A, Coyle S, Diamond D. A wearable device for monitoring
333 sweat rates via image analysis. *IEEE Trans Biomed Eng*. 2015.

334 [19] Nilsson GE. Measurement of water exchange through skin. *Med Biol Eng Comput*.
335 1977;15:209-18.

336 [20] Sakaguchi M, Ono N, Ohhashi T. A new skin moisture meter using absolute hygrosensor.
337 Technical report of IEICE. 1998;98:43-7.

338 [21] Imhof RE, Berg EP, Chilcott RP, Ciortea LI, Pascut FC, Xiao P. New instrument for
339 measuring water vapour flux density from arbitrary surfaces. *IFSCC Magazine*.
340 2002;5:297-301.

341 [22] Imhof RE, De Jesus ME, Xiao P, Ciortea LI, Berg EP. Closed-chamber transepidermal

342 water loss measurement: microclimate, calibration and performance. *Int J Cosmet Sci.*
343 2009;31:97-118.

344 [23] Sakaguchi M, Kuroda K, Nakashima K, Ohhe K, Togari Y, Ohhashi T. Development of the
345 new ventilation capsule type sweating-evaporation ratemeter -Measurements of local sweating
346 rates and evaporation rates-. Technical report of IEICE. 2006;106:65-8.

347 [24] Denny MW, Meeting ASoZ. *Air and Water: The Biology and Physics of Life's Media:*
348 Princeton University Press; 1993.

349 [25] ASHRAE. Psychrometrics. In: Owen MS, editor. *ASHRAE Handbook - Fundamentals (SI).*
350 Atlanta: American Society of Heating, Refrigerating and Air-Conditioning Engineers; 2013. p.
351 1-16.

352 [26] Wilson D, Maibach H. Transepidermal water loss: a review. In: Lévêque J, editor.
353 *Cutaneous Investigations in Health and Disease: Non-invasive Methods and Instrumentation.*
354 New York: Marcel Dekker; 1989. p. 113-34.

355 [27] Martin RF. General deming regression for estimating systematic bias and its confidence
356 interval in method-comparison studies. *Clin Chem.* 2000;46:100-4.

357 [28] Shrout PE, Fleiss JL. Intraclass correlations: uses in assessing rater reliability. *Psychol Bull.*
358 1979;86:420-8.

359 [29] Plichta MM, Schwarz AJ, Grimm O, Morgen K, Mier D, Haddad L, et al. Test-retest
360 reliability of evoked BOLD signals from a cognitive-emotive fMRI test battery. *Neuroimage.*
361 2012;60:1746-58.

362 [30] Bland JM, Altman DG. Statistical methods for assessing agreement between two methods
363 of clinical measurement. *Lancet.* 1986;1:307-10.

364 [31] Aarts LA, Jeanne V, Cleary JP, Lieber C, Nelson JS, Bambang Oetomo S, et al.
365 Non-contact heart rate monitoring utilizing camera photoplethysmography in the neonatal
366 intensive care unit - a pilot study. *Early Hum Dev.* 2013;89:943-8.

367 [32] Kobayashi M, Tomioka N, Ushiyama Y, Ohhashi T. Arithmetic calculation, deep inspiration

368 or handgrip exercise-mediated pre-operational active palmar sweating responses in humans.
369 *Auton Neurosci.* 2003;104:58-65.

370 [33] Ellaway PH, Kuppuswamy A, Nicotra A, Mathias CJ. Sweat production and the
371 sympathetic skin response: improving the clinical assessment of autonomic function. *Auton*
372 *Neurosci.* 2010;155:109-14.

373 [34] Low PA, Caskey PE, Tuck RR, Fealey RD, Dyck PJ. Quantitative sudomotor axon reflex
374 test in normal and neuropathic subjects. *Ann Neurol.* 1983;14:573-80.

375 [35] Appenzeller O, Goss JE. Autonomic deficits in Parkinson's syndrome. *Arch Neurol.*
376 1971;24:50-7.

377 [36] Cheshire WP, Tsuboi Y, Wszolek ZK. Physiologic assessment of autonomic dysfunction in
378 pallidopontonigral degeneration with N279K mutation in the tau gene on chromosome 17.
379 *Auton Neurosci.* 2002;102:71-7.

380 [37] Fealey RD, Low PA, Thomas JE. Thermoregulatory sweating abnormalities in diabetes
381 mellitus. *Mayo Clin Proc.* 1989;64:617-28.

382 [38] Korpelainen JT, Sotaniemi KA, Myllylä V. Hyperhidrosis as a reflection of autonomic
383 failure in patients with acute hemispherical brain infarction. An evaporimetric study. *Stroke.*
384 1992;23:1271-5.

385 [39] Micieli G, Tosi P, Marcheselli S, Cavallini A. Autonomic dysfunction in Parkinson's
386 disease. *Neurol Sci.* 2003;24 Suppl 1:S32-4.

387 [40] Sandroni P, Ahlskog JE, Fealey RD, Low PA. Autonomic involvement in extrapyramidal
388 and cerebellar disorders. *Clin Auton Res.* 1991;1:147-55.

389 [41] Stuart DD. Diabetic gustatory sweating. *Ann Intern Med.* 1978;89:223-4.

390 [42] Ng K, Chua H, Chung C, Loke C, Kashiwagi T, Akisawa A, et al. Experimental
391 investigation of the silica gel–water adsorption isotherm characteristics. *Appl Therm Eng.*
392 2001;21:1631-42.

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 $-0.0042 \text{ mg cm}^{-2} \text{ min}^{-1}$ 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)

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

423

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

427

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

437

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

Figure 1

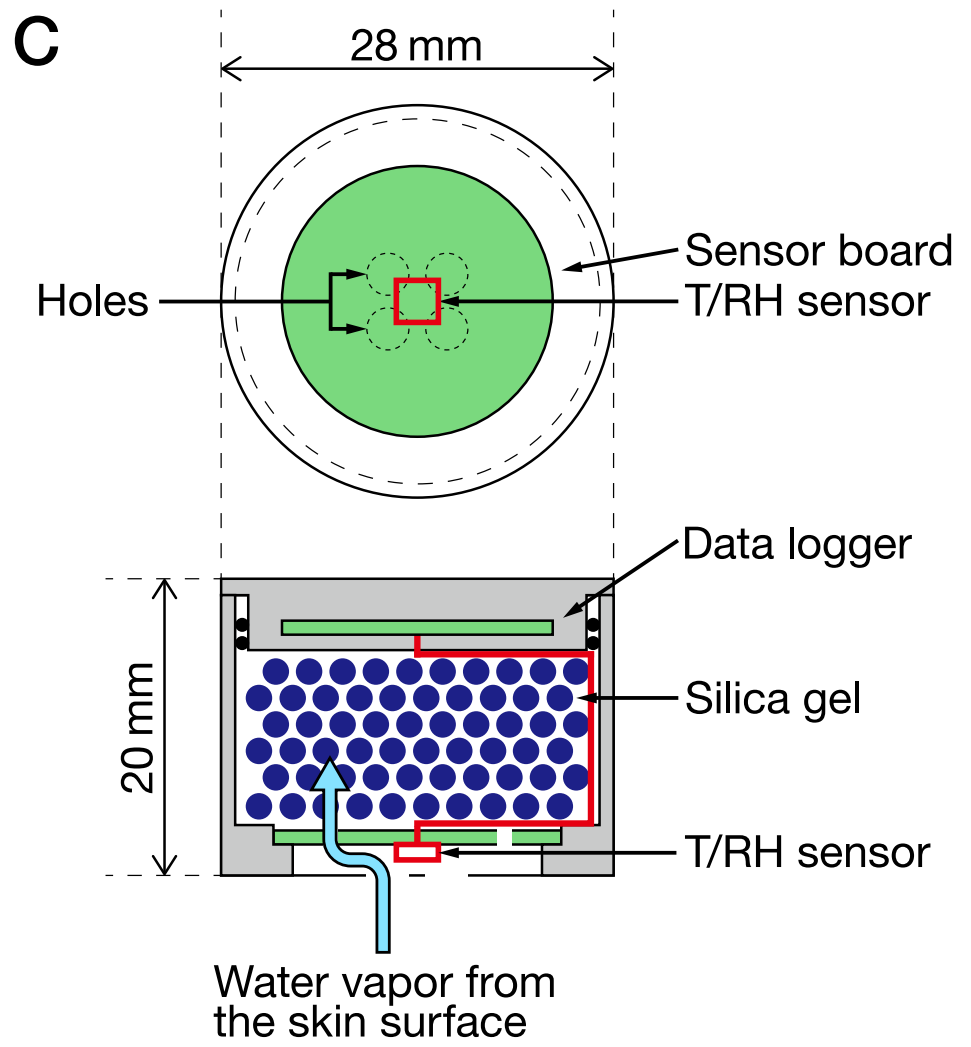
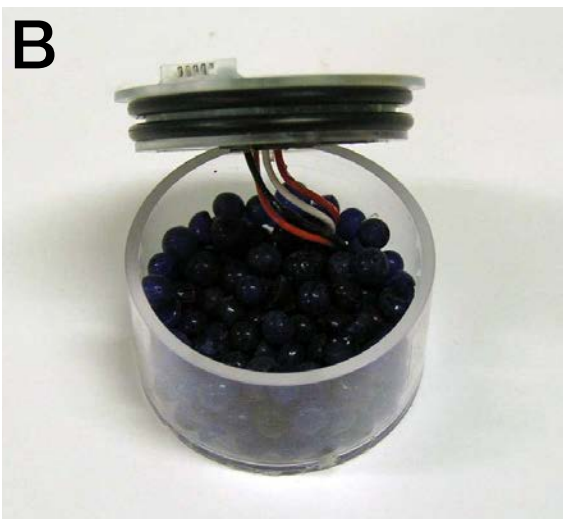
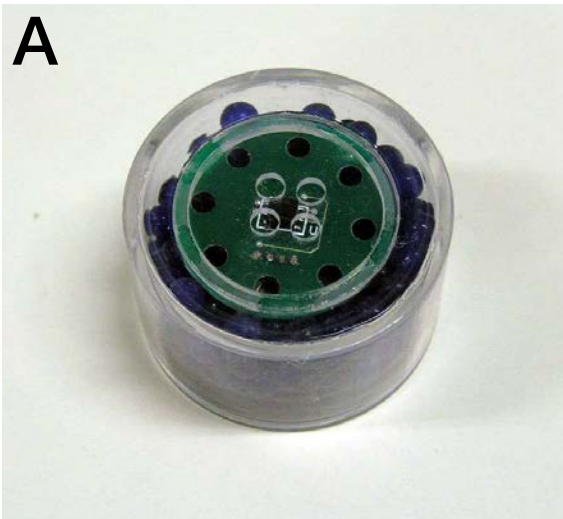
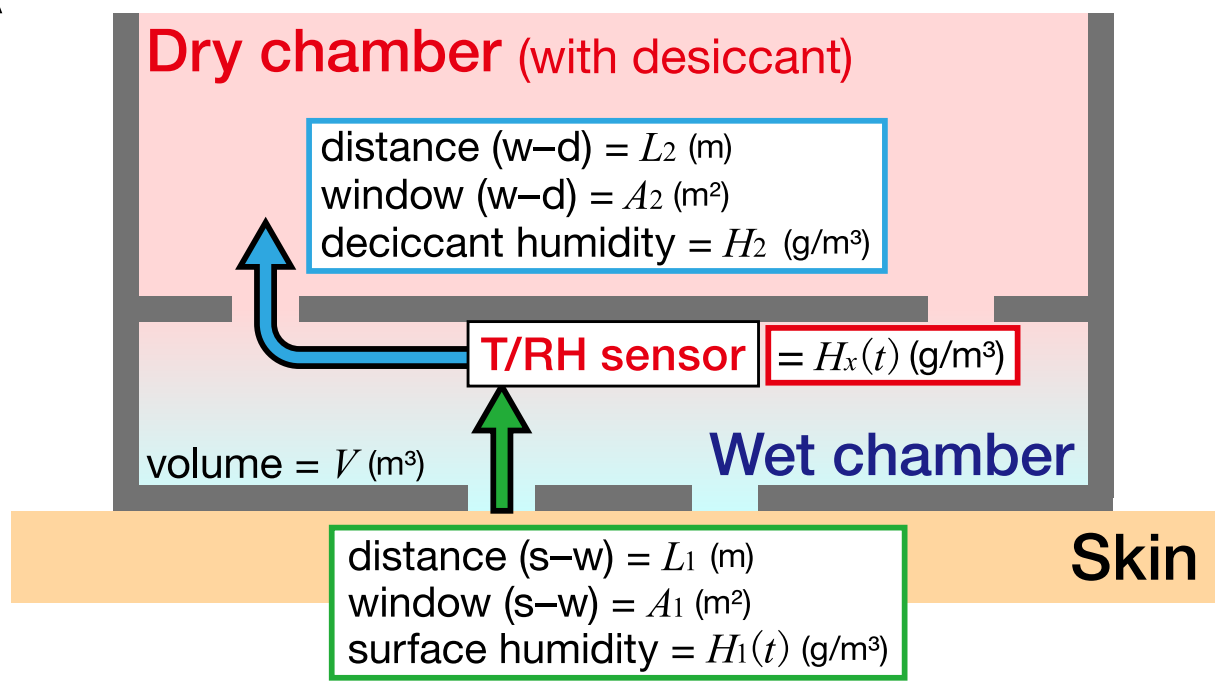


Figure2

A



B

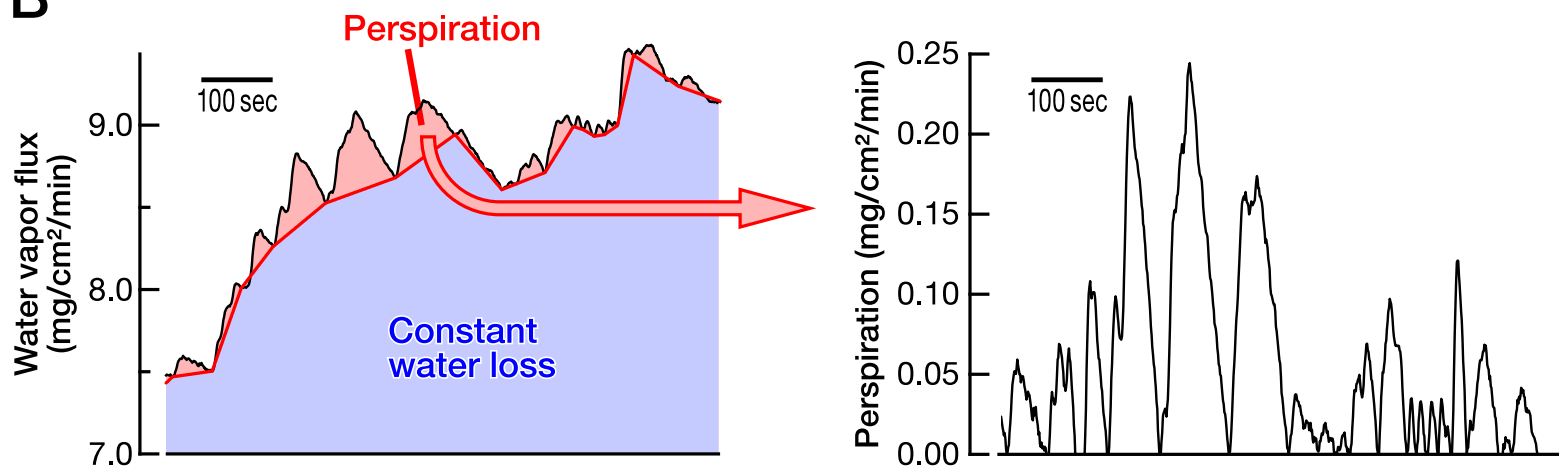


Figure 3

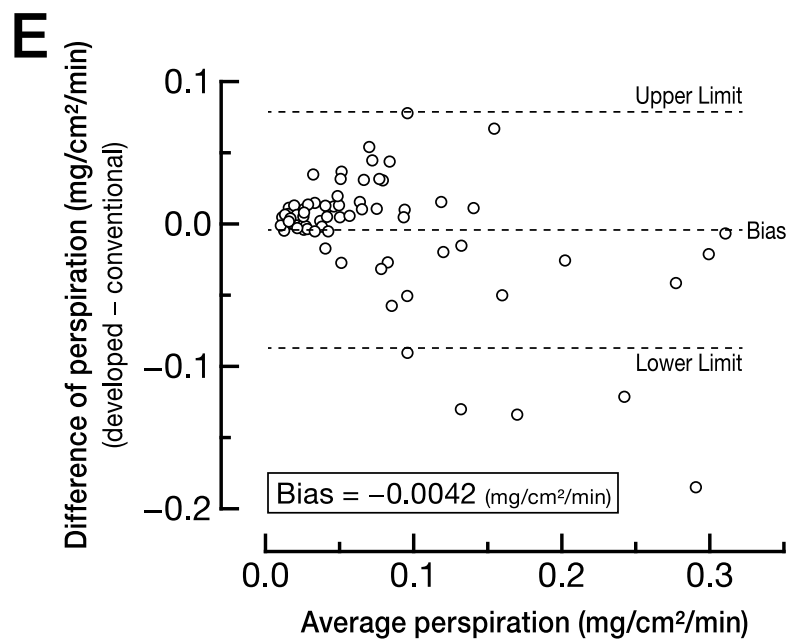
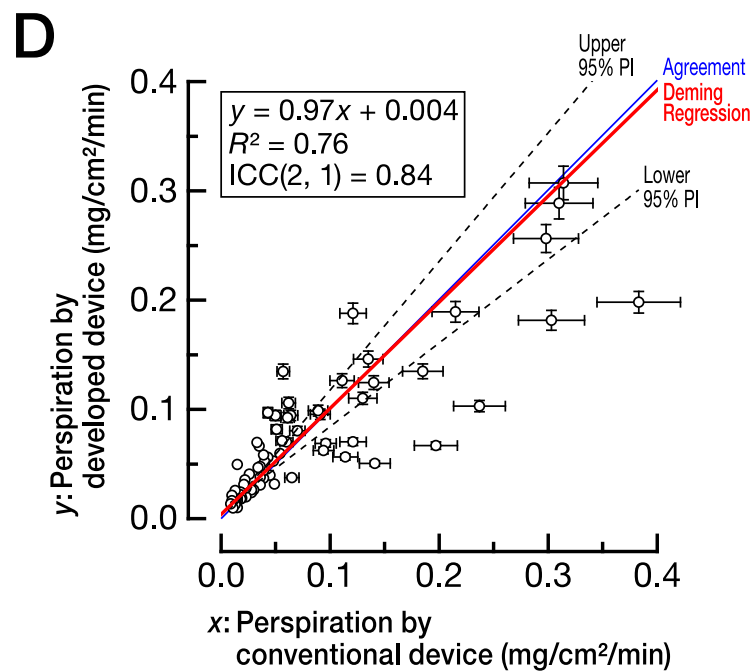
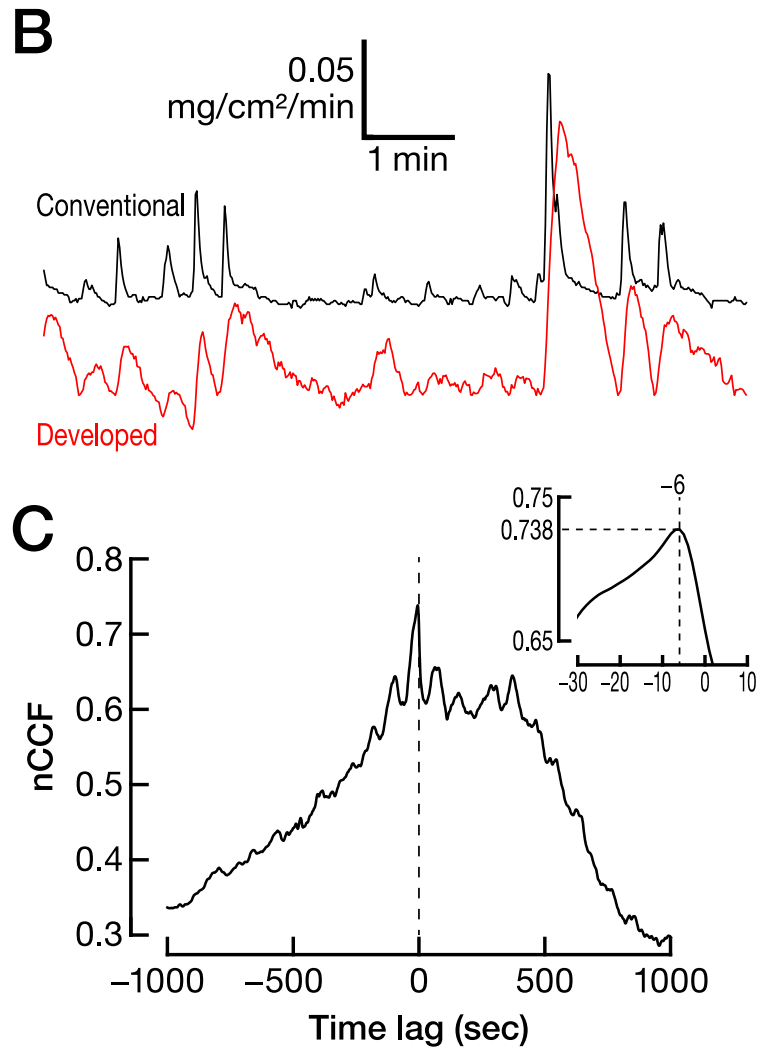
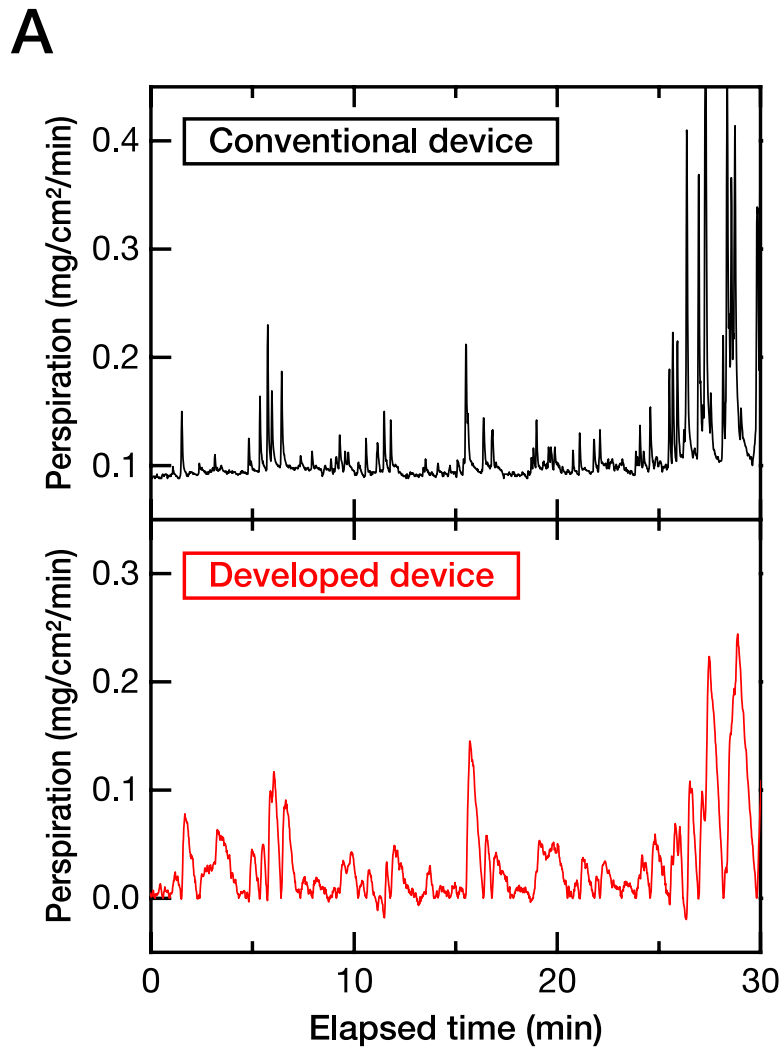
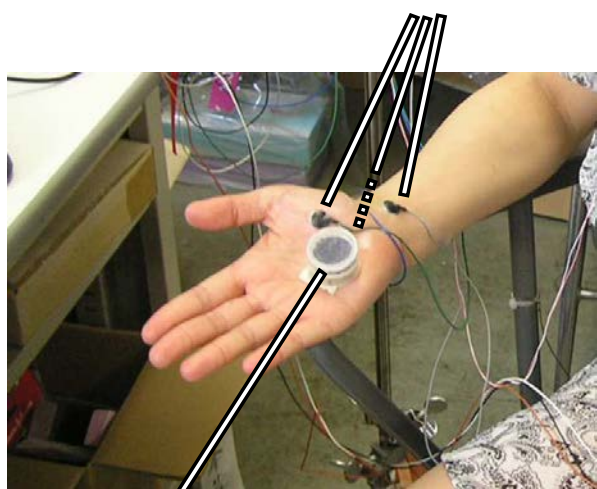


Figure4

A

SSR sensors

Two electrodes (on the palm and the back) with an indifferent electrode on the wrist



Developed device

B

