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

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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 by
23 the change in relative humidity and then adsorbed by silica gel, allowing continuous recording
24 of perspiration within a closed and wireless chamber, which has not been previously
25 achieved. By comparative experiments using the commercially-available perspiration monitoring
26 device, the developed device could measure perspiration as efficiently as the conventional one,
27 with a normalized cross coefficient of 0.738 with 6 s delay and the interclass correlation
28 coefficient [ICC(2, 1)] of 0.84. These results imply a good agreement between the conventional
29 and developed devices, and thus suggest the applicability of the developed device for
30 perspiration monitoring.

31

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

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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 devices 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) $\times$ 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
67 perspiration monitoring. To address this, a closed chamber system with water vapor condenser
68 was proposed [21,22], although the coolant (Peltier device) is required and thus power
69 consumption would be measurable. The theory of the ventilated chamber method is similar to
70 that of the vapor pressure method, except this method uses forced and constant airflow. The
71 constant airflow is injected into a small chamber adjacent to the skin, and the air with
72 evaporated water vapor is collected in an outlet chamber. The amount of water exchange is then
73 calculated with the airflow rate and difference of humidity between inlet and outlet air
74 [19,20,23]. However, it is difficult to contain air ventilator and chambers in one
75 small package such that it can be of practical use in daily life. Therefore, it was considered
76 beneficial to develop a small device that could monitor perspiration and allow prediction of
77 emotional and physiological status.

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

81

82 **2. Materials and Methods**

83 *2.1. Developed device*

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

90 facing the skin (Fig. 1C).

91

92 2.2. Perspiration rate calculation

93 The theory of this equipment is based on the vapor pressure method [19,21] with modifications.

94 Fig. 2 illustrates the method of perspiration measurement in the developed equipment. According

95 to the Fick's law of diffusion, the flux of water vapor $J(\text{g m}^{-2} \text{s}^{-1})$ between two points can be

96 calculated as Eq. (1):

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

97 where $D(\text{m}^2 \text{s}^{-1})$ is a diffusion coefficient of water vapor in the air, $dH(\text{g m}^{-3})$ is a difference of

98 concentration of water vapor, and $dx(\text{m})$ is a distance between two points. In the developed

99 equipment, two different fluxes of water vapor: from the skin surface to the T/RH sensor (Fig.

100 2A, green arrow; **s-w**), and from the sensor to the dry chamber (Fig. 2A, blue arrow; **w-d**), can

101 be theorized. The flux difference between (**s-w**) and (**w-d**) could be detected as a change of

102 humidity in T/RH sensor. The water exchange between the skin and silica gel via T/RH sensor

103 should satisfy Eq. (2):

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

104 where $V(\text{m}^3)$ is a volume of wet chamber in which the T/RH sensor exists; $H_1(t)$, $H_x(t)$, and $H_2(\text{g}$

105 $\text{m}^{-3})$ are the concentrations (i.e., absolute humidity) of water vapor at the skin surface, T/RH

106 sensor, and dry chamber, respectively; A_1 and $A_2(\text{m}^2)$ are the areas of windows at (**s-w**) and

107 (**w-d**) junctions, respectively; J_1 and J_2 are the fluxes of (**s-w**) and (**w-d**), respectively; L_1 and

108 $L_2(\text{m})$ are the distances of (**s-w**) and (**w-d**), respectively (Fig. 2A). H_2 is assumed to be constant

109 due to a buffering effect of desiccant (preliminary experiment is shown in Fig. S1). Because J_1 in

110 Eq. (2) simply represents the flux of total water vapor from the skin surface [i.e., perspiration and

111 constant transepidermal water loss (TEWL)], the Eq. (2) can be solved for J_1 as following Eq.

112 (3):

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

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

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

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

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

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

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

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

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

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

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

134

135 *2.3. Verification of developed equipment*

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

149

150 *2.4. Measurement of perspiration evoked by sympathetic activity*

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

160

161 *2.5. Ethical approval*

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

164

165 **3. Results**

166 *3.1. Perspiration recording by the developed and conventional devices*

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

179

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

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

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

192

193 **4. Discussion**

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

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

212 without the interference of ambient air in a small and wireless exterior (Figs. 1 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 to
227 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 (Fig.
230 3). The high cross correlation ($n\text{CCR} = 0.738$) and interclass correlation coefficient of the
231 peak-to-peak values [$\text{ICC}(2, 1) = 0.84$] indicate the considerable agreement between the
232 conventional and developed equipment (Fig. 3). In addition, the developed device could
233 detect perspiration evoked by mental stress (Fig. 4) like as the conventional device [32,33],
234 which 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
241 subtraction. The data of baseline themselves might have information on TEWL, although further
242 analysis 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 could
249 be achieved by incorporating such errors, although the calibration of the T/RH sensor and the
250 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.

264 3A 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 in
268 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
277 perspiration monitoring, and could predict the onset of the diseases related to
278 perspiration abnormalities.

279

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287

288 **Ethical Approval**

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

290

291 **Conflict of interest**

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

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

294

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396 **Figure legends**

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

400

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

407

408 **Figure 3** Comparison of the temporal patterns of perspiration.(A) Perspiration patterns obtained
409 by the conventional (upper) and developed (lower) devices are shown. (B) These devices
410 showed similar patterns of perspiration. (C) The normalized cross correlation function (nCCF)
411 of these devices. Note a good correlation (0.738) between the conventional and developed
412 devices with a short delay (-6 s). (D, E) Agreement of peak-to-peak values by the two devices.
413 (D) Scatter plot of conventional (x-axis) and developed (y-axis) devices with linear Deming
414 regression (red line) and 95% prediction interval (PI). These devices showed a good agreement,
415 with a high absolute interclass correlation coefficient [ICC(2, 1) = 0.84]. A blue line indicates
416 complete agreement. (E) Bland-Altman plot with the limits of agreement (bias = -0.0042 mg
417 $\text{cm}^{-2} \text{min}^{-1}$ with the limits of agreement -0.087 to 0.079).

418

419 **Figure 4** Detection of sympathetic palmar perspiration.(A) Experimental condition: the subject

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

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Figure1

Figure2

A

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Figure3

Figure4