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# Improved Estimation of Sweating Based on Electrical Properties of Skin

CHRISTIAN TRONSTAD,[1] HÅVARD KALVØY,[1] SVERRE GRIMNES,[1,2] and ØRJAN G. MARTINSEN[2,1]

[1]Department of Clinical and Biomedical Engineering, Rikshospitalet, Oslo University Hospital, Oslo, Norway; and [2]Department of Physics, University of Oslo, Oslo, Norway

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**Abstract**—Skin conductance (SC) has previously been reported to correlate strongly with sweat rate (Swr) within subjects, but weakly between subjects. Using a new solution for simultaneous recording of SC, skin susceptance (SS) and skin potential (SP) at the same skin site, the aim of this study was to assess how accurately sweat production can be estimated based on combining these electrical properties of skin. In 40 subjects, SC, SS, SP and Swr by skin water loss was measured during relaxation and mental stress. SC and Swr had high intraindividual correlations (median $r = 0.77$). Stepwise multilinear regression with bootstrap validation lead to a sweating estimation model based on the sum of SC increases, the SP area under the curve and the SS area under the curve, yielding an interindividual accuracy of $R^2 = 0.73$, rmse = 12.9%, limits of agreement of +27.6, −30.4% and an intraclass correlation coefficient of 0.84. Bootstrapping of training and test-sets gave median rmse = 15.4%, median $R^2 = 0.66$. The model was also validated for intraindividual variability. The results show that estimation of sweating is significantly improved by the addition of SS and SP measurement.

**Keywords**—Sweating, Sweat rate, Skin conductance, Skin potential, Skin susceptance, Bioimpedance, Multilinear model.

## INTRODUCTION

It is well known that the *in vivo* electrical properties of skin are heavily influenced by sweating.[15] The outermost layer of the skin, the Stratum Corneum, provides a strong electrical barrier of dead skin cells when dry, but when sweat pores are filled with sweat, pathways of ionic transport are created which greatly reduces this electrical impedance through the skin. This provides a very sensitive measurement of sweat gland activity by means of the skin conductance (SC), and is widely used as a physiological parameter within many biomedical applications.[3] Sweating also causes changes in the bioelectric potential (SP) at the skin surface, but the characteristics of this parameter are different from SC. While the SC always increases with the sweat duct filling and recovers while sweat is reabsorbed and/or pores are closed, the SP can change in both positive and negative voltage directions and produce biphasic or triphasic responses.[3] A third electrical parameter to consider is the skin capacitance, which is positively correlated with the moisture content of the stratum corneum,[19] and is most accurately measured by low-frequency skin susceptance (SS).[18] Although sweating, here defined as the amount of water loss from the skin by evaporation, and changes in the skin electrical properties both originate from the sweat gland, they are governed by completely different biophysical mechanisms. Sweating includes transport of sweat to the skin surface, evaporation and further transport of water molecules to the air. Some of the sweat is also absorbed in the skin and later evaporates at a decreasing rate, making the evaporation also dependent on the skin sorption characteristics. SC also increases as sweat fills the pores, but this is not necessarily dependent on sweat reaching the skin surface.[10] After sweat pore filling, the SC decreases as electrolytes are reabsorbed through the sweat duct wall. The same mechanisms govern the SP response although the waveform is more complex.[10] Despite these biophysical differences, mainly the movement of water vs. the movement of electrolytes, high within-subject correlations have been reported between SC and skin water loss (up to $r = 0.88$,[21] $r > 0.85$[9]) and between the positive SP response and water loss ($r = 0.93$[12]). Between-subjects correlations on the

Address correspondence to Christian Tronstad, Department of Clinical and Biomedical Engineering, Rikshospitalet, Oslo University Hospital, Oslo, Norway. Electronic mail: christian.tronstad@gmail.com

other hand, is reported to be lower than $r < 0.50$ as a rule[8] and between $r = 0.23$–$0.64$,[21] suggesting one or more individual factors in the relation between electrical and evaporative skin properties. These factors could be differences in sweat electrolyte concentration,[17] thickness of the stratum corneum[11] or the skin sorption characteristics. The SS parameter could approach some of this interindividual variance, as SS is not directly affected by sweat pore filling and reflects only the stratum corneum moisture content.[18] Recently, a method for measuring SC, SP and SS at the same skin site was introduced by Grimnes *et al.*,[14] enabling a new look on the relation between sweating and electrical properties of skin by inclusion of more parameters.

Compared to methods for estimating sweating by means of evaporation and water loss measurement, the electrical methods have numerous advantages as they enable miniaturization[27] and long-term ambulatory recording,[24] lower costs of equipment and are much less susceptible to movement artifacts as the water loss measurement method requires a constant contact pressure of the probe onto the skin.[23] The high reported intraindividual correlations suggest that the electrical method already is usable for studies involving repeated measurements on the same subjects. An improved interindividual accuracy would also enable studies involving comparison in sweating between subjects or groups based on electrical measurement.

The aim of this study was to assess how accurately sweat production can be estimated based on combining SC, SP and SS. This was implemented by the following list:

1. Measure sweat rate (Swr) and the electrical properties SC, SS and SP during variations in sweating within and between subjects.
2. Analyze individual temporal correlations between changes in Swr and electrical properties.
3. Investigate between-subjects correlations between sweating and electrical parameters.
4. Develop a model for estimating sweat production based on significant parameters.
5. Validate the model for both between-subjects and within-subject variance.

## MATERIALS AND METHODS

### *Skin Admittance and Potential Measurement*

Based on the solution described in Grimnes *et al.*,[14] a PC-based EDA measuring system for simultaneous recording of skin complex admittance (SY) and SP at the same electrode was developed, consisting of front-end electronics connected *via* a National Instruments® DAQ-card to a laptop running software written in LabVIEW® v8.5, similar to the instrumentation presented in Tronstad *et al.*[28] Instead of a saline bath immersion of the underarm as a reference electrode, a novel three-electrode system was used which ensures unipolar recording of SC regardless of electrode areas, thus allowing the use of the same electrode type at all sites. The constant current control was replaced by a Howland current source giving a 10 Hz AC current of 14 $\mu$A to the skin. In addition to SC and SP measurements, the quadrature component of the SY signal was used to acquire SS. The differential amplifier was replaced by voltage sensing by analog-to-digital conversion at both terminals with software differencing, enabling control of the electrodermal inactivity at the reference site, which is a requirement for accurate SP recording.[3] A schematic of the measuring setup is shown in Fig. 1. In brief, the PC software controls the generation of an AC excitation signal which is fed to the Howland current source, producing a constant AC current ($i$) between the measurement (M) electrode at the hypothenar site and the current sink (C) electrode at the underarm. Due to the high impedance of the stratum corneum relative to the living tissue, the AC voltage at the reference site (R) represents the AC voltage in the equipotential viable skin layers beneath the stratum corneum ($u_C$), and the unipolar skin impedance below the M electrode can thus be found from the AC voltage at the M lead ($u_C + u_M$) subtracted by the AC voltage at the R lead, divided by the known current $i$. This AC signal is then processed by phase-sensitive rectification in software to extract the in-phase and quadrature components for calculation of the SC and SS. The SP is found from the DC voltage difference between the M potential ($U_C + U_M$) and the potential at the R electrode ($U_C + U_R$) placed at the apex of the elbow which is an electrodermally inactive area. Although abrasion[8] or skin drilling[30] is a recommended pretreatment for the inactive site, no pretreatment of the skin was used in this study to avoid risk of contamination and the need for associated procedures. The electrode type used was Arbo® Kittycat™, according to recommendations in Tronstad *et al.*[29]

### *Sweat Rate Measurement*

Swr was measured by using the Q-sweat (WR Medical Electronics Co, Stillwater, USA). The Q-Sweat uses dry air (room air, drawn across a desiccant) to pick up moisture from a measurement capsule (5.06 $cm^2$ circular measurement area) placed on the skin. Constant airflow (60 $cm^3$/min) through the capsule transports the captured moisture to the temperature and humidity sensors in the device. The airflow back and forth from the capsule to the Q-sweat passes through two 2.4 m air hoses.

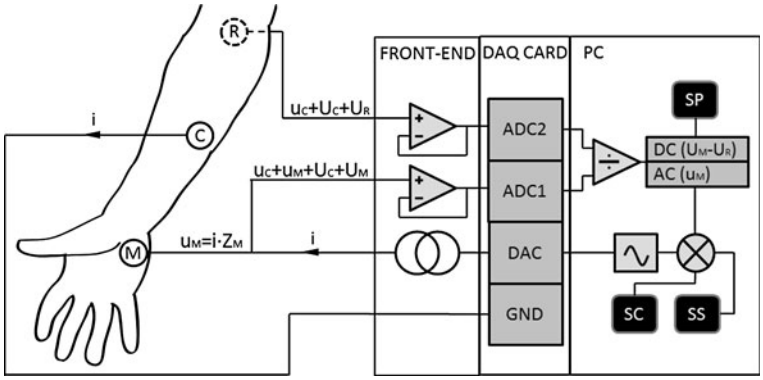


**FIGURE 1. Schematic of the setup for measuring SC, SS and SP using a PC with signal generation and processing software together with a data acquisition (DAQ) card utilizing one digital-to-analog converter (DAC) and two analog-to-digital converters (ADC) before front-end electronics for current injection to- and voltage pickup from the skin. Measuring (M), current sink (C) and reference (R) electrodes are placed at the hypothenar, underarm and elbow sites respectively. The lead voltages are represented in lower case (*u*) for AC and upper case (*U*) for DC. In the PC software, the two digitized signals are processed by differentiation before separation into a DC component used to determine SP, and an AC component used to determine SC and SS by phase-sensitive rectification.**

The skin capsule was attached on the hypothenar eminence of a randomly picked hand on each test subject (the opposite of the bioelectrical electrode). This was done by first attaching the end of the silicone band to the capsule and turn the silicone band around the wrist and back over mounting spikes on the capsule. Then the remaining half of the silicone band was turned around the palm distal to the thumb and back to the capsule, where the end of the band was attached to the mounting spikes. This way the silicone band formed a figure of eight with the middle cross attached to the capsule. One of the circles forming the figure of eight was wrapped around the wrist, similar to a wrist band. The other half was the silicone band on the meta carpus, passing to the dorsal side of the hand distal to the thumb. The capsule position was adjusted by varying the tension in the silicone band in the four directions of the cross covering the capsule. This was done to even out the pressure between the capsule and the skin around the capsule, until a tight fit was obtained and no leakage could be detected by the Q-sweat software (WR-TestWorks 2.0.1). To further prevent air leakage, the test subject was asked to find a comfortable position for the hand and move it as little as possible during the measurements.

A step-response test of the QSweat system was performed before the experiments started to test its responsiveness. By sliding a probe from a dry to a wet surface, a response delay of 8 s and a time-constant of 19.2 s was found.

### *Experimental Protocol*

40 healthy volunteers (31 male, 9 female, age 17–64), were recruited from Oslo University Hospital and University of Oslo and gave informed consent before participation. The study was approved by the regional ethics committee (REK #2010/1927a).

The experiments were done in silent laboratory or meeting rooms with only the test subject and two operators present. Room temperature was 22.5 ± 1 °C. The ambient humidity was neither controlled nor measured during the experiments. After bilateral fixation of EDA and QSweat sensors to the hypothenar area on both palms, the Swr measurement was monitored until stable levels were attained. The experiment was aborted if the Swr stabilized above the maximum of the QSweat measuring range (1000 nL/min). At least 5 min were allowed for stabilization of EDA electrodes before the recording started. In order to produce variations in sweating and thereby duct filling over time, intervals of relaxation and mental stress were alternated. The subjects were sitting comfortably in a chair and were not allowed to speak with the operators. Mental stress was induced by asking the subject to repeatedly subtract seven from a starting number of 1000. This was repeated in three 2-min intervals, with 5 min for relaxation before and after, giving a total recording session of 26 min.

### *Data Analysis and Statistics*

#### *Signal Conditioning*

Due to noisy QSweat recordings, the time-series were low-pass filtered with a third-order zero-phase Butterworth filter with a 0.025 Hz cutoff frequency in order to dampen all noisy fluctuations sufficiently. The response delay and time-constant found from the step-test was used to correct the phase of the signal and the responsiveness by backwards-filtering through an inverted RC-filter with the measured time-constant.

The Swr recording with a sampling frequency of $F_s = 4$ Hz were aligned with the EDA recordings with $F_s = 10$ Hz by resampling to 10 Hz by linear interpolation using the time-series tools in Matlab.

### *Intraindividual Time-Series Analysis*

To assess the synchronicity between the SC and Swr recordings, the cross-covariance matrix between the two signals was calculated for each subject and the relative time-shift at the maximum was used to estimate the lag between each signal pair. A one-sample $t$ test was used to evaluate the significance of the lag.

Based on an insignificant lag, the correlation between the SC and Sw time-series were evaluated by the Pearson product–moment correlation coefficient.

12 Swr recordings which either saturated the measurement range (>1000 nL/min) or produced no increase above the measuring noise (5% threshold of 50 nL/min) during the stress provocation were excluded from the time-series correlation analysis.

### *Interindividual Correlations*

Similar to the time-series analysis, the recordings which saturated the Swr measurement range were excluded, but those with low responses (<50 nL/min) were included, giving a total of 33 subjects used in this analysis.

Total sweating (Sw) in $\mu$L/5.06 cm$^2$ for each subject was calculated by the area under the curve (AUC) of the Swr recording for the whole duration of the session and was used as a dependent variable in a multilinear regression with EDA parameters as independent variables. Table 1 gives a complete list of all the variables used in the analysis.

The NSCR parameter was calculated from a peak recognition algorithm using Matlab® where a peak at time t was recognized if there was 0.5 s or more of continuous increase above 0.01 $\mu$S on the left side and a decrease or flat 0.5 s to the right side of $t$.

The correlation between each pair of variables was determined by the Pearson product–moment correlation coefficient.

### *Model Development*

In order to sift out redundant independent variables and avoid overfitting, stepwise regression was first used to find the best explanatory model. The stepwise regression method adds or removes terms from a multilinear model based on their statistical significance in a regression. Beginning with no terms, one independent variable was added at a time based on the lowest $p$ value of its coefficients being zero if added to the model, until the model could no longer be improved by including more terms within the significance level ($p < 0.05$).

In order to validate the model suggested by the stepwise regression procedure, the model from each step was evaluated by bootstrapping. Random permutations were used to assign half of the dataset to a training set and the other half to a test set. The regression coefficients were then found from the training set and applied on the test set to calculate the root mean square error (rmse) of the estimation. This was also done on the model with all parameters for comparison. Distributions of the rmse for each model from 1000 bootstrap iterations were compared by a Kruskal–Wallis one way ANOVA on Ranks with Tukey *post hoc* pairwise multiple comparison tests. The final model was selected based on the fewest number of terms and an rmse not significantly higher than any other model.

### *Model Validation*

Agreement between the model and the Sw was presented as a scatterplot using the mean coefficients from the bootstrapping, together with the coefficient of determination, $R^2$, the Pearson product–moment correlation coefficient, $r$, the rmse and percent-wise rmse. A Bland–Altman plot was constructed from the mean and differences between the measured and estimated sweating, with 1.96 std.dev limits of agreement. Reliability was assessed by the two-way mixed intraclass correlation coefficient for absolute agreement (ICC).

**TABLE 1. Overview of all parameters extracted from the recordings and used in the statistical analyses.**

| Parameter | Description | Function | Unit |
|---|---|---|---|
| Swr | Sweat rate | Dependent variable | nL/min |
| SC | Skin conductance | Intraindividual covariate | $\mu$S |
| Sw | Sweating (Swr area under curve) | Dependent variable | $\mu$L |
| AUC SC | SC area under curve | Interindividual covariate | $\mu$S min |
| AUC SS | SS area under curve | Interindividual covariate | $\mu$S min |
| AUC SP | SP area under curve | Interindividual covariate | mV min |
| NSCR | Number of SC responses | Interindividual covariate | Dimensionless |
| SCpos | Sum of positive SC responses | Interindividual covariate | $\mu$S |
| SPpos | Sum of positive SP responses | Interindividual covariate | mV |
| SPneg | Sum of negative SP responses | Interindividual covariate | mV |

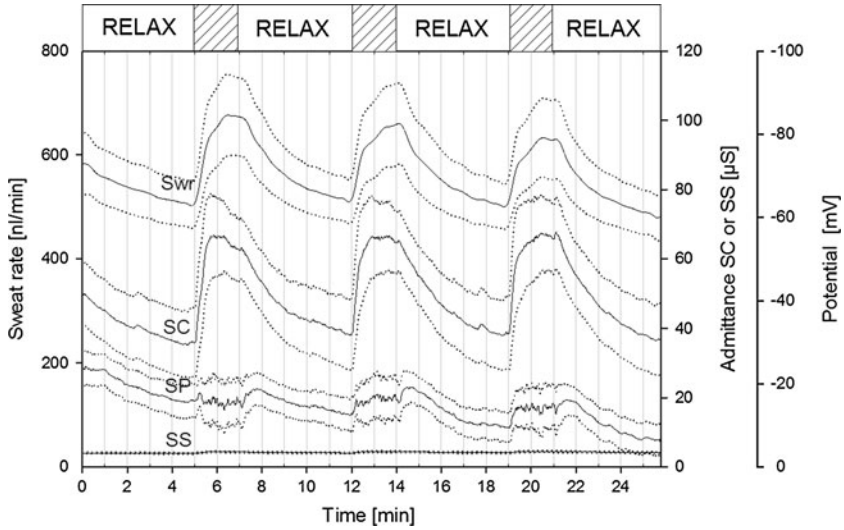


**FIGURE 2. Mean time series of Swr, SC, skin potential (SP) and skin susceptance (SS) with 95% confidence intervals (dotted lines). The upper boxes indicates the intervals of relaxation and stress by arithmetics (hatched).**

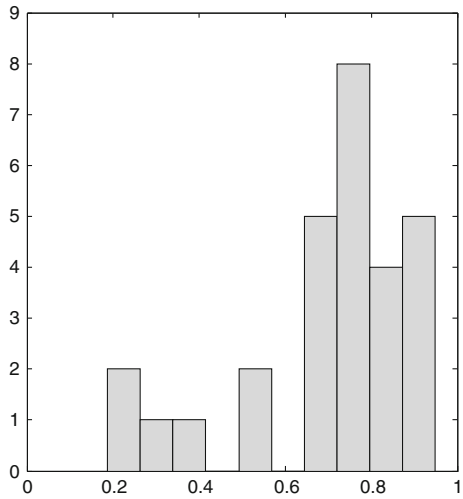

**FIGURE 3. Histogram of intraindividual correlation coefficients between Swr and SC ($N = 28$).**

In order to test the validity of the model upon intraindividual variability, estimation was assessed both during periods of high and low sweating according to the ends of the stress and relaxation intervals. Sw and all model parameters were extracted from the last minutes of all stress periods for the high sweating dataset, and during the last minutes for all relaxation periods, except the last, for the low sweating dataset. Sweating was estimated from the same mean coefficients from the total dataset. The two estimations were compared by their linear regression trend-lines from scatterplots between measured and estimated sweating within the same graph.

All data-analysis was done using Matlab R2011 a, Sigmaplot 11.0 and SPSS 19.

## RESULTS

### *Intraindividual Time-Series Analysis*

Based on the plots in Fig. 2 and the cross-covariance analysis, the SC and Swr signals were highly synchronized, with an insignificant ($p = 0.19$, one-sample $t$ test against zero) lag of the Swr with median 1.05 s, 25% percentile −3.6 s and 75% percentile of 12.05 s.

The temporal correlation between the SC and Swr time-series was median 0.77 ($p < 0.001$), 25% percentile 0.67 and 75% percentile 0.84 with a histogram as shown in Fig. 3. The correlation between the means of SC and Swr was 0.94 ($p < 0.001$).

### *Interindividual Correlations*

Pearson product–moment correlation coefficients for all interindividual parameters are presented in Table 2 ($N = 40$). Sw correlated strongest with SCpos ($r = 0.76$) and NSCR ($r = 0.65$), and these two parameters were strongly correlated to each other ($r = 0.77$).

### *Model Development*

Multilinear regression using all parameters gave an $R^2 = 0.84$, $r = 0.91$, rmse = 1.7 $\mu$L. The stepwise regression lead to a model including the significant terms of SCpos, AUC SP, AUC SS and AUC SC in the given order ($R^2 = 0.82$, $r = 0.90$, rmse = 1.7 $\mu$L). Validation of this model by bootstrapping lead to rmse distributions as shown in Fig. 4. Based on the ANOVA with pairwise comparisons, each step lead to a significant model improvement except for the last step with

**TABLE 2. Pearson product moment correlation coefficients between all interindividual parameters.**

| Pearson's *r* | AUC SC | AUC SS | AUC SP | NSCR | SCpos | SPpos | SPneg |
|---|---|---|---|---|---|---|---|
| Sw | 0.49** | 0.39* | 0.29 | 0.65*** | 0.76*** | 0.43** | 0.17 |
| AUC SC | | 0.48** | 0.02 | 0.79*** | 0.66*** | 0.02 | 0.014 |
| AUC SS | | | *0.13* | 0.23 | *0.22* | 0.11 | 0.35* |
| AUC SP | | | | −0.08 | *0.19* | 0.28 | −0.02 |
| NSCR | | | | | 0.77*** | 0.2 | 0.12 |
| SCpos | | | | | | 0.55*** | −0.17 |
| SPpos | | | | | | | −0.21 |

*$p < 0.05$; **$p < 0.01$; ***$p < 0.001$.

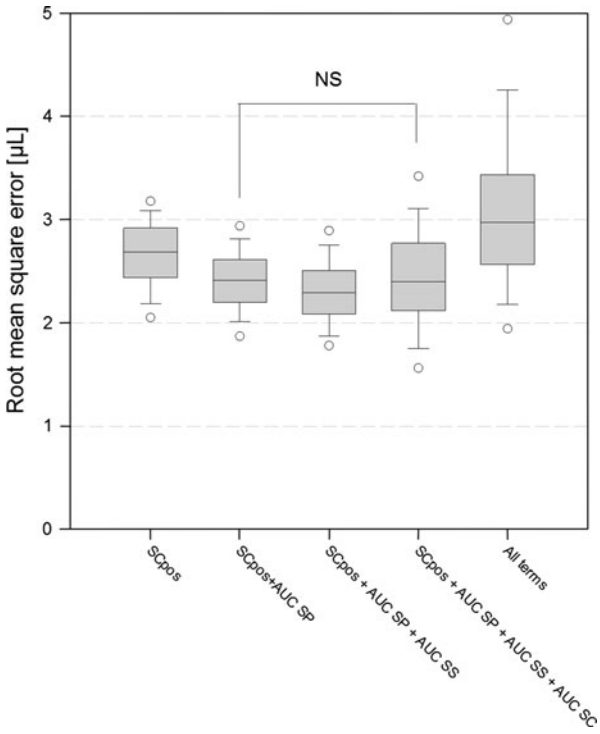


**FIGURE 4. Distributions of rmse from 1000 bootstrap validation of models from the stepwise regression and the model including all terms. All distributions were significantly different except for model number two and four ($p < 0.05$, Tukey Test for pairwise multiple comparison). Circles represent 5 and 95% percentiles.**

the inclusion of AUC SC, which significantly increased the rmse. Thus, the SCpos + AUC SP + AUC SS model was chosen for further assessment.

### *Model Validation*

Table 2 also shows that none of the selected terms were significantly correlated (italic numbers). A scatterplot and a Bland–Altman plot of this model using the mean coefficients from the bootstrapping is presented in Fig. 5. The model estimated the measured sweating with an $R^2 = 0.73$, $r = 0.85$ and rmse = 1.9 $\mu$L, which equals 12.9% of the measurement range. The limits of agreement were 27.6% above and 30.4% below the mean, which was close to zero due to the use of averaged coefficients. The reliability test of measured vs. estimated sweating gave an ICC of 0.84.

Testing of the model on the datasets from the last minutes of provoked sweating and of relaxation gave parallel (slope difference of 0.03) trend-lines with a bias of 0.28 $\mu$L. The lower $R^2$ for the relaxation dataset is due to a reduced range in sweating.

## DISCUSSION

The results from this study show that there was a high temporal correlation between Swr and SC, that inclusion of SP and SS parameters could explain some of the between-subjects variance, and that sweat production could be estimated with 12.9% average error by a model using a combination of the sum of SC increases, the SP and the SS areas under the curves.

The high temporal correlation between Swr and SC within subjects was not entirely expected in spite of the agreement with Muthny[21] and Edelberg.[9] Most previous studies have compared changes in evaporation to SC or SP during single sweat gland responses. A high correlation in this sense should lead to a similar correlation in this study assuming that both processes are equally linearly or non-linearly additive. Inspection of Fig. 2 suggests that this does not hold completely, as the SC tends to rise sharper and plateau earlier than the Swr. This could be due to a reduction in SC as the skin is hydrated, leading to swelling of the stratum corneum and sweat pore closure, an effect explained in Edelberg.[10] It is also known that evaporation follows the electrical response by roughly 5 s.[16,22] The last factor to consider is the responsiveness of the QSweat device used to measure Sw in this study, which may have provided slow recordings despite the backwards filtering correction and due to the necessary aggressive low-pass filtering.

The interindividual correlation between sweating and SC was not as high ($r = 0.49$), in agreement with both Edelberg[9] ($r < 0.50$) and Muthny[21] ($r = 0.23$–$0.64$),

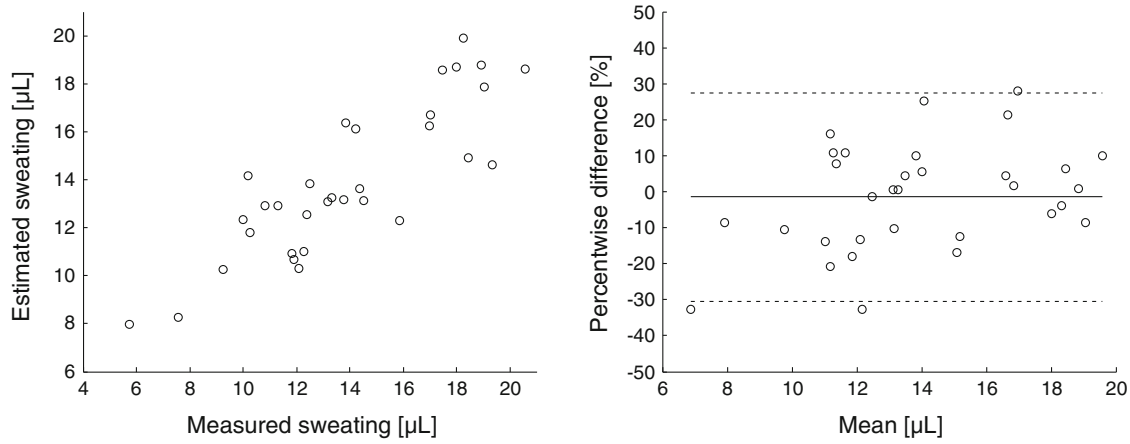


**FIGURE 5. Scatterplot of measured vs. estimated sweating based on the selected model (SCpos + AUC SP + AUC SS) with $R^2$ = 0.73, $r$ = 0.85, rmse = 1.9 µL (left) and Bland–Altman plot (right) showing the mean of the measured and estimated sweating vs. their percent-wise difference. The solid line represents the mean percentwise difference (−1.42%) while the dashed lines show the upper (27.6%) and lower (−30.4%) limits of agreement.**

although significant ($p < 0.01$). Stronger correlations were found between sweating and the sum of SC increases and the number of these responses (NSCR). In agreement with Ellaway *et al.*,[12] the positive SP responses correlated better with sweating than the negative SP responses.

Some indications of parameter redundancy could already be seen from Table 2 with the combinations AUC SC + NSCR, AUC SC + SCpos, NSCR + SCpos and SPpos + SCpos. Multilinear regression including all parameters produced an overfitting with $R^2 = 0.84$ which was reduced to 0.82 through elimination of the insignificant model parameters NSCR, SPpos and SPneg by means of stepwise regression. Validation of the suggested model from the stepwise regression by bootstrapping lead to elimination of AUC SC also. The remaining terms of SCpos + AUC SP + AUC SS were able to estimate total sweat production with a median $R^2 = 0.66$ and median rmse error of 2.28 µL, which represents 15.4% of the measurement range or 8.7% of the QSweat device range, from validation by bootstrapping. We believe that the magnitude of this error needs to be considered with respect to the intended use. This accuracy is nevertheless a significant improvement compared to conventional SC measurement (Bootstrap validation of a model based on only AUC SC gave $R^2 = 0.09$). The approximately ±30% limits of agreement between measured and estimated sweating (Fig. 5) is also difficult to evaluate due to lack of similar studies. As an example, within cardiac output measurements, a limit of agreement up to ±30% has been suggested to be acceptable.[6] This study may also be compared with Gagnon *et al.*,[13] where sweat gland activation was determined by a modified iodine-paper technique, in which a computerized method was compared with manual counts by the correlation coefficient and a Bland–Altman plot with the results of $r = 0.77$ and ±38 glands/cm$^2$ limits of agreement in a pool with a 70 glands/cm$^2$ mean.

The limits of agreement and the ICC can provide inconsistent results in agreement studies, and both are reported here according to the recommendation of Costa-Santos *et al.*[5] Similar to the Pearson's $r$, an ICC close to 1 indicated "excellent" reliability, but no agreement exists on the interpretation of the ICC. A value >0.9 has been proposed as high agreement between measurements, with an ICC between 0.7 and 0.8 representing questionable agreement in sports medicine.[1] Within psychometrics, ICCs >0.75 to >0.81 have been suggested as excellent agreement.[2,4,31] Portney and Watkins[25] suggested 0.75 as a threshold from "poor to moderate" to "good" reliability. Thus, we suggest that the ICC = 0.84 in this study represents "good" reliability between measured sweating and what could be estimated from model.

The same dataset was used to develop and to test the model in this study. Although the dataset was randomly divided in training and test sets in the bootstrap validation, all measurements were done by the same operators using the same methods and equipment on different subjects, hence this study does not assess repeatability or reproducibility. The ICC and the Bland–Altman plot, being based on an estimation from the mean model coefficients from the bootstrapping, is a best-case representation of the agreement.

Validation of the model during stress and relaxation periods gave a bias in the scatterplot (Fig. 6) intercept of 0.28 µL, indicating that the model slightly (approx. 10%) overestimates the Sw during heavy sweating. This could be due to a slightly nonlinear relation

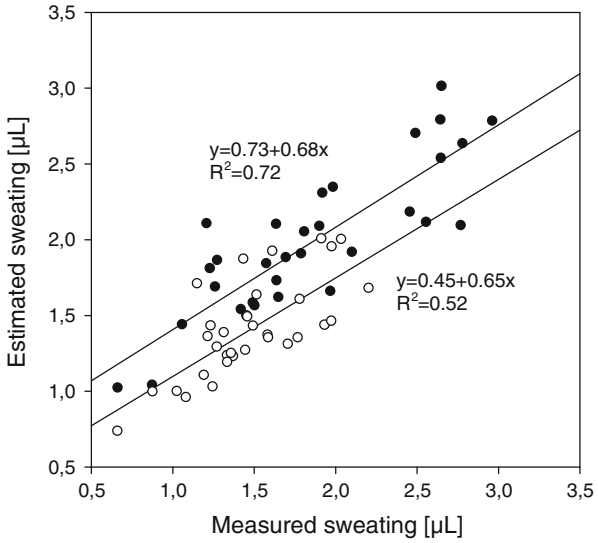


**FIGURE 6. Scatterplots of sweating estimated by the SCpos + AUC SP + AUC SS model for periods of provoked sweating (●) and relaxation (○).**

between the electrical and evaporative changes, and suggests that a more advanced model should be considered based on a larger range in Swr, which was limited in this study due to the measurement range of the QSweat® device. The slopes for sweating during stress and relaxation however, were parallel, indicating that the model's explanation of between-subjects variance is valid in both cases.

The SC alone had a strong correlation to the individual variations, but both the SS and SP significantly helped explain the between-subjects factors, also with positive signs of their linear model coefficients. As mentioned in the introduction, the SS is known to reflect the moisture content of the stratum corneum separately from the sweat duct filling, and the water uptake of the corneum is known to differ among individuals.[26] The SP contribution is more difficult to interpret, as the relation between SC and SP have previously been studied only to small extent due to lack of methods to measure them simultaneously. Contrary to the SC and SS, the palmar SP is negative with respect to an electrodermally inactive skin site and increases initially in the negative voltage direction when a sweat gland response occurs, but may shortly be followed by a response in the opposite direction towards 0 V. A positive SP model coefficient thus means that sweat production is estimated to increase as the SP approaches zero. In view of Fig. 2, SC and SP (with an opposite sign) seems to correlate well during the relaxation intervals, but as sweating increases, the SP reacts in both directions, giving a very different change in the area under curve than for the SC. Along with the low between-subjects correlation between the two parameters ($r = 0.02$), the results suggest that the two parameters bring complementary physiological information on the skin and sweating. The results in total speak in favor of utilizing simultaneous recording of SC, SP and SS for electrical measurement of sweat activity.

The results indicate that roughly 30% of the inter-individual variance in sweating could not be explained by the electrical parameters. The electrolyte concentration of sweat will directly influence the magnitude of the SC and SP responses, and relatively large variations in Na and K concentrations have been reported among healthy adults (51.9 ± 21.1 meq for males, 36.5 ± 18.7 meq for females, mean ± std.dev).[17] A sweat collector attached to the skin performing continuous conductivity measurement within a fixed volume could be included in order to increase the accuracy of the electrical measurement of sweating, and was considered in the planning of this study. Piloting however, revealed that insufficient amounts for analysis were collected during the experimental session when testing the Macroduct™ Sweat Collector. Another factor is the thickness of the corneum, which will influence both the Sw and electrical measurement, but in different ways. Egawa *et al.*[11] reported a std.dev of 37 $\mu$m from a 173 $\mu$m mean for palmar corneum thickness. Only palmar skin was measured in this study, which is different from non-glabrous skin with respect to innervation, sweat gland count,[20] corneum thickness,[11] and electrical properties.[7,29] Due to these differences, the model will not likely be valid on non-palmar skin without new calibration. A third factor could be inter-individual differences in the autonomous nervous system. However, a change in the sympathetic nervous system activity (SNA) will lead to a change in sweating, both in the way of evaporation and the electrical properties of skin, and in the same direction. In other words, the SNA is an underlying mechanism which governs both the reference measurement (evaporation) and the electrical properties used to estimate the reference measurement. Hence, when comparing these properties it was important to include variations in the SNA both inter- and intraindividually in order to cover most of the natural range. With the 40 subjects included in this study, there was a natural interindividual variation from healthy subjects, and the intraindividual variation was produced by the alternation of relaxation and mental stress periods during the experiment. The last factor to consider is the ambient humidity, which was not controlled in this study in order to allow natural variation. The ambient humidity variation could have naturally inflicted the overall evaporation and possibly the Sw measurement, but not the electrical measurements in the same way due to the occluding effect of the electrode. However, we believe that this difference

is small, as the QSweat probe confined the Sw measuring area with a small (approx. 5 $cm^3$) air-tight volume.

A future study is suggested including sweat electrolyte concentration measurement, measurements of larger ranges in Swr, and SC, SP and SS at additional skin sites with different corneum thickness, and repeated measurements over different seasons in order to assess the repeatability of the estimation.

In conclusion, estimation of sweating, defined as water loss from the skin, is significantly improved by the addition of SS and SP measurement to the SC measurement only, yielding a model for estimation of sweating with approximately 15% average error. We believe that the proposed method is useful for measurements of sweat production for both intra- and interindividual comparisons in situations where variations are large. Examples of such large variations include before-after treatment or treatment vs. control of hyperhidrosis, exercise physiology studies, nightsweats, or psychological studies involving moderate to strong stressors.

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