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A study on electrode gels for skin conductance measurements

Christian Tronstad[1], **Gorm Krogh Johnsen**[2], **Sverre Grimnes**[1,2] **and Ørjan G Martinsen**[1,2]

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

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Abstract

Low-frequency skin conductance is used within several clinical applications and is mainly sensitive to sweating and the moisture content of the stratum corneum, but also how electrodes introduce changes in the electrical properties. Four electrode gels were investigated with regard to sorption characteristics and electrical properties. Skin conductance time series were collected from 18 test subjects during relaxation, exercise and recovery, wearing different pairs of electrodes contralaterally on the hypothenar and the T9 dermatome. Pressure test was applied on the T9 electrodes. Impedance frequency sweeps were taken on the T9 electrodes the same day and the next, parameterized to the Cole model. ANOVA on the initial skin conductance level change, exercise response amplitude, recovery offset and pressure-induced changes revealed significant differences among gel types. The wetter gels caused a higher positive level change, a greater response amplitude, larger recovery offset and greater pressure-induced artifacts compared to the solid gels. Sweating on the T9 site led to negative skin conductance responses for the wetter gels. Correlations were found between the desorption measurements and the initial skin conductance level change (hypothenar: $R = 0.988$ T9: $R = 0.901$) RM-ANOVA on the Cole parameters revealed a significant decrease in R_s of the most resistive gel. Clinical implications are discussed.

Keywords: sweat, skin conductance, electrodes, water sorption, stratum corneum, electrodermal activity, sweat glands

1. Introduction

The electrical admittance of the skin is dependent on several factors. The epidermal stratum corneum (SC) layer, consisting of mainly dead skin cells, gives a large impedance in series with

the viable skin, dominating the measurement at low frequencies (below 10 kHz) (Martinsen *et al* 1999). The low-frequency skin admittance measurement then becomes highly sensitive to mechanisms occurring in this layer, mainly the ionic shunting provided by the sweat ducts as they are filled with sweat and the absorption of water in the SC. The admittance contribution from the sweat duct is predominantly conductive (Martinsen *et al* 1998), while the SC moisture content also significantly influences the low-frequency capacitive part of the admittance (Martinsen *et al* 1998, Martinsen and Grimnes 2001). The high sensitivity of the skin conductance to sweat activity has opened up several clinical applications.

Within the field of psychophysiology, skin conductance is best known as a measure of electrodermal activity (EDA) (Boucsein 1992), and due to the link between palmoplantar sweating and the sympathetic nervous system, this parameter has found its use as a tool for stress assessment (Healey and Picard 2005, Setz *et al* 2010), assessment of psychiatric disorders (Iacono *et al* 1999, Nilsson *et al* 2006) among other uses, often in combination with other physiological parameters. Within anesthesia, it has been proposed as a tool in pain assessment (Ledowski *et al* 2007, Hullett *et al* 2009). Skin conductance at other skin sites has been proposed for the diagnosis and treatment evaluation of hyperhidrosis (Tronstad *et al* 2008), and may be a promising tool for diabetes (Hoeldtke *et al* 2001), cystic fibrosis (Quinton 2007) and various neurological disorders.

It is known that the choice of electrodes for electrical measurements will heavily influence the signals that are measured (McAdams *et al* 1996, Mayotte *et al* 1994 and Rahal *et al* 2009), and so it is important to find the best-suited electrode for the intended use. Depending on how the measurements are interpreted, artifacts introduced by the electrodes could lead to incorrect diagnostic assessments.

Studies of electrode properties have previously been carried out in different ways (Rahal *et al* 2009, Eggins 1993), but not with focus on monopolar skin conductance measurements combined with an *in vitro* sorption study of the water holding properties of the electrode gels. In addition, the authors are not aware of any studies where parameters from skin conductance time series have been compared between different electrode gel types recorded simultaneously on the same test subjects.

In this study we investigated four different types of electrode gels for skin conductance measurements. As a low-frequency monopolar setup will focus mainly on the SC, its general hydration state, and sweat filling in particular, will be very dominant for what is being measured. SC hydration may be affected by occlusive effects, and also through a direct and sometimes rapid hydration from the electrode gel. We compared the amount of free water in the electrode gels *in vitro* with the results from electrical *in vivo* measurements with the aim of finding how different electrode types will affect the skin conductance recordings. In particular, we wanted to investigate the stability of the skin conductance level, the response to sweating stimuli, the post-stimulus recovery and the susceptibility to mechanical artifacts. As both SC properties and innervation differ according to skin site, especially on palmoplantar skin compared to non-palmoplantar skin (Kuno 1956), we included the T9 dermatome on the abdomen in addition to the hypothenar site on the palm for this study.

2. Materials and methods

2.1. Electrode types

Four electrode types were investigated, all with similar Ag/AgCl metallic part, but with different composition of the gel. The *a priori* gel properties of the different electrodes are listed in table 1.

Table 1. Overview of the electrode gel types examined in this study according to gel material, the intended use, effective electrode area (EEA) and type of adhesion.

Type	Gel material	Intended use	EEA (cm^2)	Adhesion
A	Solid hydrogel	ECG, neonatal	5.05	Adhesive gel
B	Wel gel	ECG, short term	2.54	Outer adhesive part
C	Isotonic EDA jelly	EDA	1.95	Outer adhesive part
D	Karaya solid gel	ECG, hypoallergenic	6.16	Adhesive gel

The brands of the electrode types were as follows.

- Type A: Kendall™ KittyCat™ 1050NPSM
- Type B: Ambu® Blue Sensor Q-00-A
- Type C: discount disposables, model TD-246 skin resistance—skin conductance electrode paste
- Type D: Kendall™ Arbo™ H83V.

2.2. In vitro sorption characteristics

The sorption characteristics of the four different electrode gels were obtained using a gravimetric method where water loss and water gain could be monitored carefully. A dynamic vapor sorption instrument (DVS) from the Surface Measurements Systems (London, UK) was used for this purpose. This instrument enables very sensitive gravimetrical measurements that can be performed under controlled environmental conditions, ensuring high-precision and reproducible results. The mass resolution of the instrument is 0.1 μg, and the gel samples in our study had a mass in the range of 4–10 mg at the measurement onset. This means that the sample masses during the entire sorption cycle were well above the resolution of the instrument, resulting in smooth time series. The gel samples were loaded into a metal pan in a closed chamber where temperature and relative humidity (RH) were pre-programmed and carefully monitored. The two solid gel samples, A and D, were prepared by means of sterile scalpels to be of approximately the same size and geometrical shape (cubics). The wet gel B was scratched off the attached surface of the electrode and placed in the metal pan as a small droplet.

Finally, the cream-based gel C was loaded into the pan by gently squeezing the bottle containing the cream. The samples were measured one at a time, and all four measurements were initiated immediately after sample preparation was finished. The RH was pre-programmed to increase in steps of 10%, from 0% up to 90%, and then back to 0%. The temperature was kept stable during the entire measurement at 25 °C. The mass reading found at 0% RH was taken to be the dry mass of the sample. A step in RH was initiated when the mass rate, dm/dt, was less than 20 ppm for a period of at least 10 min, ensuring the samples to reach their equilibrium state at every level of RH.

The characteristic time constants (τ), which yield the time needed for the sample mass to step approximately 63% of the entire increase needed to reach its stable state value, were obtained by fitting the time courses for each increment in RH to $y = a\exp\left(\frac{t}{\tau}\right) + c$.

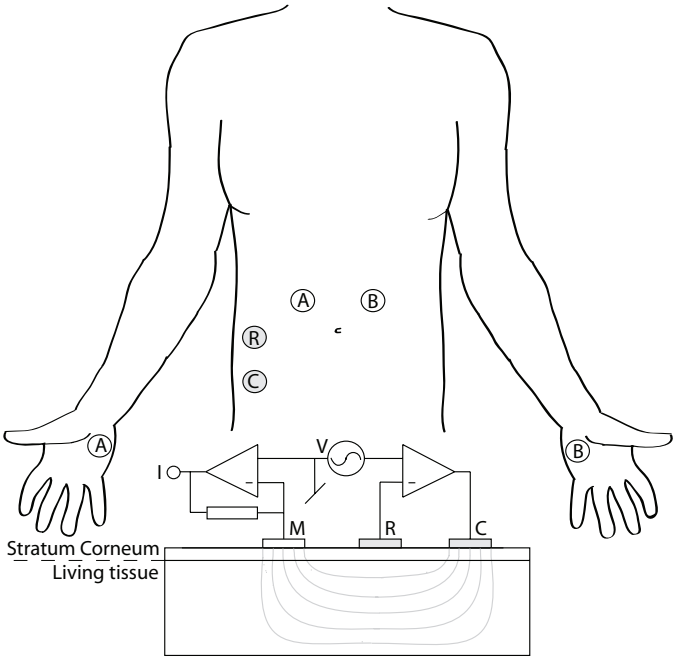

Figure 1. Measuring principle, electrode configuration and electrode placement for this study.

2.3. In vitro electrical characteristics

Electrical impedance frequency sweeps were taken on fresh pairs of each electrode type attached front to front using a Solartron 1260+1294 setup (Solartron Analytical, UK), with a constant voltage excitation of 30 mV RMS and a frequency range from 1 Hz to 100 kHz.

2.4. Skin conductance measurement

The three-electrode system, shown in figure 1 (Grimnes 1983b and Grimnes *et al* 2009), was used to obtain unipolar measurements of the skin surface conductance density (SSCD) of the area below the EEA of the measuring electrode. The current at the current injecting electrode *C* is regulated by remote currentless sensing at the reference electrode *R*, providing a potential below the *R* electrode equal to *V*, thus eliminating the contribution from the *C* electrode and the underlying SC to the measurement. Due to the vastly higher impedance of the SC compared to living tissue, the measurement becomes exclusively sensitive to the admittance of the SC area below the EEA of the measuring electrode *M*, which is proportional to the voltage at *I*. This principle is extendable to simultaneous measurements at multiple sites, and the electrode placement used in this study is shown in figure 1 where two electrode types A and B are compared (corresponding to *M* in the circuit). AC measurement was used to avoid polarizing the skin or electrodes and to avoid contribution from endogenous potentials in the skin. For single-frequency SSCD time series recording in four channels, the Sudologger (Tronstad *et al* 2008) was used. Impedance frequency sweeps from 1 Hz to 100 kHz were taken using the Solartron 1260+1294 equipment. Constant voltage excitation of 30 mV RMS magnitude was used for both equipments to keep the current density low and the measurement within the linear range.

2.5. Test subjects and protocol

Eighteen (14 males, 4 females) healthy volunteers who gave informed consent were each assigned to two of the four electrode types for contralateral comparison of the measured SSCD parameters. Initially, the test subject (TS) was asked to sit down on a chair while measuring electrodes were placed simultaneously on the left and right hypothenar sites of the palms and about 4 cm to the left and right of a point slightly above the navel on the T9 dermatome. The measuring electrodes on each side of the body were of the same type, and the selection was based on a matrix of all six possible pairwise combinations of the four types.

After electrode fixation, the SC was recorded for 10 min, while the TS was sitting still on the chair. During the last four minutes, the TS was asked to remain quiet and focus on relaxing in order to obtain a baseline. At the tenth minute, a loud sound (2.5 s sound sample of a glass breaking played by a set of speakers at an intensity measured to be 82 dB 1 m in front) was played to induce a startle response, and the TS was asked to stand up and perform squats for 2 min to induce thermal sweating. This was followed by 15 min of sitting relaxation until a test for pressure artifacts on the abdomen electrodes was performed by an operator. The TS was asked to relax the diaphragm while the operator pushed the electrodes equally and simultaneously inward on the abdomen for 10 s, pushing as far as to the point where the tissue would no longer compress easily. This was followed by 2 min of recording to check whether any SSCD offset would remain or recover to the previous level.

The palmar electrodes were then removed and electrical impedance frequency sweeps were taken on both of the abdomen electrodes. The TS was then wearing these electrodes until returning the next day, approximately 24 h later, for a control measurement. For the prewired electrodes, the wire was taped to the skin in order to avoid pull on the electrode. The TS was instructed to pursue normal activities and not to give the electrodes any consideration except for not showering the abdomen area. For all the electrodes which were still attached on the next day, new impedance sweeps were taken both with the electrodes as they were and after a refixation of the electrodes by gentle pressure on the adhesive area. After gentle removal of the electrodes, photographs of the skin areas were taken. The pictures were evaluated by a dermatologist for signs of skin irritation caused by the electrode gel.

2.6. Data analysis

Sudologger measurements were scaled to a unit of μS cm^{-2} (SSCD) for comparison between electrodes of different effective electrode area (EEA). For the SSCD time series, the following parameters were extracted.

- Level change during the initial 10 min after electrode–skin contact.
- Amplitude of the SSCD change during the 2 min exercise interval, calculated as the maximum deviation from the prestimulus baseline.
- Offset between the prestimulus baseline and the level reached after 15 min of recovery following the exercise interval. In some cases there were spontaneous responses during the end of the recovery, and the offset was then calculated as the minimum value within the last 2 min of recovery subtracted by the prestimulus baseline.
- Change induced by the pressure test on the electrodes, calculated as the maximum deviation from the prestimulus baseline during an interval of the applied pressure time + 5 s.
- Offset between the level 2 min after the pressure test and the level before the applied pressure.

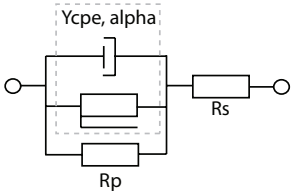

Figure 2. The Cole system electrical equivalent model with the components R_s, a resistor in series with a parallel combination of the resistor R_p and a constant phase element (CPE). The CPE is characterized by its admittance Y_{cpe} and its alpha(α) exponent ($Y_{cpe} = G_1(\omega\tau)^{\alpha}$).

Separate but equal statistical analyses were done for the hypothenar and abdomen measurements. ANOVA was done on all the parameters. Test for normality was done using the Shapiro–Wilk test and the Kruskal–Wallis ANOVA on ranks was used when the test for normality test or equal variance did not pass. For pairwise multiple comparison, the Holm–Sidak *t*-test was used when the normality test passed and the Student–Neuman–Keuls test was applied otherwise. For cases of missing values, the Dunn test was used instead.

For the Solartron impedance frequency sweeps, the measurements were parameterized using a Cole system electrical equivalent model (Grimnes and Martinsen 2005) shown in figure 2. Using the curve-fitting tool in the ZView 3.0a software (Scribner Associates Inc.), the Cole-parameters R_p, R_s, Y_{cpe} and alpha were determined with a median accuracy of 1.77% fit error. In a few cases with poor electrode-skin contact on day 2, the impedance measurement was inaccurate leading to a poor fit. The parameters with an error larger than 10% were excluded from the analysis. Repeated measures ANOVA were done on each of the Cole parameters, grouped into the day 1 measurements and both the day 2 measurements. The same tests as for the time-series parameters were applied for pairwise multiple comparison.

For comparison between the wetness of the gels and the skin conductance measurements, Pearson product moment correlation coefficients and *p*-values were calculated between the initial relative mass change of the gels in the DVS at 0% RH and the median initial level change of the SSCD time series.

3. Results

3.1. In vitro results

Figure 3 shows the relative water uptake of the four different gels as the RH was altered in steps from 0% up to 90%. The mass value at 0% is taken as the dry mass of the gel, meaning that the initial values at the far left in the figure represent the relative amount of water in the gel as it will be before any electrode to skin attachment in the electrical measurements. The wet gel in our study, the type B gel, contained the highest amount of water initially, almost doubled compared to the cream-based C gel, whereas types A and D were very similar, with little amounts of free water to be depleted into the environments. The type B and C gels did not regain their initial water content, even at 90% RH, although B took up an amount of water corresponding to several times its own dry weight. The A and D gels, on the other hand, more than doubled their weight during the absorption process.

There were few signs of hysteresis in our measurements, which would correspond to a gap between absorption and desorption steady-state mass values at a given value of RH. Only for the B gel there was hysteresis in a range where the RH value was between 60% and 80%,

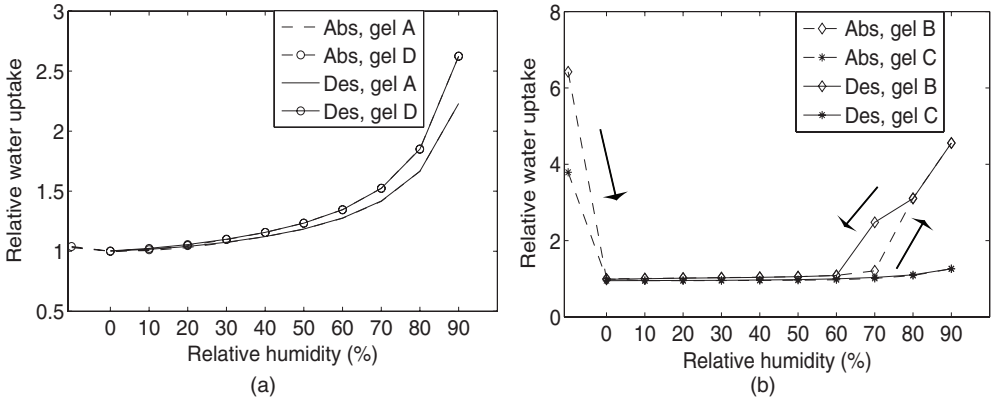

Figure 3. Relative water uptake of gels A and D in (a) and B and C in (b). The arrows, in the figure, to the right indicate the direction of the sorption process, both initially after sample placement *n* the DVS chamber and in the hysteresis. Absorption and desorption curves are overlapping in (a).

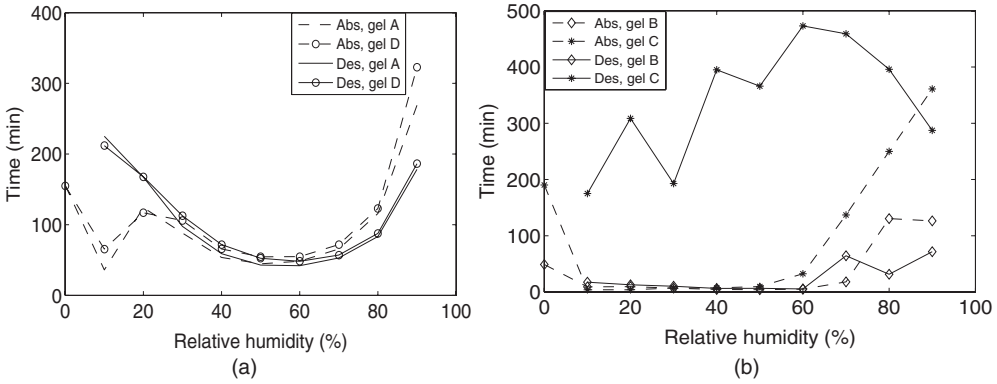

Figure 4. Time constants for gels A and D in (a) and B and C in (b). The points to the far left in the graphs illustrate the characteristic time for the gels to reach equilibrium with the dry chamber, starting from their initial hydration state.

where the gel retained more of its water content during desorption compared to absorption. The time courses following a step in the RH were found to closely follow an exponential curve which is in accordance with Fick's law of diffusion.

The characteristic time constants for water entering and leaving the gels are shown in figure 4. The values to the far left in the graphs are the typical values for how rapid the gels, starting from their initial (manufactured) hydration state, will reach equilibrium with the dry air in the DVS chamber at 0% RH. The smaller this value was, the faster the gel deposited water into its environment. Again, the A and D gels showed very similar patterns, both during absorption and desorption. The B gel gained and released water very rapidly for almost all RH values, whereas the cream-based C gel gained water with a markedly decreasing rate as the RH was increased. During desorption the C gel needed long time to reach its equilibrium state for each decrement in RH.

Table 2. Gel ac conductivity (μS cm^{-1}) at low and high frequencies for all the gel types presented as median (bold), 25% and 75% quartiles.

Frequency	A	B	C	D
1 Hz	**166.0** (162.8, 174.4)	**313.8** (301.3, 320.6)	**282.0** (265.7, 290.5)	**3.600** (3.299, 3.676)
100 kHz	**399.1** (390.3, 407.4)	**1480** (1445, 1523)	**670.6** (597.9, 737.1)	**71.81** (70.58, 73.27)

The *in vitro* electrical measurements of the gels revealed a higher ac resistance of the gel-type D than the others, represented in table 2 at 1 Hz and 100 kHz for the calculated conductivity. These measurements are sensitive to the dispersions from both the metal-gel polarization impedance and the gel, and would not parameterize with a good fit according to a single Cole system.

3.2. In vivo results

The ANOVA revealed significant differences among the electrode types for all the time-series parameters. Figures 5(a) and (b) show that on both the hypothenar and T9 sites, the electrode of type B had the highest change in SSCD after electrode fixation followed by electrode C. Types A and D had lower and roughly equal medians, but with a slightly lower interquartile range (IQR) for electrode A. Figure 5(c) reveals that on the hypothenar site, the electrode of type B gave a significantly higher response amplitude than the other types. This is also true for the T9 site; however, the response of the type B gel is negative in this case. The ability to recover toward the baseline after a period of sweating is presented in figures 5(e) and (f). Again, we see the highest offset for type B on both sites, while type A has the lowest median/mean offset and IQR/variance. A difference between the sites is seen on the type C offset, with clearly lower ability to recover on the T9 site than on the hypothenar site.

Correlations were found between the initial change in SSCD (figures 3(a) and (b)) and the initial relative mass change in the DVS at 0% RH with an $R = 0.988$, $p < 0.05$ for the hypothenar and $R = 0.901$, $p < 0.1$ for the T9 site.

Significant differences were also found in the susceptibility of the electrode types to pressure artifacts. As figures 6(a) and (b) show, types B and C were more affected by the pressure test than types A and D. Again, type A scored the lowest offset and IQR with type D not far behind.

The differences between the SSCD measurements from different electrode types were most clearly seen on the recordings using certain pairs of electrode types simultaneously on the same subject. Typical examples of these findings are presented in figure 7. The tendency of types B and C to initially increase the SSCD was absent for types A and D in the same subjects. Figure 7(b) displays how type B can feature a *negative* response to sweating on the T9 site, while type A at the same time expectedly gives a positive response on the same site. In figures 7(a) and (c), the most visible difference between the electrode types on the hypothenar is in the long-term changes of the SSCD level where type A seems to be more stable than types B and D. However, there were also some differences in the spontaneous SSCD responses recorded, where a few responses could be seen in type B which were absent or vanishingly small in type A. Figures 7(a) and (d) show the tendency of either type B or C to remain elevated after the sweating compared to type A or D. An intraindividual comparison of types B and C is shown in figure 8, exhibiting a special case on the T9 site where types B and C initially drift apart, converge and overlap during the exercise interval and proceed to drift apart again.

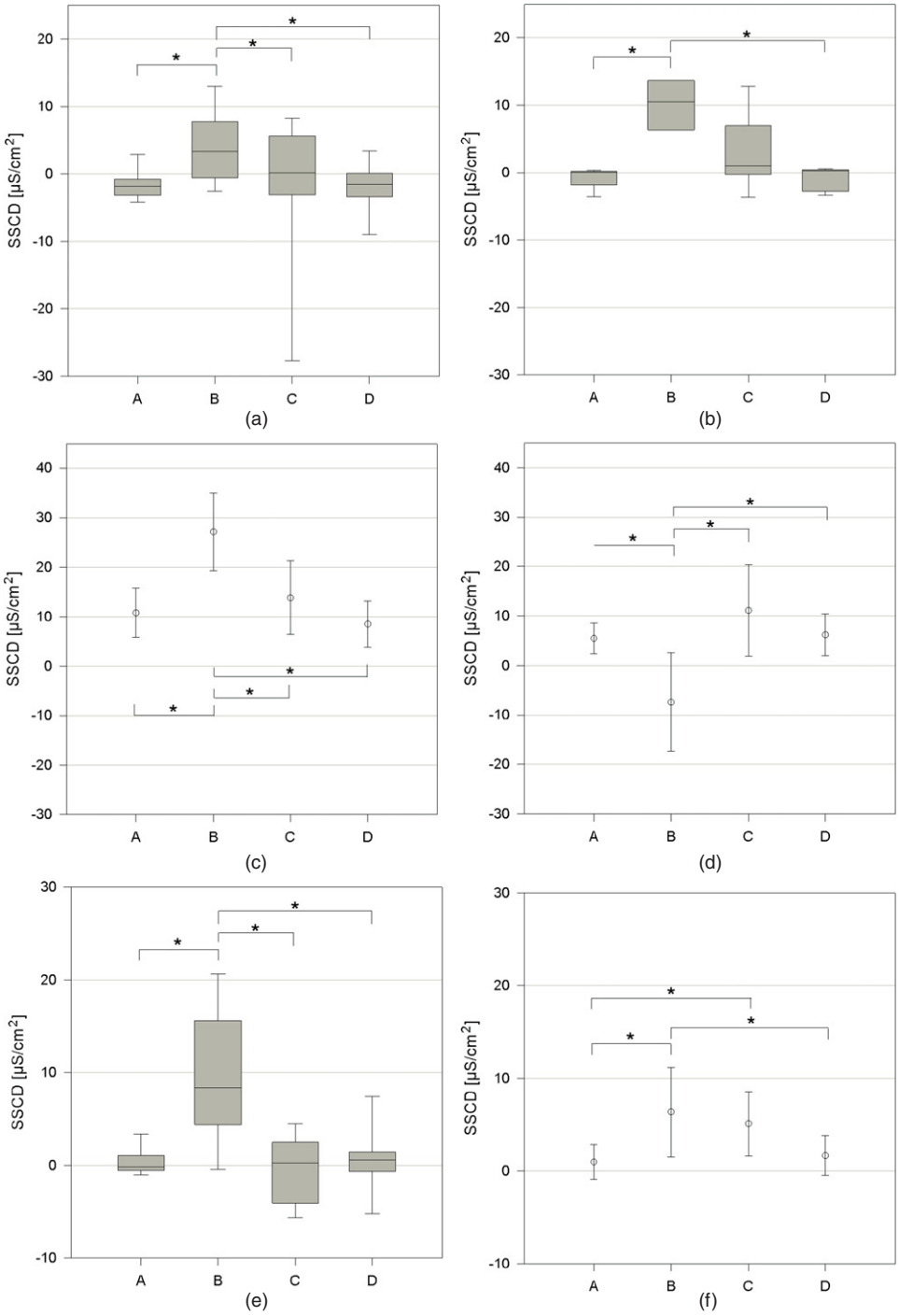

Figure 5. Time-series parameters from the hypothenar (left) and the abdomen (right): (a) and (b) initial difference in SSCD after 10 min of relaxation; (c) and (d) maximum change from the baseline during sweating stimulus; (e) and (f) offset from the baseline at 15 min after the end of sweating stimulus. $^*p < 0.05$.

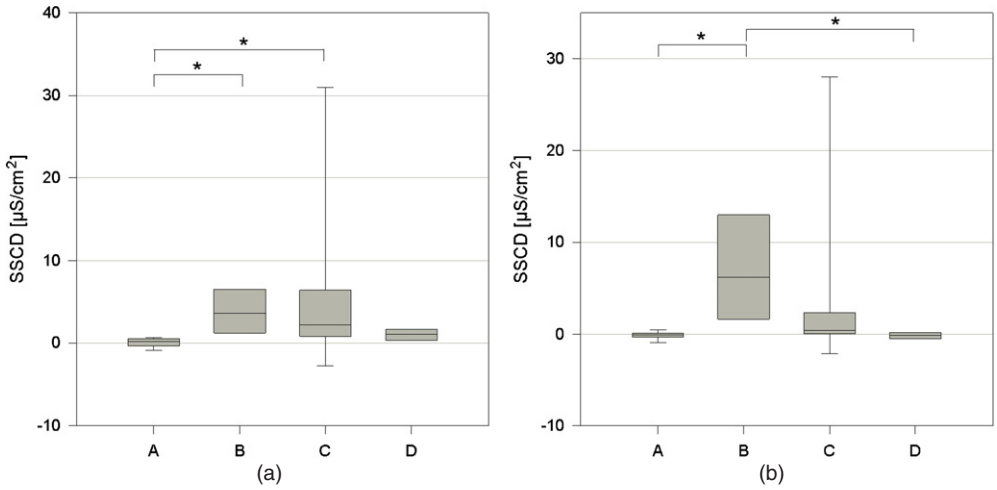

Figure 6. Quantization of the pressure-induced artifact on the electrodes. (a) Maximum deviation from the baseline during pressure time + 5 s. (b) Offset from the baseline 2 min after the end of applied pressure. $^{*}p < 0.05$

For the Cole-parameters from the impedance frequency sweeps, the ANOVA revealed no significant differences between days 1 and 2 except for the following three cases.

- The Y_{cpe} for electrode A was significantly higher on day 1 than on day 2, also after refixation of the electrode with a difference in means of 77 nS.
- The R_{s} for electrode D was significantly higher on day 1 than on day 2, also after refixation of the electrode with a difference in means of 212 Ω.
- The alpha parameter for electrode D was significantly lower on day 1 than on day 2, also after refixation of the electrode with a difference in means of 0.072.

The dermatologist assessment of the pictures revealed no significant signs of skin irritation. There were some signs of very faint erythema, but its degree was always lower than on the areas which had been taped.

4. Discussion

In this paper we have shown differences in both the *in vitro* properties and the *in vivo* electrical measurements of the gels, which are discussed below.

4.1. In vitro evaporation

Figure 3 shows that the gels we tested in our study contained very different amounts of water free to evaporate, and this means that they also likely will hydrate the SC very differently after the electrode onset. However, the amount of water released from the gels initially in the *in vitro* sorption measurements is not necessarily the same as the amount of water entering the SC from the gel after the electrode onset. The transport of water to or from the gels is in general determined by the difference in partial pressures of water between the gel and the material it is in contact with. The water in the SC will in general have a different partial pressure than the air initially in the DVS measuring chamber. Also, other transport mechanisms such as

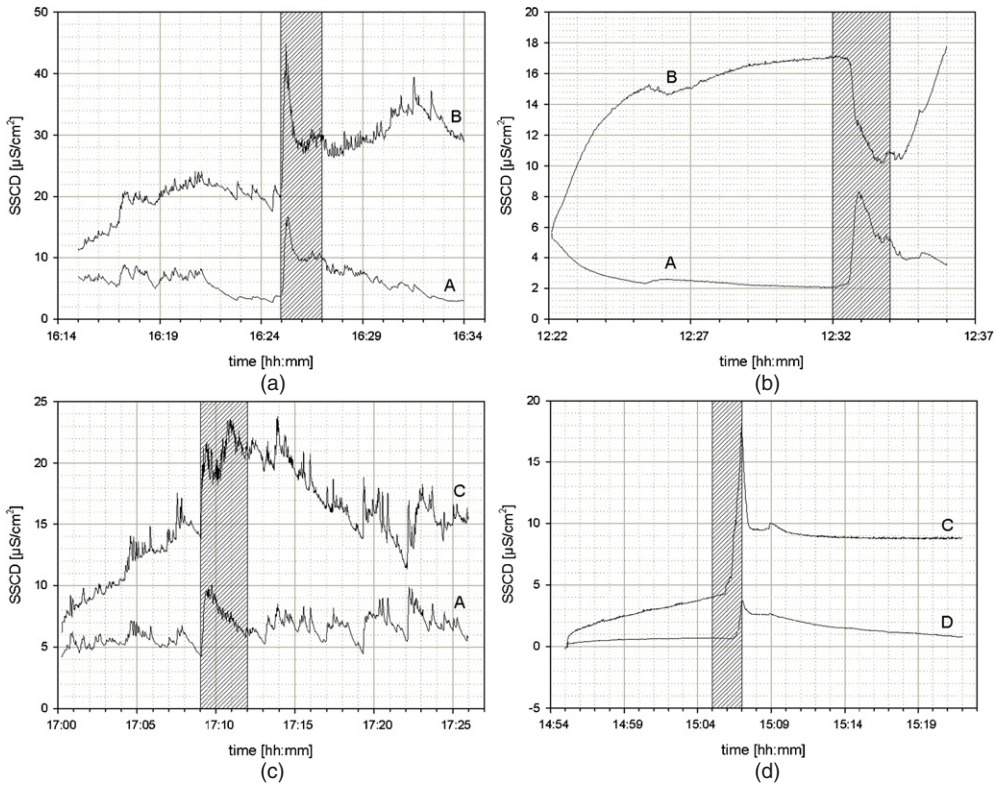

Figure 7. Examples of features found with simultaneous recordings using different electrode types on the same subjects: (a) hypothenar recording using types A and B; (b) T9 recording using types A and B; (c) hypothenar recording using types A and C, and (d) T9 recording using types C and D. The hatched area indicates the 2 min exercise interval.

skin occlusion and osmotic transport of water to or from the electrode gel, due to differences in electrolytic concentrations between the SC and the gel, can be expected to alter the skin conductance. The type B and C gels contained large amounts of water, whereas A and D barely lost any water to the dry environments at 0% RH in the DVS measuring chamber. This is correlated with the results from the skin conductance measurements, especially in how the SSCD changes initially after the application of the electrode. This indicates that the moisture content and viscosity of the electrode gel has a major impact on the quality of the skin conductance recordings. A too wet gel such as type B will bring moisture to the SC or penetrate sweat ducts if the viscosity is low enough, introducing errors in the skin conductance measurement. For electrodes A and D the lack of initial free water to hydrate the SC opens for the possibility that the SC was depleted of moisture for a short period of time after the electrode onset, depending on the ionic concentration of the hydrogel, by means of osmotic transport of water into the gel. It is seen from figures 5(a) and (b) that skin conductance actually decreased during the first 10 min of relaxation. However, a slight reduction in the skin conductance level due to a decrease in sweat activity is to be expected, as the test subject is sitting down and relaxing. Also, the relatively cold gel may cause contractions in skin pores, resulting in a small decrement in sensed conductance values (Eggins 1993). Finally, the sorption results showed that the type A and D gels had very similar *in vitro* patterns, both regarding the relative water

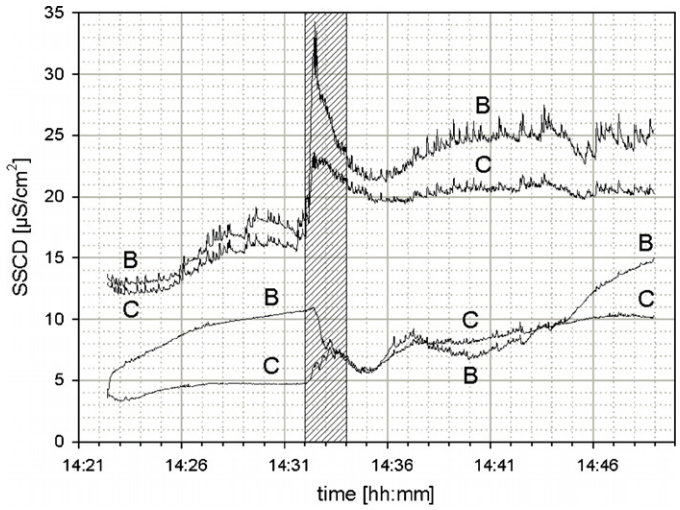

Figure 8. Example measurement from one test subject wearing electrode types B and C on the hypothenar (upper plots) and the T9 site. The hatched area indicates the 2 min exercise interval.

uptake and how rapid water entered or left the gel samples, supporting our findings from the skin conductance time series.

4.2. In vitro sorption time constants

Both the amount of water free to evaporate from the gel and how rapid such a diffusion process takes place can be expected to influence the skin conductance values. From figure 4 we see that the type B gel deposited its free water more rapidly into the dry DVS chamber than the other gels. A corresponding effect was seen in our electrical time-series measurements, e.g. figure 7(b). The type C gel, although containing high amounts of free water initially, gave lower initial skin conductance level change (figures 5(a), (b)) than gel B. This effect is probably a combination of less free water, but also a larger time constant initially for this water to evaporate from the gel into the dry chamber air. Depending on the initial hydration state of the SC, a similar trend can therefore be expected to be seen also when water diffuses from the gel into the SC. The evaporation of water from a sample, and hence the time constant, is depending on the geometry and size of the sample (Crank and Park 1968). However, as the four samples were of relatively equal size and mass, the patterns in time constants seen in figure 4 can be expected to give a representative measure of the hydration characteristics of the gels. Although gels can be expected to hydrate the SC giving increased conductance levels after the onset, the water transport may potentially also go from the SC to the gel, especially during a sweat response when the moisture content locally in and around the sweat ducts increase significantly. In that sense the gel with the lowest absorption time constant (gel B) will pick up this water most rapidly, whereas gel C will not be able to gain much water. Such effects are probably slow, and only significant in long time monitoring.

4.3. Inverse sweat response

As shown in figure 5(c), all electrode types gave increased SSCD values at the hypothenar sites during the sweating stimulus, and electrode type B, the wettest gel, gave the highest response. However, for the T9 site, electrode B gave a significant *decrease* in SSCD during the sweat

response initiated by the squats. It is our belief that this inversed sweat response is due to the effect of electrode gel having penetrated down into the sweat duct. As the conductivity of the wet gel of the type B electrode is much higher (roughly 300 μS cm^{-1}) than the sweat electrolyte solution (5.56 μS cm^{-1}, Licht *et al* 1957), a penetration of gel down into the ducts, that are empty or partly filled (Grimnes 1983a), will result in an increased conductivity. On the other hand, during a sweat response, the higher conducting gel will be substituted with a lower conducting sweat liquid causing a decrease in the measured skin conductance (figures 7(b) and 8). Figure 8 shows a special case where the type B and C SSCD initially drift apart, but lie on top of each other for a while during and after the exercise interval. Within the overlapping period, there are equal proportions of sweat and gel in the ducts below both electrodes, yielding equal SSCD. The question remains why the inverse sweat response occurred on the T9 site and not on the hypothenar site. The dependent factor may be the thickness of the SC being several times thicker on the palmar skin than the abdominal skin (Montagna and Parakkal 1974, Ya-Xian *et al* 1999). Comparing figures 5(a) and (b), we see that the initial SSCD change on the T9 site was more than three times as large as the change on the hypothenar. This points to the electrical shunting effect caused by gel penetrating the sweat ducts being dependent on the SC thickness and thereby the sweat duct length within the SC. It is likely that more complete shunts will occur in thinner SC than in thicker SC where part shunting will contribute a lot less to the measurement, due to the main restricting factor to the skin conductance being the SC. Thus, a larger proportion of ducts completely shunted by the gel at the T9 site may explain why sweating induces a negative response on this site contrary to the hypothenar where the thicker SC yields a larger proportion of partly shunted ducts.

The part shunting of the hypothenar ducts may also explain the higher SSCD amplitude of type B (figure 3(c)) by a sensitivity to a greater number of sweat glands where the sweat is not pumped all the way to the orifice of the duct, which also explains why some responses are present for this type but absent for a solid hydrogel type (figure 5(a)). This could not be due to differences in EEA as type B had the smallest EEA of all the gels.

4.4. Post-sweating recovery

After the end of the exercise epoch, the SSCD is expected to return to a level close to the baseline within a few minutes as the sweat activity diminishes. As seen in figures 5(e) and (f), there is still a large positive offset after 15 min of relaxation for electrode type B on both sites, and type C on the T9 site. We believe that this can also be attributed to the penetration of gel into the sweat ducts along with the sweat reabsorption process in the ducts or glands. As sweat is reabsorbed in the glands following the sweating period, gel is again allowed to enter the ducts, perhaps even actively transported through the ductal reabsorption process. This may explain why type B exhibits a positive offset even though the response during the exercise was in the negative direction. Another plausible explanation for the recovery offsets, also attributed to the solid gels, is the absorption of sweat in the SC following the sweat emergence, leading to a change in the total conductance. A third plausibility, concerning only the wetter gels, is that there may have been cases of a constant wetting of the SC during the whole recording period, leading to a somewhat steady increase in level as with type B in figures 7(a) and 8.

These hypotheses are also supported by the lower variance of the solid gel types in all the results presented in figure 7, suggesting that the artifacts introduced by the wet gels depend on the intersubjective properties of the SC such as its thickness.

4.5. Pressure artifacts

The pressure artifacts showed the same pairwise pattern as the sorption results with electrodes A and D to be very similar and relatively unaffected, but with large effects on electrodes B and C. The two latter electrode types have a gel that is believed to have a much less mechanical stability and are thus sensitive to applied pressures, smearing the gel giving both better contact and possibly also an increased EEA. Due to its low viscosity gel B may actively be pushed into the sweat ducts, which in that case would explain the significantly elevated level in skin conductance also after the final relaxation period.

4.6. Changes in Cole-parameters from day 1 to day 2

Although the repeated measures ANOVA resulted in few significant changes in the Cole parameters from day 1 to day 2, the test may have suffered from unapparent adhesion changes of the EEAs. Even though the electrodes were gently re-fixed for a second measurement on day 2, persistent poor adhesion could alter the impedance measurements. This, along with a lowering of the number of samples due to electrodes having fallen off, may have weakened the ability of the test to find significant changes in skin impedance changes caused by the electrodes. From day 1 to the control on day 2, one of type A, two of type B, and two of type D electrodes had fallen off the skin of the TS. This should not be interpreted as a result on adhesion properties, as the electrodes are constructed differently in this regard. In the three electrodes of type C, the gel chamber was completely void of gel, evident after removal of the electrode and corresponding to very high impedance values which were not included in the Cole-parameter assessment.

We believe that the significant change found with type D, evident in both the R_s and the alpha parameter, could be attributed to a skin reaction occurring below the SC, but can also be explained by a lowering of the gel series resistance, as type D had a relatively high front to front resistance compared to the other gels (table 2). The latter case is also supported by the dermatologist assessment of skin reaction, where no sign of any reaction beyond very faint erythema was found. Thus, it is likely that this gel may cause a drift in long-term skin conductance measurements because of the initial low-frequency resistance of type D which is relatively high compared to the SC resistance during sweating. This high gel resistance may also explain the slightly lower SSCD response on the hypothenar for this type compared to the other solid gel type A (figure 3(c)). Another effect that may have been responsible for the changes in the type D gel is its ability to gain an amount of water corresponding to about 2.5 times its own dry weigh under humid conditions, as seen in figure 3. During a 24 h electrode placement, water is building up underneath, giving a much increased water content locally beneath the electrode. Some of this water may potentially have been taken up by the gel, which surely would have resulted in a much reduced series resistance. However as such a water uptake typically has a characteristic time constant of a few hours, this effect is not seen initially, but will be present in long-time measurements.

The three (33%) cases of empty gel chambers on day 2 for type C may be due to absorption of all its moisture in the skin corresponding to its low *in vitro* ability to reuptake moisture.

The decrease in Y_{cpe} from day 1 to day 2 may have been due to a net reduction in the EEA due to only part electrode contact with the SC. Thus, a reduced number of ducts in contact with the gel would result in a net reduced low-frequency admittance as a reduction of electrode area would probably also yield less water build-up in the SC due to occlusion.

4.7. Clinical implications

The inverse sweat response (figures 5(d) and 7(b)) can obviously lead to misinterpretation in sweating-related measurements. Even though this effect did not occur on the hypothenar site in this study, we cannot exclude the possibility that it could occur on this site as well. Type B recordings were often seen to have a steady increase in the SSCD level throughout our 29 min recording (e.g. figure 7(a)), and it is not unlikely that inverse sweat responses could occur also on this site after a longer time of electrode-skin contact when the gel has been allowed to penetrate deeper into the ducts.

The implications of these electrode-related phenomena depend on how the measured signal is interpreted or computationally parameterized. Traditionally, both the skin conductance level and the characteristics of the skin conductance response have been used as psychophysiological parameters (Boucsein 1992) and the results in this paper point to the potential sources of errors introduced by the electrodes regarding both features. The frequency of the skin conductance responses is a parameter which may not be as susceptible to these errors as the abovementioned. However, the response detection usually depends on an amplitude criterion beyond a certain threshold, which may cause electrode gel dependence in this parameter as well due to differences in amplitudes. In addition, these results indicate that the gel to duct penetration may alter the sensitivity to the sweat glands below the EEA. In other words, a moist gel with low viscosity could detect a greater number of responses than a solid gel or gel-free electrode. For the wetter gels, the results will likely to have a significant dependence on the time since electrode attachment and the SC thickness.

The pressure artifacts are more relevant in situations where the patient is moving or physically active during the recordings than for controlled setups in the laboratory. Pressure on the electrodes is also likely to occur during sleep monitoring as the test subject is turning around. Push or pull on the electrodes may give transient artifacts which could be detected as single responses, but the greatest source of error comes from the electrodes which cause changes that remain after the pressure is no longer applied.

The mentioned errors are probably equally relevant for dc and ac methods of skin conductance measurement, as long as these are performed within the linear range.

4.8. Summary of findings

- The electrode gels can introduce large changes in the level of the skin conductance over time, corresponding with the amount of free water in the gel and its viscosity.
- Sweating can lead to negative skin conductance responses for wet gels with low viscosity. This inverse sweating response effect probably depends on the thickness of the SC.
- Wet gels may provide an increased sensitivity to sweat activity in partially filled ducts.
- Solid gels have in general a better ability than the wetter gels to return the skin conductance to the baseline during recovery after a period of sweating.
- Wet gels seem to give less repeatable measurements due to the changes introduced by the gel which likely depend on the interindividual SC properties.
- Expectedly, the gels with lower mechanical stability are more susceptible to pressure artifacts.
- The Karaya-based solid gel may introduce a drift in the skin conductance measurement due to lowering of its initially high resistivity.
- The isotonic EDA jelly (type C) may be completely absorbed by the skin during long-term wear.

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