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Skin impedance and electro-osmosis in the human epidermis

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Abstract—*It is well known that skin conductance is dependent on current flow through the skin. It is shown that this may be due to an electro-osmotic effect. This can explain why in some cases negative and in other cases positive potentials increase skin conductance. Electro-osmosis implies current rectification, a powerful skin breakdown mechanism with potentials larger than 50 V, and that negative d.c. potentials are more dangerous than positive on dry skin. Electro-osmosis may be a source of error in skin ionic permeability studies with electrical parameters.*

Keywords—*Electrical safety, Electro-osmosis, Skin electrical breakdown, Skin impedance, Skin permeability*

1 Introduction

IT IS well known that the flow of electric current through the skin often leads to increased conductivity, particularly with negative potential at the measuring electrode (ROSENDAL, 1943, 1944). An electro-osmotic explanation was proposed early on by MUNK (1873) and GÄRTNER (1882). It was also proposed by FLOTTORP (1953) and TREGEAR (1966). REIN (1924) found electro-osmotic transport in the skin *in vitro*, but accepted that it was not the cause of the conductance variations measured *in vivo* (REIN, 1926). GILDEMEISTER (1919, 1928) and EBBECKE (1923) both believed that the conductivity variations were due to changes in the membrane permeability of the living cells of the stratum germinativum. ROSENDAL (1944) did not put forward electro-osmosis as an explanation of the large difference he found between anodic and cathodic conductivity. EDELBERG *et al.* (1960) used the same explanation as Gildemeister and Ebbecke, and thought that the membrane was 'easily accessible from the surface'. Other explanations have been proposed: STEPHENS (1963), LEEMING *et al.* (1970). CARTER *et al.* (1969) reported movement of fluids into and in the skin without proposing any explanation.

In this paper current flow through the skin has been investigated by recording small-signal admittance curves or large-signal current-potential curves. 'Dry' skin has been examined particularly, and the applicability of electro-osmotic theory is discussed in relation to conductance variations, electric breakdown of the skin and as a source of error in electrical studies of skin ionic permeability.

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2 Methods

For the experiments the following items were used:

Measuring electrodes:

(*a*) dry disc covered by hard silver chloride, a part of commercial e.c.g. electrodes (3M, model 2246). Area 56 mm^2. Used when not otherwise stated.

(*b*) dry discs made of aluminium, brass, copper, platinum or stainless steel. Area 56 mm^2.

(*c*) pregelled e.c.g. electrodes (3M, model 2246), effective electrode area (skin area wetted by gel) 3–5 cm^2.

Conductive paste/gel:

Hewlett Packard Redux high viscosity, highly conductive electrode paste (resistivity 9 Ω cm) under the indifferent electrode (16 cm × 8 cm, used on the same arm as the measuring electrode) and for the result of Fig. 7. Fig. 8 was obtained with the gel contained in the pregelled electrode (KC1, low concentration of chloride ions of about 1%).

Potentials of more than 10 V were delivered by d.c. coupled amplifiers especially designed and built by the author. Key specifications included: a gain equal to 100, potential capability of 450 V, current limited to 3 mA, rise/fall time about 10 μs. Large-signal current-potential curves were recorded on a two-channel storage oscilloscope, the current was read by a 1 kΩ shunt resistor. Small-signal (less than 100 nA cm^{-2}) admittance values were measured with a lock-in amplifier (Princeton Applied Research, model 129). It measured either the in-phase and quadrature components of the admittance *Y*, that is conductance

G and capacitance C_p (Figs. 6, 7, 8), or the magnitude of $|Y|$ and the phase angle ϕ (Fig. 3). A d.c. potential up to ±5 V was superimposed on the small a.c. signal, Figs. 6, 7 and 8. Further details can be found elsewhere (GRIMNES, 1982).

Dry skin is defined as the state of the skin when no electrolyte or conductive gel has been applied, after an adequate stabilising period during which the subject relaxes so that his sweat gland activity has reached a stable minimum and GSR-waves are not elicited at nonpalmar sites (GRIMNES, 1982).

3 Results

All skin sites tested, repeatedly showed results similar to those to be reported, except palmar skin. Here skin conductance was much less dependent on current flow.

3.1 *Dry condition*

Fig. 1 shows large-signal current curves found with the dry disc electrode and a.c. potentials of 0·2, 23, 90, 500 Hz. Fig. 1*a* shows the result in slow-motion. The measuring electrode connected to the potential source was placed on the skin at the moment of sweep start. In the first positive half wave the current was very small. A process started in the first negative swing, and the

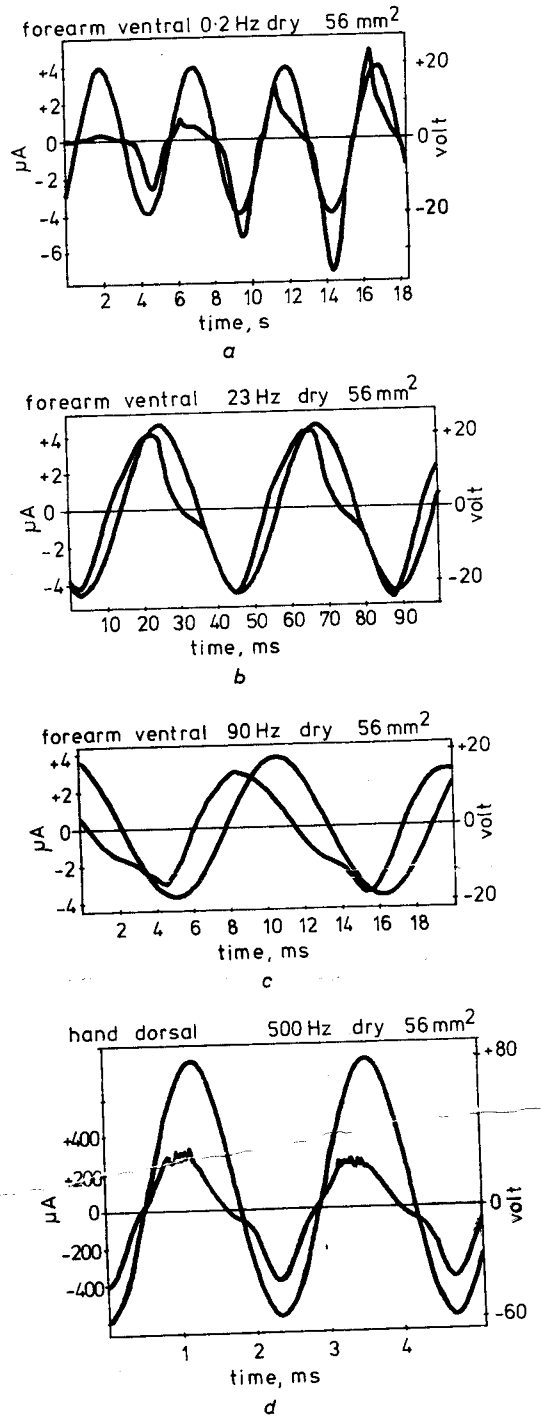

Fig. 1 *Large-signal current curves with the dry disc electrode at four different frequencies. Applied potential is shown as a reference together with the current curves*

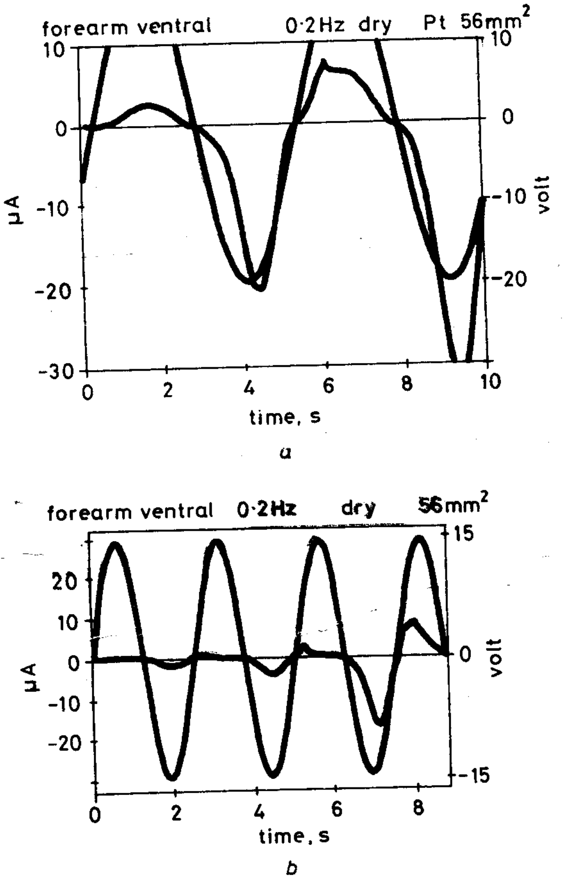

Fig. 2 *Similar conditions as for Fig. 1a, but, (a) with a dry platinum disc electrode, (b) with another test subject*

current peak lagged the potential peak. In the next positive swing the current abruptly stopped, but for each positive swing the time to current stop was longer. For the results in Fig. 1*b*, 1*c*, 1*d* the electrode was placed on the skin, and within a second thereafter the oscilloscope sweep was started. Fig. 1*b* shows both the positive potential current stop and negative current increase clearly. In Fig. 1*c* the effects are less clear, even if the negative potential effect is still discernible. At 200 Hz the current amplitude was about 10 μA and almost purely sinusoidal. At 500 Hz the current curve was a sinusoid with a peak current of about 20 μA. To have a clear nonlinear effect at 500 Hz it was necessary to increase the potential swing. Fig. 1*d* shows the result with +80 V, −60 V swing. The asymetrical potential was necessary to prevent rapid skin breakdown. Positive-potential current stop and negative-potential current increase are retained.

Fig. 2*a* was obtained under similar conditions as Fig. 1*a*, but with higher sweep speed and with the platinum disc electrode. All the dry-disc electrodes made of different metals gave current curves similar to Fig. 1*a*. Fig. 2*b* shows the result with the AgCl disc electrode but with a different test subject. Four different test subjects were measured and similar results obtained.

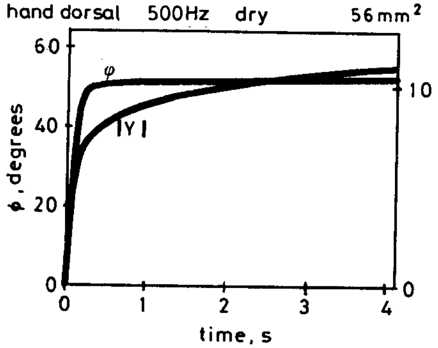

Fig. 3 *Small-signal admittance $|Y|$ and phase angle ϕ from the moment of electrode placement*

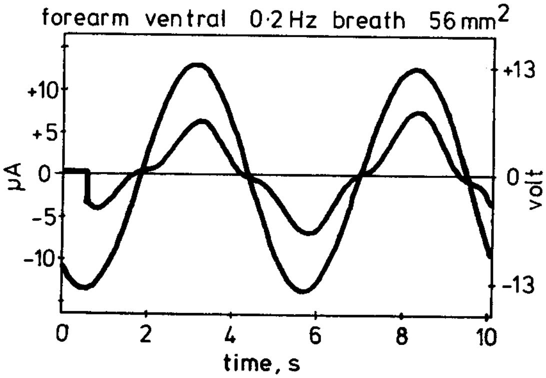

Fig. 4 *Similar conditions as Fig. 1a, but with applied breath*

Fig. 3 shows small-signal admittance $|Y|$ and phase angle ϕ from the moment of electrode onset. The phase angle was capacitive and surprisingly stable right from the start. From Fig. 3 $\phi = 50°$, usually the values were in the range 50°–60°. At lower frequencies the phase angle showed larger variations, but usually the values at 20 Hz were about 35°, and at 0·5 Hz (frequency limit of the lock-in amplifier) in the range 5°–15°.

Fig. 4 was obtained after a short breath had been applied to the skin just before the electrode was placed. Similar curves were sometimes obtained when the test subject had not waited for dry skin condition as defined in this paper. The increased surface humidity reduced the negative potential effect; a rather strong nonlinearity appeared around 0 V. This was a low-frequency effect which was reduced at 20 Hz. Both the negative and positive current peaks slightly lagged the respective potential peaks.

Fig. 5 shows examples of rapid current run-away accompanied by a pricking sensation. Figs. 5*a* and 5*b*

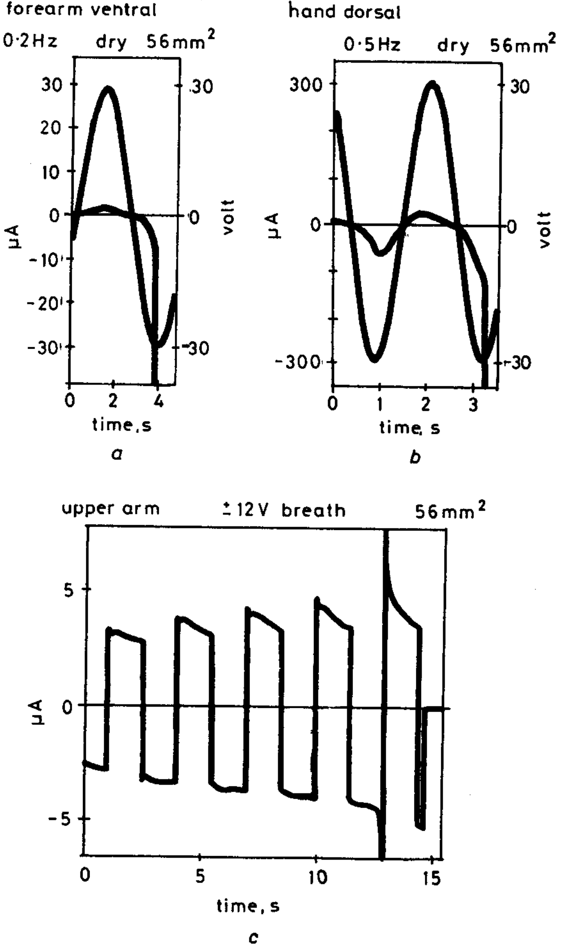

Fig. 5 *Rapid current run-away accompanied by pricking sensation: (a) and (b), two different test subjects, (c) example of reversible current run-away*

show current curves obtained with two different test subjects. Rapid run-away only occurred with negative potentials, but rarely with potentials below 10 V. Fig. 5*c* shows that in the early phase of a run-away the process was completely reversible. The negative current was large enough to elicit a sudden sensation, but the next positive swing was able to reduce the current to the original level. Electrode-skin contact was broken a short time after the last negative swing had appeared. Fig. 5*c* was obtained after a breath had been applied to the skin, and it shows that under that condition a large part of the conductance did not take part in the run-away process.

Figs. 1, 2, 4 and 5 all show large-signal current curves. Fig. 6 shows how the small-signal admittance values changed when a large d.c. potential was applied. It was recorded after the electrode had been in contact with the skin for about 1 min.

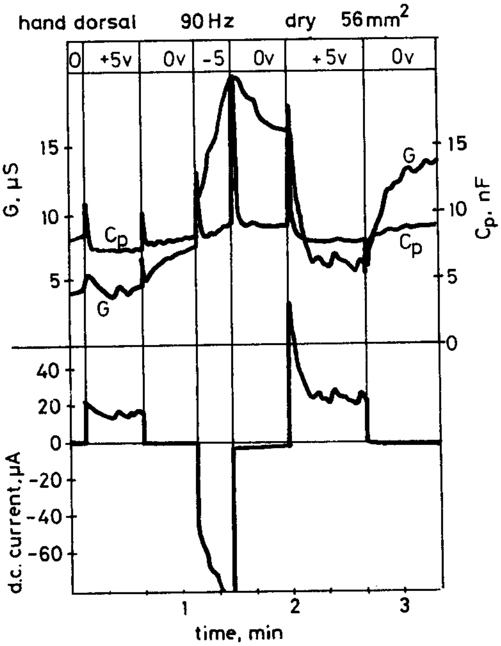

Fig. 6 *Small-signal admittance as a function of applied d.c. potential, dry condition*

3.2 *Results with applied conductive paste/gel*

For the recording of large-signal current curve, potential amplitudes with dry electrodes up to 80 V were used. With a conductive paste/gel between the skin and the electrode disc the potential was reduced to keep the current below the threshold of sensation. Low d.c. potentials of ±5 V maximum were used, with a correspondingly slow response, Fig. 6 has already been presented as an example. Fig. 7 shows the result obtained with a highly conductive, high-viscosity paste under the disc electrode. Effective electrode area (skin area wetted by the electrolyte) was measured as 135 mm^2. Fig. 7 shows that only negative d.c. potentials increased the conductance G. Fig. 8 shows a different result with the weak, lower-viscosity gel. It was recorded right from the moment the test subject arrived at the measuring station. The initial conductance decrease during the first half hour shows a typical transition from nondry to dry skin as defined in Section 2.

Note the constant skin capacitance C_p during the large initial change in G. -2 V did not increase the conductance, -3 V did slightly, while $+3$ V increased the conductance strongly. Note that the return to 0 V after positive polarity reduced the conductivity as if gel was driven out of the ducts by an active process.

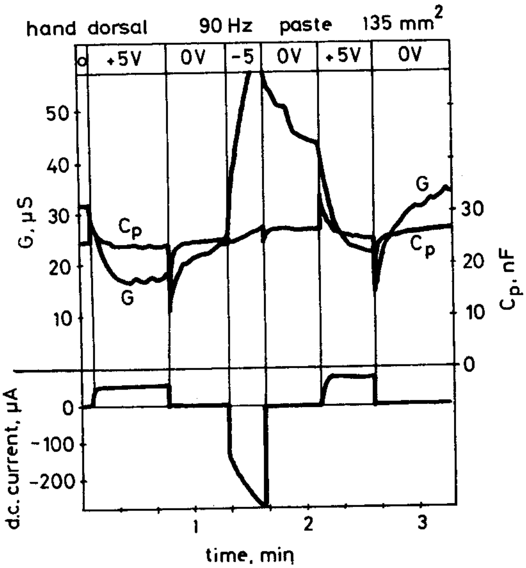

Fig. 7 *Similar conditions to those of Fig. 6, but with high-viscosity, strong paste electrolyte*

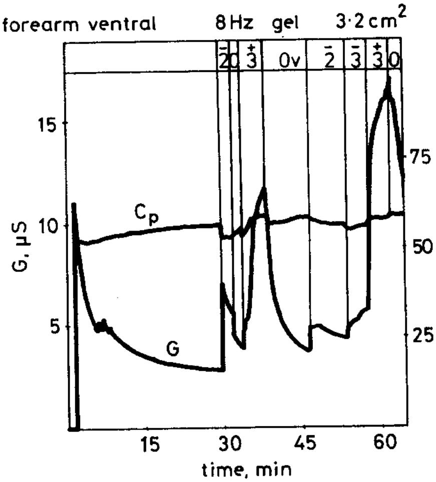

Fig. 8 *Similar conditions to those of Fig. 6, but with lower viscosity, weak gel electrolyte*

4 Discussion

The current curve shown in Fig. 1*a* reveals a runaway process when the potential was negative, but that positive potential again reduced the current. First we must discuss whether the results can have been influenced by electrode effects, either by electrode polarisation impedance or the formation of passivating layers on the electrode surface. Electrode polarisation impedance is usually less than $1\,\mathrm{k}\Omega\,\mathrm{cm}^2$(10 Hz) (ROSENDAL, 1940), while in our case skin impedance was larger than $300\,\mathrm{k}\Omega\,\mathrm{cm}^2$ (8 Hz), Fig. 8. The polarisation impedance will be larger because the electrolytic film on the electrode is thin, but control experiments have revealed that the polarisation impedance is approximately doubled (0·5 Hz) when film thickness is reduced from more than 1 mm to about 20 μm. The influence of electrode polarisation impedance was therefore negligible, as it often is with skin surface electrodes at low frequencies (BLANK *et al.*, 1946; SWANSON *et al.*, 1974). Effects of passivating layers are not probable because all metals, platinum included, showed the same effect, Fig. 2*a*. These conclusions do not exclude the possibility that the effect shown in Fig. 1*a* is partly due to a change in the (wetted) contact area of the electrode surface, this will be further discussed below.

We will try to explain the result of Fig. 1*a* as the result of an electro-osmotic transport of water (electrolyte). In dry skin most of the pores are empty or semifilled, but conductive films are always present (RANDALL, 1946; THOMAS *et al.*, 1957; GRIMNES, 1982). With a dry electrode on dry skin a conductance increase can occur when water is transported from deeper layers towards the skin surface and the electrode. Water will always be transported towards the skin surface by diffusion (insensible respiration). Under an electrode plate it will be trapped and the conductance will increase. The process is slow however, typically 10 min (BLANK *et al.*, 1946). We postulate that the process can be accelerated by electro-osmotic transport of water, resulting in an increase in pore filling, skin surface films and wetted electrode area.

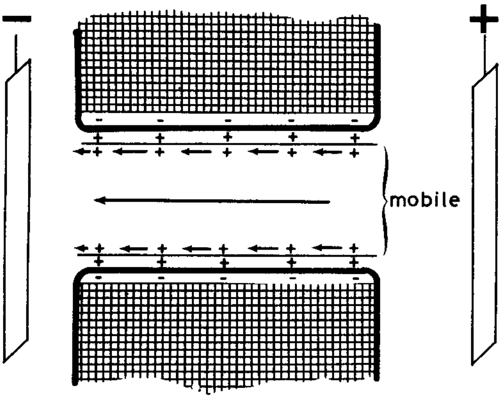

Fig. 9 *Model of electro-osmosis in a capillary membrane*

Electro-osmosis is a capillary transport mechanism of bulk liquid caused by viscous forces when the mobile part of the charged double-layer is moved by an electric field*. This is illustrated in Fig. 9 where the whole central part (core) of the capillary is moved by the mobile (diffuse) part of the double layer.

The flow F through a filled capillary per second is (GLASSTONE, 1946):

$$F = \frac{\zeta \varepsilon r^2 V}{4 l \eta} = \frac{\zeta \varepsilon I \rho}{4 \pi \eta} \qquad (1)$$

Where ζ = electrokinetic potential, r = radius of capillary, l = length of capillary, ε = dielectric constant of liquid, V = potential difference, I = total electric current, ρ = resistivity of liquid, η = viscosity of liquid.

Although electro-osmosis is an effect controlled primarily by potential, the flow per second as a function of current (right-hand side of eqn. 1) is a particularly useful expression because it contains no data on capillary dimensions. It is equally valid for a capillary of variable radius r or length l, and it can easily be shown that it is also valid in the case of an empty capillary with liquid films on the wall. For further information on electro-osmosis, see GLASSTONE (1946) and SCHMID (1951).

REIN (1924) reported that solutions of NaCl in the skin are transported towards the cathode as the result of electro-osmosis, and that the transport increases with decreasing NaCl concentration in the range 5–0·5%. GLASSTONE (1946) also reported that the stronger the electrolyte, the weaker the electro-osmosis.

The current $i(t)$, in the negative half period of an applied potential $U \sin \omega t$, through a membrane with electro-osmotic conductance increase in the capillaries, is derived in Appendix A:

$$i(t) = G_0 U \sin \omega t \, e^{E(\cos \omega t + 1)} \qquad (2)$$

where E is an electro-osmotic factor defined in eqn. 3 in Appendix A. From eqn. 2 and Fig. 11 it is clear that the electro-osmotic effect introduces a current lag with respect to the applied potential, corresponding to an apparent inductance. In Appendix B it is shown that the electro-osmotic acceleration of inert liquid mass is not a rate-determining factor.

It is generally accepted that skin admittance is of electrolytic origin, and that the stratum corneum is

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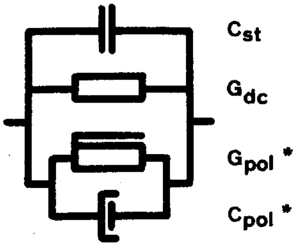
* *The expressions electro-osmosis and iontophoresis have been used differently in the literature. Here iontophoresis will be used in relation to the migration of ions caused by an electric field. Electro-osmosis is the transport of bulk liquid by viscous forces exerted by the migrating ions of the diffuse part of the double-layer formed in the electrolyte on the capillary walls. Fig. 9. If the whole core of a capillary is moved by the surrounding migrating ions, the number of bulk molecules so transported can be much larger than the number of driving ions.*

the admittance-determining part of human skin (ROSENDAL, 1945).

The equivalent circuit of the small-signal skin admittance is usually an ohmic resistor in parallel with a 'polarisation' capacitive admittance (TREGEAR, 1966; YAMAMOTO and YAMAMOTO, 1978). It is shown in Fig. 10 that a capacitor has been added for the electrostatic skin capacitance. The polarisation admittance Y_{pol} has the peculiar property of a frequency-independent phase shift (BARNETT, 1938). The magnitude of Y_{pol} decreases towards zero when frequency decreases towards d.c. because it is caused by bound ions which are not free to move large distances. At very low frequencies the ohmic resistance G_{dc} therefore dominates, and the corresponding phase shift is small. G_{dc} is dependent on the number of free ions in the stratum corneum, which again is dependent on the amount of electrolyte there. Our postulate is that electro-osmosis can affect this amount and thereby also G_{dc}.

With increasing frequency the polarisation current through Y_{pol} increases, at the same time the electro-osmotic effect decreases according to eqns. 2 and 3. With the potential swing used, ±20 V, the current was more and more sinusoidal and led the potential as frequency was increased; at 500 Hz it was a pure sinusoid. Consequently Fig. 1 can be explained as follows.



* *Frequency dependent*

Fig. 10 *Equivalent circuit of skin admittance*

At 0·2 Hz, the polarising current through Y_{pol} is negligible. The current is mainly due to G_{dc}, and in negative half periods it increases according to eqn. 2 and Fig. 11 in Appendix A. With increasing frequency the polarisation current through Y_{pol} gradually dominates. At the same time the electro-osmotic effect is also reduced and at 90 Hz it was only just visible. At 500 Hz and increased negative potential (Fig. 1*d*), the phase shift due to Y_{pol} and electro-osmosis counteract so that the current is roughly in phase with the potential. From Fig. 3 we know that the small-signal phase shift is about 50°. Under the assumption (not examined in this paper) that the large-signal phase shift due to Y_{pol} is also about 50°, the electro-osmotic phase-shift ϕ_e is about 50°, and from Fig. 12 this corresponds with an electro-osmotic factor E equal to about 2.

The current curve shown in Fig. 1*a* has an abrupt fall with positive potential. This fall can easily be explained from electro-osmotic theory. Positive potentials transport water away from the electrode, and as there is no reservoir at the electrode, liquid rupture will occur near the electrode surface. The current then falls abruptly, and the water transport stops. It is easy to presuppose that electro-osmotic transport would pump water back and forth with applied a.c., but Fig. 1*a* and electro-osmotic theory indicate a rectifying mechanism with a net transport of water and a corresponding rectified d.c. component. This explanation also covers the noisy part of the current curve shown in Fig. 1*d* at large, positive potential. Water is violently thrown back from the electrode, but as soon as current falls it pours back to give renewed contact and current rise, and so on. MUELLER *et al.* (1953) also found this noisy part with positive potential, but they interpreted it as a reversible

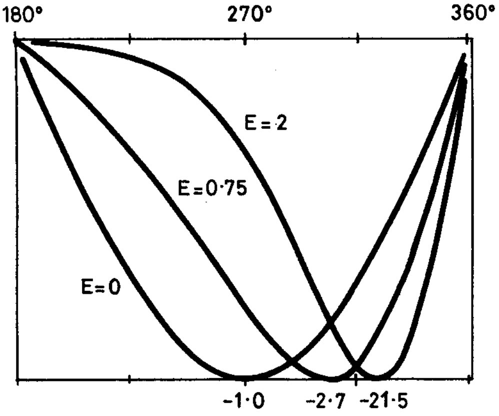


Fig. 11 *Current i(t) through a semifilled capillary, calculated from eqn. 2 for three different values of E. Normalised curves with current amplitudes given underneath*

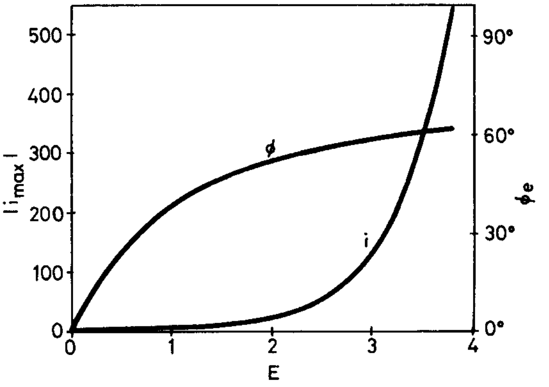


Fig. 12 *Electro-osmotic phase-shift ϕ_e from eqn. 4, and current maximum $|i_{max}|$ relative to the current maximum in the preceding positive potential swing*

dielectric breakdown dependent on the dielectric strength of the skin. It has been shown that dielectric breakdown of the skin (breakdown of C_{st} in Fig. 10) is improbable at potentials less than 100 V (GRIMNES, 1983).

Eqn. 1 and the literature (REIN, 1924; GLASSTONE, 1946) indicate that electro-osmosis is strong with weak, nonviscous electrolytes, but weak with strong, viscous electrolytes. With applied electrode electrolyte the case with strong, viscous electrolytes should be similar to the dry case, because the electro-osmotic transport of the electrode electrolyte is negligible. This is in agreement with Fig. 7. With a weak, less viscous electrolyte the case is complicated. Electro-osmosis and increased G_{dc} should be possible from both sides, dependent on which side has the strongest effect. Fig. 8 shows that it was possible to have a strong increase in G_{dc} with positive potentials, this has never been registered with dry disc electrodes. The result is in agreement with the results of ROSENDAL (1944), he found a larger difference between anodic and cathodic conduction the stronger the contact electrolyte. However, he interpreted the result differently.

Figs. 6, 7 and 8 show constant skin capacitance C_p (the peaks were due to lock-in amplifier overload transients). This is in agreement with the equivalent circuit of Fig. 10 and shows that Y_{pol} is independent of G_{dc}, as earlier found by YAMAMOTO and YAMAMOTO (1978) and GRIMNES (1982*a*), and even valid during the first half hour transition shown in Fig. 8. Earlier results (GRIMNES, 1982*a*) and Figs. 6, 7 and 8 show that without electro-osmosis there is an active absorption process emptying the pores of electrolyte, just as the well known sweat re-absorption (KUNO, 1956). The emptying process can be stronger than the penetrating effect of applied electrolytes. Also with an applied electrolyte there will be many nonfilled pores for many hours, and electro-osmotic filling will still be of interest.

Electro-osmosis is not the only conceivable current-induced admittance-increasing factor in the skin. Generally the following effects may be listed:

(i) temperature rise, temperature coefficient of electrolytic conductance is about +2%/°C

(ii) high-field dielectric breakdown

(iii) high-field electrolytic conductance increase (Wien-effect)

(iv) excitation of sweat glands

(v) iontophoresis, electrolysis

(vi) mechanical changes, e.g. caused by water vapour pressure

(vii) electro-osmosis.

The stratum corneum consists of dead cells, and no changes in the cell membranes of living cells, as earlier proposed, can be directly involved. Indirectly however, the living cells under the stratum corneum, e.g. the sweat glands, can intervene.

Which of the listed factors can explain the result in Fig. 1*a*? Temperature effects, Wien-effect and excitation of the sweat glands (WILKINS *et al.*, 1938), can be rejected because of the strong polarity dependence found. Dielectric breakdown involves mechanical destruction. Mechanical destruction is irreversible, but Fig. 5*c* showed that the process, at least in an early phase, was reversible. Iontophoresis could very well include a strong polarity dependence, but it is an electrolytic effect dependent on a proper amount of electrolyte. It is difficult to see how the products of electrolysis at the electrode or in the deeper, vascular parts of the skin, with the very low initial current, can cause as large a conductance increase as seen in Fig. 1*a*. It is also difficult to see how the process can be fast enough, and how the product of electrolysis can stop the current so effectively with 20 V applied amplitude. An electro-osmotic explanation is supported by the following facts.

(*a*) Skin impedance is of electrolytic origin (ROSENDAL, 1945). With a dry electrode on dry skin the electrolytic conductance is small, but it is not zero (THOMAS *et al.*, 1957; GRIMNES, 1982). A substantial electrolytic conductance increase is only possible if the amount of electrolyte in the stratum corneum, and in contact with the electrode plate, is increased. Thus the basic conditions for an electro-osmotic effect are present: a membrane with empty pores but a nonzero initial conductance, and with an electrolytic reservoir on one side and an electrode on the other.

(*b*) The current increase with a dry, negative electrode is in agreement with the results of REIN (1924).

(*c*) The current stop and current noise with positive potential are readily explained from electro-osmotic theory.

(*d*) The results with weak and strong electrolytes are in agreement with electro-osmotic theory, and also with the classical result of ROSENDAL (1944).

(*e*) Rapid skin breakdown with dry electrodes occurred only with negative potentials, Fig. 5. The reversibility in the initial phase of current run-away is in agreement with electro-osmotic theory, and cannot be due to a destructive effect on the stratum corneum cells.

It is concluded that under certain conditions electro-osmosis is an important factor controlling skin current. Our conclusion is a confirmation of old concepts dating back to the last century, and sometimes revived, sometimes rejected. Some authors take electro-osmotic transport in the skin for granted (ABRAMSON, 1938; ABRAMSON and GORIN, 1939; GELDARD, 1972). ABRAMSON and GORIN (1939) also assumed electro-osmotic filling of skin pores; this must necessarily be accompanied by an increase in skin conductance. FLOTTORP (1953) and TREGEAR (1966) assumed that the current increase was due to electro-osmosis, but did not pursue the question. TEORELL (1959) demonstrated electro-osmotic conductance variations and rectification in an artificial membrane, but only analysed completely filled capillaries separating electrolytes of different concentrations. Our conclusion is well in agreement with the general statement of TEORELL (1953) in his chapter on electro-endosmosis that '...to include the electrically induced water flow, is of the utmost importance also for those dealing with biological systems, where the water transport is as striking a phenomenon as the molecular, or ionic transport.'

Our conclusion is in contrast with those of EBBECKE (1923), GILDEMEISTER (1919, 1928), ROSENDAL (1944) and EDELBERG *et al.* (1960). ROSENDAL (1943) in his introduction favoured electro-osmosis as a factor changing the 'electrolyte content of the skin'. He knew that the negative conclusions of REIN (1926), EBBECKE (1923) and GILDEMEISTER (1919, 1928) were based on the misleading interpretation of skin impedance results (use of too high measuring frequency and a corresponding Y_{pol} domination). However, he did not use electro-osmotic theory in the discussion of his results. Later he did not mention electro-osmosis and discussed conductance variations on the basis of supposed impermeability of the skin to inorganic cations, the size of anions etc., i.e. iontophoretic parameters (ROSENDAL, 1944). As already mentioned his experimental results are in agreement with our conclusion.

Our conclusion implies that iontophoretic effects are usually much smaller than electro-osmotic effects, even if the former cannot be neglected in special cases. The statement of STEPHENS (1963) that the origin of nonlinearity of skin resistance is due to 'changes in the ionic population of the skin, produced by ion migration' is, according to our conclusion, inaccurate because only very few of the migrating ions are of importance for the change; those loosely attached to the double-layer on the capillary walls. The idea of LEEMING *et al.* (1970) that products of electrolysis (alkali under the negative electrode) gradually destroy the insulating epidermis, is also an iontophoretic explanation. It was partly rejected by MOLITOR *et al.* (1939), but most probably should be considered as a possible long-term effect. YAMAMOTO and YAMAMOTO (1981) also found that very low-frequency, large potentials caused variations mainly in skin conductance, but did not discuss the physical processes involved.

SCHEUPLEIN *et al.* (1971) wrote that little is known about ionic permeability in the skin, but the scarce results indicate that the difference between different electrolytes is small. Our results indicate that the sweat ducts of nonpalmar skin are emptied by an active process, whether they have been filled from the surface by electro-osmosis or by the sweat glands. Ionic permeability will therefore be highly dependent on the sweat gland activity and degree of 'dryness' of the skin just as skin admittance is. Electrical methods used to examine skin ionic permeability must be used with caution because the amount of liquid forming ionic paths will be influenced by the current flow.

Fig. 5 shows examples of rapid current run-away accompanied by a pricking sensation, giving a subjective impression of skin breakdown. Fig. 5*c* proves that the pricking was an early warning before skin destruction started. In the initial phase the current run-away was reversible. The warning is early because the current is concentrated to pores with very high local current density and a high probability of nerve excitation. It is well known that electricity is more dangerous with wet skin than with dry skin. Electro-osmosis draws water out through the dry stratum corneum so that the case becomes more like wet conditions. Conductors connected to a negative potential source are therefore more dangerous to touch with dry skin than positive conductors.

Below 10 V the run-away was slow. The experiment in Fig. 1*d* implied a potential swing of 60 V, and skin breakdown was then so rapid that an asymmetrical potential swing was used. The electro-osmotic effect increased from slow to very rapid in the potential range 10–50 V. This is in accordance with eqns. 2 and 3 because the current is dependent on the exponent of the potential. With power-line potentials in the range 110–250 V, electro-osmotic transport will be very rapid, while dielectric breakdown is still slow (GRIMNES, 1983). It is possible from eqn. 6 to find the time necessary to reach either the ordinary threshold of sensation (1 mA) or the dangerous level (20 mA) in an electroshock situation with 230 V r.m.s. (325 V peak). From Fig. 1*a* G_0 is estimated to 10 nS, and $K = 0{\cdot}05\,\mathrm{SC}^{-1}$. The time T will be 1·1 s and 1·7 s respectively. The process is quicker because of temperature effects. It should be possible according to electro-osmotic breakdown theory to touch a live powerline conductor with dry skin for a fraction of a second without being hurt. This is in agreement with verbal reports from experienced electricians. It is also in agreement with the delayed current rise found with dry electrodes on a corpse (own results, unpublished). It is in agreement with the result of CARTER *et al.* (1969), but not for as high a potential as 250 V r.m.s.

It is difficult to control the adequacy of the mathematical modelling of Appendix A and eqns. 2 and 9 accurately because the actual values of the parameters involved are uncertain. However, a rough estimate can be done from Fig. 1*a*. From Fig. 1*a* the conductance increase per quantity of electricity can be estimated to $K = 0{\cdot}05\,\mathrm{SC}^{-1}$, then E is equal to (0·2 Hz) 0·8 from eqn. 3. From eqn. 4 $\phi_e = 33°$, this is roughly in agreement with the phase shift seen in Fig. 1*a*. On the other hand the peak current is about 10 in the first negative swing relative to the current in the first positive swing. From Fig. 12 this corresponds to

ϕ_e of about 50°, which is larger than the above cited values. The discrepancy can hardly be explained from a certain capacitive contribution from Y_{pol}, even if the small-signal phase-angle at 0·5 Hz was found between 5° and 15°. A probable source of error is the assumption made as to how the transported water is distributed. Eqns. 2 and 9 are based on the fact that the water is assembled in the end of a capillary tube of uniform radius. However, the real geometry is more complicated; the water must pass the pore orifice and a part of the skin surface before it is stopped at the electrode plate.

References

ABRAMSON, H. A. (1938) Production of wheals in the human skin. *Science*, **87**, 299.

ABRAMSON, H. A. and GORIN, M. H. (1939) Relationship of skin permeability to electrophoresis of biologically active materials into the living human skin. *J. Phys. Chem.*, **43**, 335–346.

BARNETT, A. (1938) The phase angle of normal human skin. *J. Physiol.*, **93**, 349–366.

BLANK, I. H. and FINESINGER, J. E. (1946) Electrical resistance of the skin. *Archs. Neurol. Psychiat.*, **56**, 544–557.

CARTER, A. O. and MORLEY, R. (1969*a*) Electric current flow through human skin at power frequency voltages. *Br. J. Ind. Med.*, **26**, 217–223.

CARTER, A. O. and MORLEY, R. (1969*b*) Effects of power frequency voltages on amputated human limb. *Ibid.*, **26**, 224–230.

EBBECKE, U. (1923) Ueber Gewebsreizung und Gefässreaktion. *Pflügers Arch*, **199**, 197–216.

EDELBERG, R., GREINER, T. and BURCH, N. R. (1960) Some membrane properties of the effector in the galvanic skin response. *J. Appl. Physiol.*, **15**, 691–696.

FLOTTORP, G. (1953) Effect of different types of electrodes in electrophonic hearing. *J. Acoust. Soc. Am.*, **25**, 236–245.

GÄRTNER, G. (1882) Untersuchungen über das elektrische Leitungsvermögen der menschlichen Haut. *Mediz Jahrb Ges Arzte* (Wien), 519–549.

GELDARD, F. A. (1972) *The human senses*. John Wiley & Sons, 261.

GILDEMEISTER, M. (1919) Über elektrischen Widerstand, Kapazität und Polarisation der Haut. *Pflügers Arch*, **176**, 84–105.

GILDEMEISTER, M. (1928) Die passiv-elektrischen Erscheinungen im Tier-und Pflanzenreich. Die Elektrizitätserzeugung der Haut und der Drüsen. In *Handbuch der normalen und pathologischen Physiologie, Energieumsatz*, BETHE, A., BERGMAN, G., EMBDEN, G., ELLINGER, A. (Eds.), Springer, 666.

GLASSTONE, S. (1946) *Textbook of physical chemistry*. Van Nostrand.

GRIMNES, S. (1982) Psychogalvanic reflex and changes in electrical parameters of dry skin. *Med. & Biol. Eng. & Comp.*, **20**, 734–740.

GRIMNES, S. (1982*a*) Skin capacitance as a measure of effective electrode area. Proc. World Cong. Med. Phys. Biomed. Eng., Hamburg, 7.16.

GRIMNES, S. (1983) Dielectric breakdown of human skin *in vivo*. *Med. & Biol. Eng. & Comp.*, **21**, 379–381.

KUNO, Y. (1956) *Human perspiration*. Thomas, 61.

LEEMING, M. N., RAY, C. and HOWLAND, W. S. (1970) Low-voltage, direct-current burns. *JAMA*, **214**, 1681–1684.

MOLITOR, H. and FERNANDEZ, L. (1939), Experimental studies on the causes and prevention of iontophoretic burns. *Am. J. Med. Sci.*, **198**, 778–785.

MUELLER, E. E., LOEFFEL, R. and MEAD, S. (1953) Skin impedance in relation to pain threshold testing by electrical means. *J. Appl. Physiol.*, **5**, 746–752.

MUNK, H. (1873) Ueber die galvanische Einführung differenter Flüssigkeiten in den unversehrten lebenden Organismus. *Arch. Anat. Physiol. Wissenschaftl. Med.*, 505–516.

RANDALL, W. C. (1946) Sweat gland activity and changing patterns of sweat secretion on the skin surface. *Am. J. Physiol.*, **147**, 391–398.

REIN, H. (1924) Experimentelle Studien über Elektroendosmose an überlebender menschlicher Haut. *Z. Biol.*, **81**, 125–140.

REIN, H. (1926) Untersuchungen über den Gleichstromwiderstand. *Ibid.*, **84**, 118–142.

ROSENDAL, T. (1940) The conducting properties of the human organism to alternating current. Thesis, Copenhagen, Munksgaard, 82.

ROSENDAL, T. (1943) Studies on the conducting properties of the human skin to direct current. *Acta. Physiol. Scand.*, **5**, 130–151.

ROSENDAL, T. (1944) Further studies on the conducting properties of human skin to direct and alternating current. *Ibid.*, **8**, 183–203.

ROSENDAL, T. (1945) Concluding studies on the conducting properties of human skin to alternating current. *Ibid.*, **9**, 39–49.

SCHEUPLEIN, R. J. and BLANK, I. H. (1971) Permeability of the skin, *Physiol. Rev.*, **51**, 702–747.

SCHMID, G. (1951) Zur Elektrochemie feinporiger Kapillarsysteme. Elektroosmose. *Z. Elektrochem.*, **55**, 229–237.

STEPHENS, W. G. S. (1963) The current-voltage relationship in human skin. *Med. El. Biol. Eng.*, **1**, 389–399.

SWANSON, D. K. and WEBSTER, J. G. (1974) A model for skin-electrode impedance. In *Biomedical electrode technology*, MILLER, H. A. and HARRISON, D. C. (Eds.), Academic Press.

TEORELL, T. (1953) Transport processes and electrical phenomena in ionic membranes. In *Progress in Biophysics*, BUTLER, J. A. V. and RANDALL, J. T. (Eds.), Pergamon Press.

TEORELL, T. (1959) Electrokinetic membrane processes in relation to properties of excitable tissues. *J. Gen. Physiol.*, **42**, 831–845.

THOMAS, P. E. and KORR, I. M. (1957) Relationship between sweat gland activity and electrical resistance of the skin. *J. Appl. Physiol.*, **10**, 505–510.

TREGEAR, R. T. (1966) *Physical functions of skin*. Academic Press, 66, 70.

WILKINS, R. W., NEWMAN, H. W. and DOUPE, J. (1938) The local sweat response to faradic stimulation. *Brain*, **61**, 290–297.

YAMAMOTO, T. and YAMAMOTO, Y. (1978) Dispersion and correlation of the parameters for skin impedance. *Med. & Biol. Eng. & Comp.*, **16**, 592–594.

YAMAMOTO, T. and YAMAMOTO, Y. (1981) Non-linear electrical properties of the skin in the low frequency range. *Med. & Biol. Eng. & Comp.*, **19**, 302–310.

YUAN, S. W. (1970) *Foundations of fluid mechanics*. Prentice-Hall, 291.

Appendix A

Let us consider a membrane with semifilled capillaries and a reservoir of water (electrolyte) on one side. Let us assume that water is drawn from this reservoir into the capillaries when a negative potential is applied to an electrode on the dry side of the membrane. Let us further assume that the increase in conductivity dG is proportional to the amount of water transported into the capillaries. According to eqn. 1 this amount is proportional to the quantity of current $i(t)dt$ through the capillaries. Let K denote a positive constant equal to the increase dG caused by $|i(t)dt|$. With a sinusoidal potential the current is:

$$i(t) = (U \sin \omega t) K \int i(t)dt$$

By differentiating both sides we obtain a linear, homogeneous, first-order differential equation. The solution of this is:

$$i(t) = K_1 \sin \omega t \exp\left(\frac{KU}{\omega} \cos \omega t\right)$$

where K_1 is a constant which must be determined from the initial conditions. We consider electro-osmosis during a negative half period, starting at $\omega t = \pi$. Let G_0 be the initial conductance at $\omega t = \pi$, then:

$$i(t) \to G_0 U \sin \omega t \quad \text{as } t \to \pi$$

then

$$K_1 = G_0 U e^E$$

where

$$i(t) = G_0 U \sin \omega t \, e^{E(\cos \omega t + 1)} \qquad (2)$$

and

$$E = \frac{KU}{\omega} \qquad (3)$$

Maximum current i_{max} can be found by differentiating eqn. 2, finding the phase angle of the driving voltage at which a current maximum occurs, and using this phase angle in eqn. 2. The phase shift due to electro-osmosis ϕ_e is this phase angle minus 270°, that is:

$$\phi_e = \arcsin\left[\sqrt{\left\{1 + \frac{1}{(2E)^2}\right\}} - \frac{1}{2E}\right] \qquad (4)$$

The current according to eqn. 2 is shown in Fig. 11 for three different values of E. The electro-osmotic effect creates a current lag, indicating that the effect has inductive properties. From eqn. 4 it can be seen that the phase shift is only dependent on the electro-osmotic factor E. When $f \to 0$, $\phi_e \to 90°$, corresponding to a pure inductance. When $f \to \infty$, $\phi_e \to 0°$, the *apparent* reactance becomes negligible. *Apparent* inductance when conductance increases under current flow is a well established term in general membrane theory (TEORELL, 1953).

Fig. 12 shows $|i_{max}|$ and ϕ_e as a function of E. The values of $|i_{max}|$ are given relative to $|i_{max}| = 1$ for $E = 0$, that is the case without electro-osmosis.

From eqn. 2 it is found that for $\omega t = 2\pi$ (360°), the conductance has increased by the factor e^{2E} during the negative half cycle. When n half periods have passed the conductance G is:

$$G = G_0 e^{2En} = G_0 \exp\left(\frac{KUT}{\pi}\right). \qquad (5)$$

and the time T to reach a certain conductance G:

$$T = \frac{\pi}{KU} \ln \frac{G}{G_0} \qquad (6)$$

This is under the condition that the current has been zero during the whole positive half period; our experimental results indicate that this condition is not completely fulfilled.

Eqn. 2 shows that the conductance increase is frequency dependent through E and eqn. 3. Eqns. 2 and 4 are valid for only one half period, E is therefore a factor determining the current waveform. Eqns. 5 and 6 show that the total conductance increase over more than one period is dependent not on E or the frequency, but on K and the total time elapsed.

The constant K is the conductance increase per quantity of current, that is SC^{-1}, corresponding to the amount of water transported per coulomb M. M can be found from eqn. 1:

$$M = \frac{\zeta \varepsilon \rho}{4\pi\eta} \qquad (7)$$

The corresponding conductance increase is dependent on how M is distributed in a semifilled capillary. Let us assume a capillary of uniform radius and a conductive wall film of uniform thickness. Let A_t be the cross-sectional area of the empty volume and A_v that of the conductive film. If M is collected at the end of the capillary, then:

$$K = \frac{M}{\rho l^2} \frac{A_v}{A_v + A_t - \dfrac{M}{l}}. \qquad (8)$$

When $M << (A_v + A_t)$, eqn. 8 can be written:

$$K = \frac{M}{\rho l^2} \frac{A_v}{A_v + A_t} \qquad (9)$$

Eqn. 9 shows that K is a constant because it is proportional to M, eqn. 3 is then also correct. When M is large however, eqn. 8 shows that K is increasing faster than M. In this case E is increasing during a negative half period.

Appendix B

It is necessary to examine the electro-osmotic acceleration of bulk liquid as an inert mass. We assume that the diffuse part of the double-layer reaches a constant velocity at the moment the potential is applied. Consider then an infinite plate in an infinite volume of liquid. The plate is suddenly put into motion with a constant velocity. The velocity penetration into the liquid, defined as the boundary layer b, is given by (YUAN, 1970):

$$b = \sqrt{\left(\frac{\pi\eta t}{D}\right)} \qquad (10)$$

where D is the liquid density.

The time t_d for the velocity to penetrate to a distance δ is:

$$t_d = \delta^2 \frac{D}{\pi\eta} \qquad (11)$$

If we put film thickness $\delta = 10\,\mu\mathrm{m}$ (filled sweat duct) and $D = 10^3\,\mathrm{kg\,m^{-3}}$ we get $t_d = 40\,\mu\mathrm{s}$. We can therefore conclude that the acceleration of bulk liquid is not rate determining, the electro-osmotic transport is viscosity determined according to the theory presented in Appendix A.