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A Noninvasive Thermal Microsensor for Skin Characterization

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A noninvasive microsensor was designed for the study of the time course of both water content and microcirculation of the skin related to physiological, pharmacological, and cosmetics experiments. Five homogeneous phantoms were designed for the calibration of the probe-instrument system and one bilayer phantom was designed to simulate the epidermis and the dermis. The feasibility and the reliability of the method were shown in the experiment performed *in vivo* on human skin.

1. Introduction

Skin characterization is currently in high demand from cosmetic and pharmaceutical industries and as well as from consumers. This characterization requires quantitative, repetitive and rapid *in vivo* measurements on human skin. The laws of citizen protection require these measurements to be noninvasive, painless and nontoxic.

A noninvasive instrumentation using a continuous heating method for characterizing skin has been designed by Dittmar *et al.*[1] However, this method needs a different probe with a specific measurement depth to explore each skin layer: one for the measurement of blood flow in capillary layers and one for the measurement of the water content of the superficial layers of the skin.[2, 3] The transient thermal method allows exploration of all

skin layers using only one probe. This method was first developed to determine tissue blood flow with implantable microprobes.[4, 5]

The aim of this research is to design a probe-instrument system to study the time course of both skin water content and skin microcirculation related to physiological, pharmacological and cosmetic experiments.

2. Materials and Methods

2.1 *Principle of measurement*

The skin water content and microcirculation estimations are carried out by the measurement of thermal conductivity. The thermal conductivity of a perfused tissue depends on the relative amount of proteins, lipids and water, and tissue blood flow.[6]

The principle of measurement is based on a two-phase cycle (Fig.1).

Passive phase (duration = 24 s): the baseline skin temperature is measured by the probe-instrument system during this phase with no heating.

Active phase (duration = 6 s): the probe's temperature is regulated by a proportional, integral, derivative (PID) controller to maintain it at a constant temperature increment above the skin baseline temperature measured prior to heating. The heating power (a few milliwatts) necessary to maintain this temperature increment is proportional to first the water content (micro-thermal field propagation in the nonvascularized superficial layers of the skin) and second the microcirculation (micro-thermal field propagation in the capillary layers of the skin).

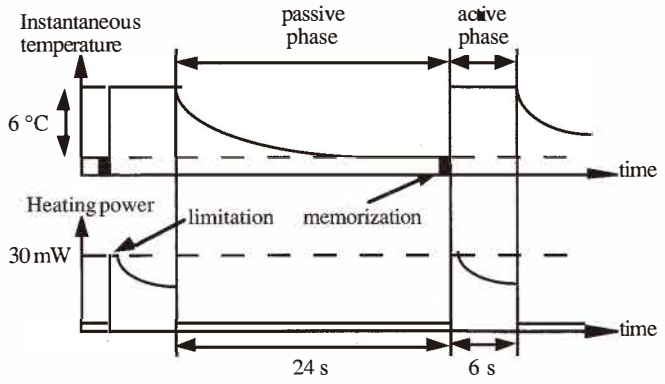

Fig. 1. Principle of measurement.
The probe is maintained at a constant temperature increment above the skin baseline temperature during the active phase (the skin baseline temperature being measured and memorized at the end of the passive phase). At the beginning of the active phase, the heating power is maximal (= 30 mW) and then it decreases. This decreasing phase of the heating power is used to determine both thermal conductivities of the skin.

2.2 *Instrumentation*

The probe consists of a disc (8 mm diameter and 2.5 mm thickness) whose active part is a microthermistor (Betatherm® 10K3MCD2). This thermistor (active part: volume = 3/100 mm^3, weight = 30 μg) operates in the self-heating mode, i.e., it is used to heat skin and to measure skin temperature simultaneously.

This probe is associated with an instrument system (electronic instrumentation and microcomputer) which controls the probe's operation and indicates the thermal conductivity of the medium under study.

3. Results

3.1 *Experiments on phantoms*

Five homogeneous phantoms with well-defined constant thermal conductivities were developed: three polyacrylamide gels (with water and ethylene glycol as solvents in different quantities), vaseline and polystyrene foam. The probe-instrument system calibration is performed in vaseline and in the polyacrylamide gel with water as solvent. The three other homogeneous phantoms are used to verify the probe-instrument system calibration.

A bilayer phantom has been designed to simulate the epidermis and the dermis. This phantom consists of the polyacrylamide gel with water as solvent, covered by one or several polyvinyl chloride sheets (PVC). The gel thermal conductivity is equivalent to that of the capillary network of the dermis and the PVC thermal conductivity is equivalent to that of the epidermis (nonperfused layer). Experiments performed on this phantom allowed the study of the micro-thermal field propagation relative to different parameters,[7] such as the thickness of the PVC layer (epidermis) and the duration of the active phase.

3.2 *In vivo results*

The influence of the application of a water-occlusive patch placed on one forearm for a period of 4 h was studied on two healthy subjects.

For each subject, measurements were performed on symmetric sites on both forearms at the beginning ($t0$) and at the end ($t4$h) of the experiment: right forearm, reference area and left forearm, water-occlusive patch area.

The results obtained from the two subjects are similar. The effects of the application of this water-occlusive patch are shown in Fig. 2. A large increase (1 mW/cm°C) in epidermal water content estimated by the measured thermal conductivities was observed but no modification of the skin microcirculation because Δ[water content] $\approx \Delta$[water content + microcirculation].

The relationship between the epidermal water content and the thermal conductivity (k) is given by:

$$k = 1/100 \times \{[\%\ \text{water content} \times 5.93] + [\%\ (\text{proteins} + \text{lipids}) \times 1.75]\} \quad (1)$$

and

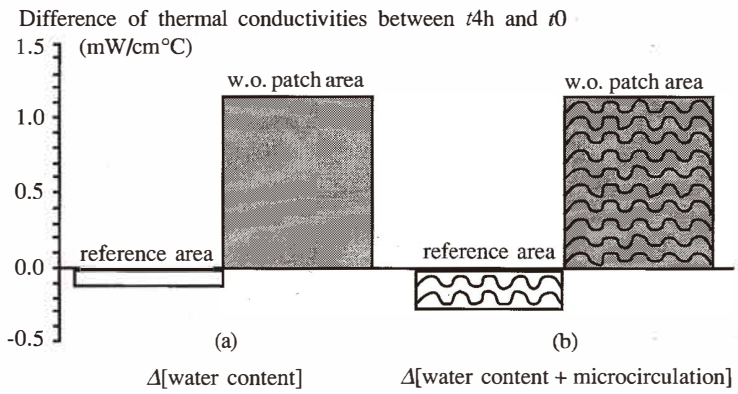

Fig. 2. Application, for a period of 4 h of a patch on the forearm of a 23-year-old healthy male subject.
w.o. patch area: area with the water-occlusive patch (left forearm)
reference area: area with no patch (right forearm)
Patch application induced an increase of the epidermal water content (a) but no modification of the microcirculation [(a) ≈ (b)].

$$\% \text{ water content} + \% \text{ (proteins + lipids)} = 100. \tag{2}$$

From eqs. (1) and (2):

$$\% \text{ water content} = (100 \times k - 175) / 4.18 \tag{3}$$

The increase of the epidermal thermal conductivity between $t0$ and $t4$h on the water-occlusive patch area corresponds to an increase of water content in the epidermis from 45% to 72% (Fig. 3).

4. Conclusion

The experimentation performed *in vivo* on human skin has proven the feasibility of the method and its practicability. The noise and drift amplitudes of the measurements are lower than those of physiological noise. Intra- and intersubject comparisons are therefore possible over short or long periods of time. This noninvasive sensor is very convenient for testing drugs and cosmetics, and performing physiological measurements in a laboratory.

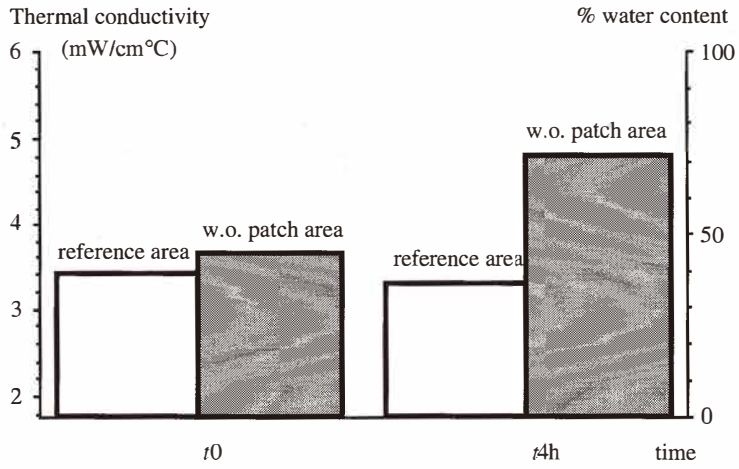

Fig. 3. Relationship between the measured epidermal thermal conductivities and the percentages of the epidermal water content (measurements on forearms of a 23-year-old male subject at *t*0 and *t*4h).
w.o. patch area: area with the water-occlusive patch (left forearm)
reference area: area with no patch (right forearm)

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