

Human Skin Temperature Response to Absorbed Thermal Power

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ABSTRACT -- Devices including ultrasound and magnetic resonance imaging probes can “overheat” and burn human skin unless they are carefully designed and tested. An empirical study was performed to determine how much thermal power the skin can absorb without raising skin temperature to the damage point. Steady-state power and temperature measurements were recorded from seven healthy adults. Small skin areas, 1.8 to 25 cm², were heated. The data indicates a “safe” absorption level of ~40 mW/cm². Near the overheating point, skin temperature increases ~0.8°C for each additional 10 mW/cm² of absorbed power.

Keywords: heating, human skin, thermal safety, absorbed power, temperature, ultrasound, MRI.

2. INTRODUCTION

During normal operation, medical devices such as ultrasound transducers, MRI surface coils, and fiberoptic illuminators are often heated as a side effect of performing their intended functions. Touching a heated device to human skin initiates a thermal transient followed by steady state. Although both the transient and steady state are important and can produce pain or skin burns, this paper focuses on steady-state considerations.

Today’s typical approach to burn prevention is simply a device-temperature limit. To verify thermal safety a designer places a finished device in ambient air, heats the device to steady state, measures device surface temperature and compares to a “known-safe” temperature like 41°C [2]. If measured temperature is less than 41°C, the designer concludes that the device will not cause pain or burning of the skin.

Although the above outlined method for checking thermal safety is easily understood and often

appropriate, several factors can limit its applicability. (1) When compared to human skin, the employed ambient air presumably provides a higher thermal resistance to heat moving from the tested device. The higher thermal resistance forces the device to reach a higher temperature than it reaches when contacting skin. Using ambient air as the thermal load likely produces conservative test results; but, if device performance improves with increasing power dissipation the test may be unjustifiably conservative. (2) No finished device is available for testing. (3) Device surface temperature is difficult to measure. (4) Expensive facilities may be required for the test. For example, certain surface coils used in magnetic resonance (MR) imaging are ideally tested while in an operating MR facility where test time is expensive.

In cases where the limitations to the “known safe” device-surface temperature approach are significant, skin-temperature response to absorbed power can provide an alternate route to ensuring thermal safety and/or choosing among design options. For example, consider a situation where the heat output for a given area of the device can be estimated from knowledge of the power supplied to the device and device surface area. If skin-temperature response is known, the heat output per area will allow estimation of the resulting skin temperature without calculating or measuring device temperatures.

A first understanding of skin temperature response to absorbed thermal power can be obtained with the aid of Figure 1. It outlines heat flow and temperature control under normal basal conditions. Bodily activities (mostly metabolism) release energy into the body core as heat, Q . An unshown control system closely regulates core temperature to a constant value, T_c , which is roughly uniform over the body. This temperature regulation works by modulating the flow of heat through the body’s “thermal insulation”

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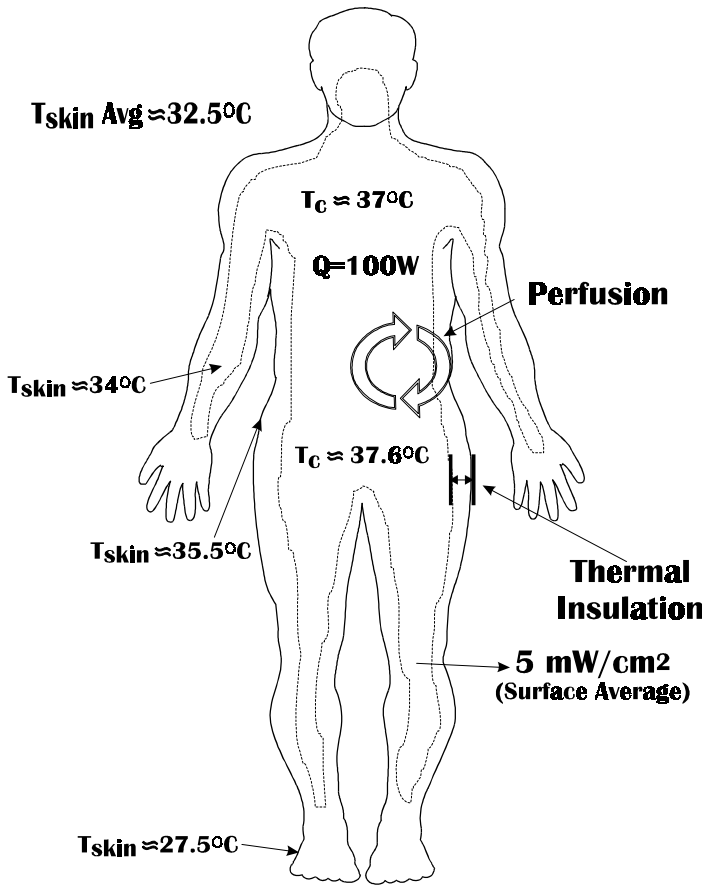


Figure 1. Schematic representation of temperature distribution and heat transfer for an adult human. At rest, metabolism releases about 100 watts into the body. The thermal insulation conducts some heat from the core and into the environment. Perfusion carries additional heat from the core and to the skin, $\sim 2 \text{ m}^2$, for release into the environment. Biological temperature sensing and feedback regulate perfusion rate to control core temperature, T_c . Both core and skin temperatures vary some with anatomical position [5, 7]. The dotted line represents the core region of the body.

(skin, subcutaneous tissue, and fat layer) and into the environment. The insulation value varies with location on the body but is stable over time. Blood flow or perfusion changes (for example with exercise or environmental temperature) and can modify heat transfer by $\sim 10X$. However, at comfortable room temperatures, humans are basically vasoconstricted and perfusion changes little with a few degrees of room temperature variation [4].

The preceding paragraph describes heat flow from the body core, through the skin and into the environment. The key variables are perfusion and thermal resistance of the insulating layer. When a medical device warms a small area of skin, we assume that (1) the core temperature remains unchanged, and (2) the perfusion and thermal resistance connect the heating device with the core so that perfusion and resistance determine the human-skin temperature response to absorbed power.

Young peoples' metabolism (watts/unit surface area) drops rapidly until approximately 20 years of age. After age 20, the metabolic-rate differences among people (young and old, male and female) is only $\sim 25\%$ [4].

Since adults have roughly the same core temperatures and metabolic rates, they must also have the same thermal power density passing through their insulating layers. Hence, we anticipate roughly the same average skin-temperature response for different adults. However, response could vary with position on the body (caused for example by variation of fat layer thickness). The response could also vary with heated-device surface area. These points are empirically explored herein.

In transient conditions the materials from which the device is constructed contribute significantly to the amount of heat transferred to touched skin. However, this paper focuses on thermal steady-state and material choices can be ignored.

This paper describes an empirical study of skin-temperature response; we find that human skin can safely absorb up to $\sim 40 \text{ mW/cm}^2$. This facilitates design and/or testing for thermal safety in situations where the established temperature-limit method is limited.

3. METHOD

Two heat gauges have been built for the purpose of measuring the steady-state power required to bring human skin to measured temperatures. Each gauge consists of an electrical heater, a heated aluminum block with an exposed surface for skin contact, a

thermal insulator to cover that portion of the block that does not contact skin, and a thermocouple for monitoring the block temperature; see Figure 2. A thin layer of silicone heat sink compound helps ensure good thermal coupling between the block and skin. We assume that block temperature equals skin temperature. The exposed surface areas are $\sim 1.8 \text{ cm}^2$ and $\sim 25 \text{ cm}^2$ for the two gauges to allow exploration of the importance of surface area.

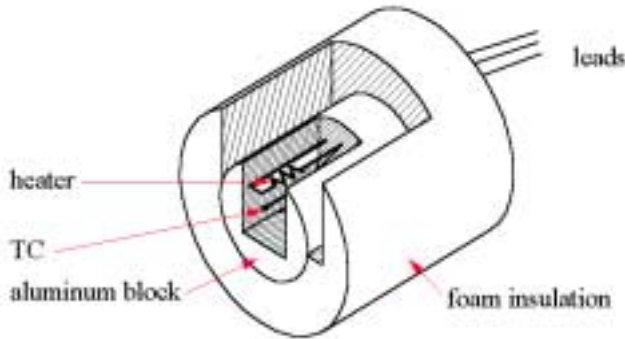


Figure 2. Cut-away view of heat gauge. Electrical power heats resistors which subsequently heat an aluminum block that connects with human skin. A thermocouple senses block temperature which is assumed equal to skin temperature at steady state. Polymer foam insulation minimizes the flow of “leakage” heat from the block and into the environment. Thermal leakage resistance was measured as $155^\circ\text{C}/\text{watt}$ and $80^\circ\text{C}/\text{watt}$ for the small and large heaters respectively. Drawing not to scale. TC - denotes thermo-couple.

In use, a heat gauge is placed on the skin and a set amount of electrical power applied to the heater until steady state is reached. The block temperature and electrical power are recorded. A calibration procedure establishes thermal resistance from the block, through the insulator and to ambient air. Thermal power leakage is computed from this resistance, the measured block temperature, and air temperature. Reference [10] provides additional details.

The heat gauges were used to collect power (watts/cm^2) and temperature ($^\circ\text{C}$) measurements from the abdomen, the calf of the leg, and the under side of the forearms of several volunteers. An hour or more was allowed for the gauge and subject to

reach thermal steady state at each measurement point. We did not collect every measurement from every subject and most of the measurements were collected from the abdomen and forearm.

The seven volunteer subjects included two females and five males. Ages ranged from 23 to 58 years; weights ranged from 123 to 230 pounds; and heights ranged from 5’7” to 6’0”. All volunteers are working adults in good health. Tests were conducted in an air conditioned office-like setting with air temperature between 21°C and 25°C (70° to 78°F). During abdominal measurements, the subject laid face-up on a bed with the heat gauge stationary on the abdomen. Data from the arm and leg were collected with the subject seated comfortably. On a day when measurements were taken, subjects engaged in normal work activities but avoided strenuous activity which might have altered the metabolic rate.

To make data collection faster, we turned on the heat-gauge electrical power and raised the heater block temperature to roughly 40°C before starting to collect measurements. This allowed us to find steady-state temperature within one or two hours after putting the gauge on the skin. Steady state was declared when the skin temperature changed $\leq \pm 0.1^\circ\text{C}$ for 20 minutes. After measurements were collected for a given power setting, the heat gauge was removed from the skin and the heater contact area visually inspected. Data was discarded unless the silicone grease left a cleanly-defined round circle on the skin. The round circle was taken to mean that the heater had remained stationary while the temperatures were read.

4. RESULTS

Figure 3 shows measured skin temperature versus absorbed power for one subject; the data was collected with the small heater placed on the abdomen. The straight line is a least-squares fit to the collected data points; over the narrow temperature range of interest ($\sim 10^\circ\text{C}$) a straight line provides a reasonable representation of the data. Over an hour or two, five or ten time-temperature pairs were recorded to arrive at one of the plotted steady-state data points shown in Figure 3. Thus, 30 to 60 time-temperature readings were taken to find the line shown. For simplicity of discussion, we define a least-squares fitted straight line as shown in

Figure 3 to be a “data line”. For compactness of notation, we can now specify a data line with one point on the line and the slope of the line. Since 41°C is generally accepted as a safe device temperature, we chose 41°C and the absorbed power required to raise skin temperature to 41°C (q_{T41}) as our defining point. For the Figure 3 data line $q_{T41} = 0.082$ watts/cm², and slope $m = 51.7^{\circ}\text{C}\text{-cm}^2/\text{watt}$. After collecting measurements for the first few data lines, we saw that the data points deviated only slightly from the data line. Because of this and the time required to take measurements, we soon began collecting data at a low temperature ($\leq 35^{\circ}\text{C}$),

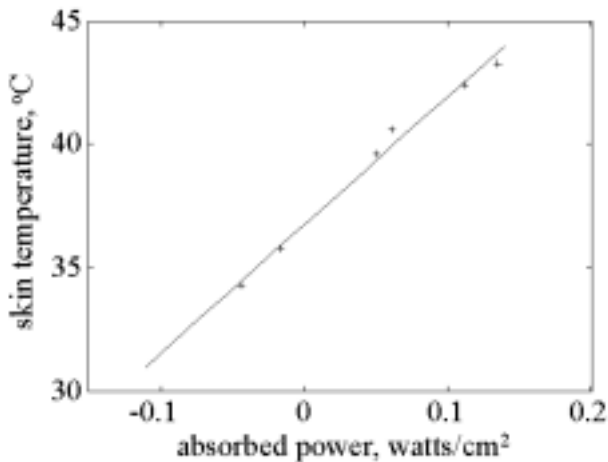


Figure 3. Abdominal skin temperature response to absorbed power. Experimental data points collected using the small heat gauge. Straight line obtained by least-squares fit to the indicated, +, data points.

a high temperature ($\geq 40^{\circ}\text{C}$), and one intermediate temperature.

Table 1 lists the calculated q_{T41} and m (slope) for each data line collected in the study. The arm and torso represent large and small body parts. Data lines from three legs were included to broaden the range of anatomical parts tested. The numerical values in Table 1 may facilitate comparison to data collected by other investigators in the future.

As an approximate way of representing all the data, we average the q_{T41} values and the m values to obtain:

$$q_{T41} \sim 0.078 \text{ W/cm}^2$$

$$m \sim 81.5^{\circ}\text{C}\text{-cm}^2/\text{watt}$$

In other words, on average the skin absorbs about 80 mW/cm² to reach 41°C. In addition, skin temperature increases by 0.8°C for each 10 mW/cm² of absorbed power.

Table 1 summarizes the q_{T41} and m values for all conducted experiments. However, the data spread may be easier to see when the data lines are plotted as in Figure 4. The q_{T41} variation is about +/-50% of average. The variation covers the range of subjects, two heater sizes, and three anatomical sites.

To obtain a sense of the skin-temperature response

Table 1. Summary of experimental results from all subjects.

	Arm		Leg		Torso	
	q_{T41} (w/cm ²)	m (°C-cm ² /w)	q_{T41} (w/cm ²)	m (°C-cm ² /w)	q_{T41} (w/cm ²)	m (°C-cm ² /w)
Small Heater	0.120	59.3	0.0961	77.7	0.088	52.0
	0.0939	74.4	0.0712	79.9	0.0765	61.8
	0.0847	77.1			0.0823	51.7
	0.100	80.0			0.0566	62.8
	0.138	65.3				
	0.0892	77.2				
Large Heater	0.0729	103	0.0581	119	0.0614	94.5
	0.0453	132			0.0488	98.8
	0.0665	101			0.0661	96.9
	0.0719	78.0			0.0600	69.6

variation with anatomical site and heater-block size, consider Figures 5 and 6. Figure 5 contrasts the data lines obtained with the small gauge on the torso to the lines obtained with the large gauge on the arm. We might expect that the relative slopes associated with the heaters would tend to reverse if we switch the heater locations. Figure 6 shows that such a slope reversal appears to exist; however, our data set is too small to warrant forceful arguments on this point.

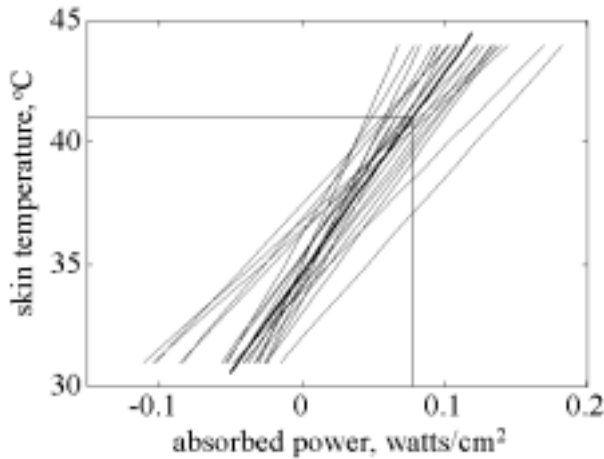


Figure 4. Data lines from seven subjects, two sizes of heat gauge and three anatomical sites. The bolder line is the average data line. The vertical and horizontal lines mark 0.078 W/cm² and 41°C, respectively.

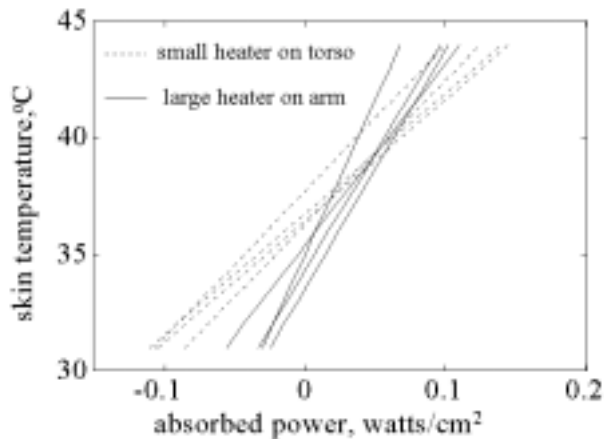


Figure 5. Dashed data lines show torso-temperature responses measured with the small heat gauge. Solid data lines indicate arm-temperature responses measured with the large heat gauge.

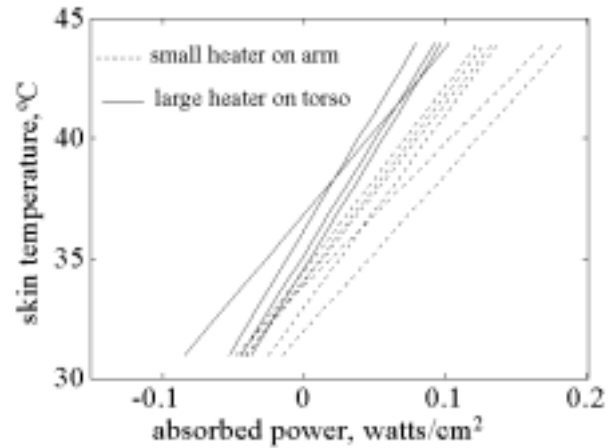


Figure 6. Dashed data lines show arm-temperature responses found with small gauge. The solid data lines are torso-temperature responses seen when using the large gauge.

Of the measurement arrangements employed, large-heater on the arm exhibits the highest slope, i.e., most temperature increase per absorbed watt/cm².

5. DISCUSSION

Data reported in this empirical study provides certain design guidance for medical devices that contact human skin. When the designer has an estimate of the thermal power that will be conducted to the skin, a minimum required surface area is readily calculated as:

$$\text{required area} = \text{thermal power} / q_{T41}$$

If this criteria is not met, one must find a way to either decrease thermal energy dissipation in the device or redirect the flow of energy away from the human. Consider an example device which conducts 1 watt into the skin. This device must have a 12.5 cm² area (for example, a 4 cm diameter circle) contacting the skin. Unless the contact area is uniformly heated, the contact area must be somewhat larger to prevent an area of excessive power density.

In the preceding calculation, we used q_{T41} to find required device surface area; but, no physiologic event is likely to happen at exactly 41°C. For example, 40°C or 43°C might be acceptable “safe”

temperatures. Also, we are using the average q_{T41} rather than the worst case which we do not precisely know. The point is that some safety factor must be chosen. With presently available data, it seems reasonable to use a safety factor of about 2 and assume that human skin can absorb at least 40 mW/cm² without pain or injury.

The data lines reported herein come from normal volunteers (mostly authors of the article). Since adults have similar metabolic rates and core temperatures, we expect modest q_{T41} differences between the volunteers. The experimental data meets this expectation. However, our knowledge will be extended when data from sick and elderly patients can be gathered and published.

Heater blocks of ~1.8 cm² and ~25 cm² skin-contact area were employed in the heat gauges. Results varied with area, but only modestly. Thus, our q_{T41} values can likely be interpolated to design problems wherein the device covers a small fraction of the body surface area. On the other hand, if a device covered and heated a complete limb, for example, the relevant q_{T41} might be considerably different from our value.

Data in this paper describes steady-state skin temperature response to absorbed power. For short time periods, i.e., seconds, the skin can safely absorb much higher levels of thermal power [1]. As previously shown [10], steady-state data can be combined with transient data to produce a more generally applicable model of skin heating. It may also be illuminating to combine the data from this empirical study with a mechanistic model such as the bio-heat transfer equation [6, 8, 9].

In summary, devices that heat small areas of human skin can safely emit about 40 mW/cm² of thermal power.

6. ACKNOWLEDGMENT

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7. REFERENCES

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