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Review article

Deep temperature monitoring using a zero-heat-flow method

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Introduction

The historical description of the measurement of body temperature dates back to Hippocrates (400BC), who put wet clay on the patient's body surface and then cut the abscess on which the clay first dried up [1]. Temperature scales were established by Celsius (°C) and Fahrenheit (°F) in the 18th century. The mercury thermometer, which was invented by Ehrle in 1866 [2], is now used clinically throughout the world because of its accuracy, serviceability, low cost, and ease of use. However, the mercury thermometer has some disadvantages: continuous measurement of body temperature is impossible, rapid temperature change cannot be monitored, and there is a risk of mercury pollution. Advanced equipment for the measurement of body temperature has been invented and used for basic research and medical examination and treatment since the 1960s [3].

Thermistors are now used by many anesthesiologists to measure the rectal temperature during an operation. Rectal temperature measured by a thermistor is usually an accurate indicator of body core temperature during an operation [4,5]. However, in operations such as cardiac surgery with cardiopulmonary bypass (CPB), in which the body temperature changes quickly, rectal temperature is not a reliable indicator of body core temperature [6,7]. Another disadvantage of measurement of rectal temperature with a thermistor is the risk of perforation of the rectum by the probe, especially in children [8].

In 1971, Fox and Solman [9,10] invented a noninvasive deep temperature thermometer using a zero-heat-flow method. This thermometer enables measurement of the deep body temperature indirectly from the intact skin surface. The technique of deep temperature monitoring using this thermometer has been improved, mainly by Togawa's group and Terumo Co. [11,12]. Monitoring of deep body temperature, especially from the forehead, is now widely used in cardiac surgery in Japan. Deep temperature monitoring has also been used in intensive care units [13] and for monitoring of circulatory failure [14]. The equipment used for deep temperature monitoring has recently been improved, and the disadvantages of the conventional device have been overcome.

Here, we review the principle of deep body temperature measurement, its characteristic features and clinical applications, and improvement in the equipment used for measurement.

Principle of measurement

It is important to measure the so-called core temperature (deep temperature), because important organs, such as the brain, maintain the core temperature precisely at around 37°C in a normal situation. The methods generally used for measuring deep body temperature in humans involve direct insertion of a thermistor probe into a natural body orifice. Thermometers require pockets into which their probes can be inserted, but there are no such suitable pockets in the human body. Certain sites, e g., the mouth, esophagus, rectum, nasal cavity, and ear, are presently used for deep temperature measurement, but placement of the thermometer probe in these sites causes discomfort to the patient and is unsatisfactory for long-term use,

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especially in awake patients. Furthermore, these measurements can induce complications [8,15,16] and show errors [6,7,17,18]. Intravascular catheter thermometers are also not generally suitable, because of the invasive nature of their application.

The indirect measurement of deep tissue temperatures from the skin surface has, therefore, been considered a desirable method for some time. One of the major problems encountered in noninvasive measurement of deep body temperature is related to the thermal properties of the skin. The thermal conductivity of the skin is generally poor and is strongly influenced by the skin blood flow [19]. The most widely used technique to prevent heat loss from the skin is thermal insulation. If an ideal thermal insulating pad is applied to the skin surface, heat loss from this area will be reduced to zero, and after a while the skin surface temperature beneath the pad will be equilibrated to the deep tissue temperature. However, ideal thermal insulators have not yet been developed.

Fox and Solman [9] first provided a principle of an electronic servocontrolled system to achieve almost complete thermal insulation. A schematic drawing of the first reported servocontrolled thermal insulation technique is shown in Fig. 1. The temperature-measuring probe (a flat square probe, 60 × 60 × 6 mm) has two thermistors separated by a thermal insulator, with an electrical heating element mounted at the rear of the probe. The temperatures on the two sides of the insulating layer, as detected by the thermistors, are compared by a differential amplifier. The error signal is used to control the heater current in such a way as to achieve a situation in which no temperature gradient arises across the insulating layer, and thus no heat flows out through this layer. This technique is called the zero-heat-flow method. As long as a zero-heat-flow condition is maintained, the probe is equivalent to an ideal thermal insulator, i.e., heat loss from the skin surface beneath the probe is prevented, and, after a sufficient time, the skin surface temperature will equilibrate with the deep tissue temperature. The skin surface temperature may then be measured by the lower thermistor in contact with the skin.

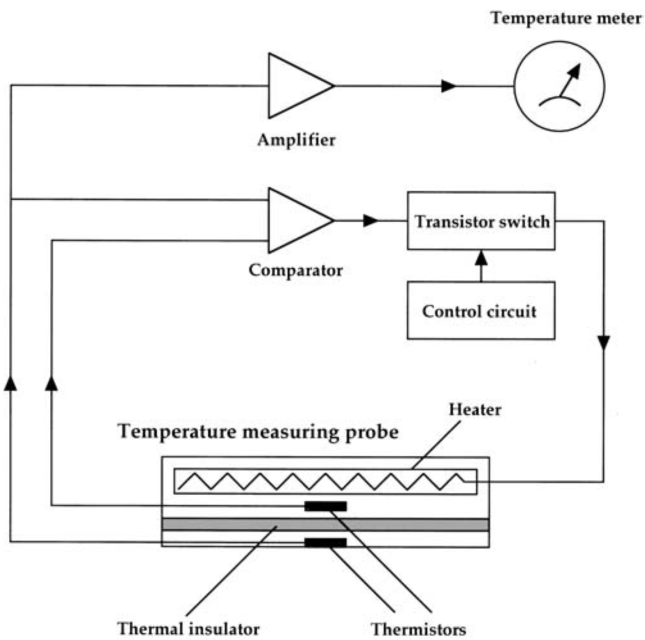

Fig. 1. Schematic diagrams of the control circuit and cross-section of the probe of the deep body thermometer. The system, originally designed by Fox and Salmon [9] (Deep Thermometers, Little Eversden, Cambridge, UK), has a flat square probe 60 × 60 × 6 mm in size. The probe contains two closely matched thermistors at the center, a piece of nylon gauze separating them, and a thin-film electric heater element. The components are encapsulated in clinical-grade silicone rubber using a single vulcanizing process

Characteristic features

Initial response time

The initial response time of a deep temperature thermometer, measured from the time when the probe is placed on the body surface to the time when the measured temperature becomes stable, seems to be long. The duration partly depends on the initial temperature of the body surface and the blood flow rate of the tissue. Figure 2 shows changes in temperatures measured by the probes of deep temperature thermometers placed on the forehead, chest, sternum, and abdomen. It takes about 15 min to obtain the final equilibrium temperature within 0.1°C.

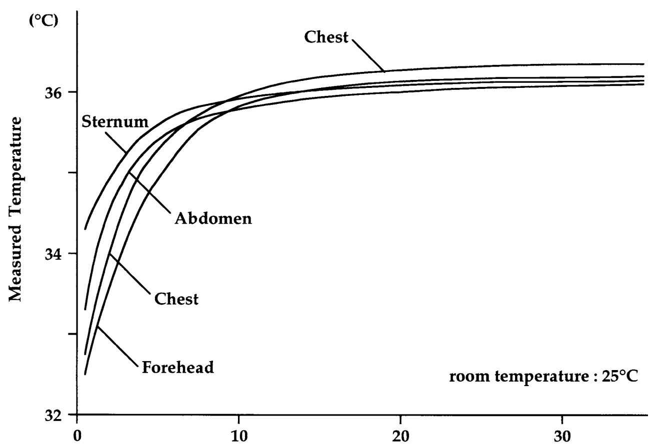

Fig. 2. Representative changes in temperature measured by the probes of deep temperature thermometers placed on the forehead, chest, sternum, and abdomen. Approximately 15 min is required to obtain the final equilibrium temperature within 0.1°C

The initial response time can be shortened by prewarming the probe. However, even with prewarming of the probe, it still takes a long time to obtain the equilibrium temperature when the body surface is cool and the blood flow rate and heat conductivity in the tissue are both low.

Response time for internal temperature changes

The initial response time does not seem to correlate with the response time for internal temperature changes. The response time of the deep body thermometer is still longer than that of conventional catheter thermometers, e.g., a catheter thermometer for measurement of rectal temperature. However, under normal physiological conditions, the body has a large thermal inertia, and internal temperature changes that cannot be monitored accurately by using a deep body thermometer do not occur so rapidly. Fox et al. [10] reported an experiment in which fever was induced by means of an intravenous injection of endogenous pyrogen in an air-conditioned room at 28°C. The thermometer probe was fixed on the upper sternum of the subject. The auditory canal temperature and the intestinal temperature measured by a radio pill were recorded simultaneously (Fig. 3). The sternum deep body temperature was not delayed in comparison with the other internal temperatures, even at the onset of the fever.

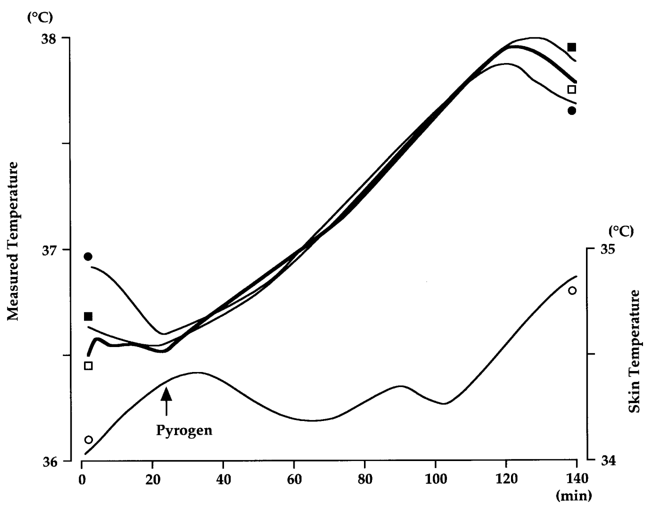

Fig. 3. Internal and skin temperatures during the induction of a pyrogen-induced fever. *Filled circle*, intestinal temperature measured by a temperature-sensitive radio pill; *filled square*, auditory canal temperature; *open square*, upper sternum deep body temperature; *open circle*, skin temperature. There seems to be no delay in measurement of the sternum deep body temperature compared with the other internal temperatures, even at the onset of the fever (redrawn from Togawa et al. [11] with permission)

On the other hand, in patients undergoing cardiac surgery with CPB [20] or with malignant hyperthermia, in which the body temperature can change very quickly (at a rate of more than $1.0°C \cdot 30\,min^{-1}$), deep temperature monitoring does not seem to be as reliable as measurement of the blood temperature from the CPB or jugular vein temperature.

Ambient temperature

An important feature of the deep body thermometer with servocontrolled heating is that ambient temperature changes do not affect the temperature sensor mounted in the probe. When the probe is attached to the head or torso of a human subject, the observed temperature is never affected by ambient temperature changes. This was confirmed by an experiment [21] using a probe 50 mm in diameter with a guard and a skin temperature probe attached to the side of the head and forehead. The observed deep temperature remained almost constant with a 20°C change in ambient temperature, while the forehead skin temperature changed significantly.

Thickness of tissue

The thickness of the tissue between the skin surface and the deep temperature region, across which a temperature gradient is developed, influences both the accuracy and the response time of deep temperature measurement. Although there is no clear boundary between the temperature gradient layer and the uniform deep temperature region in natural tissue, a simplified model of a core-and-shell structure is helpful in understanding the operating depth of the probe. In the model, the shell corresponds to a layer, including the skin and subcutaneous fat, in which a relatively large temperature gradient develops, and the core corresponds to deep tissue having an almost uniform temperature.

On the assumption that the core-and-shell model is applicable, the accuracy and response characteristics of the thermometer were evaluated by a simple model experiment using a structure with a rubber shell [22]. In the experiment, the thermal conductivity of the rubber was approximately $1.7 \times 10^{-3}\,J \cdot cm^{-1} \cdot °C^{-1}$, which is similar to the thermal conductivity of unperfused tissue. The deep body thermometer probe was placed on the rubber shell. An increase in the shell thickness to more than 9 mm caused significant increases in both the error and the response time (Fig. 4). As long as the thermal properties of the skin and subcutaneous tissue do not differ significantly from those of the material used in the model experiment, it is reasonable to assume a similar relationship to that shown by the results of this experiment. The implication of this is that accuracy will be achieved if the response time is comparatively short,

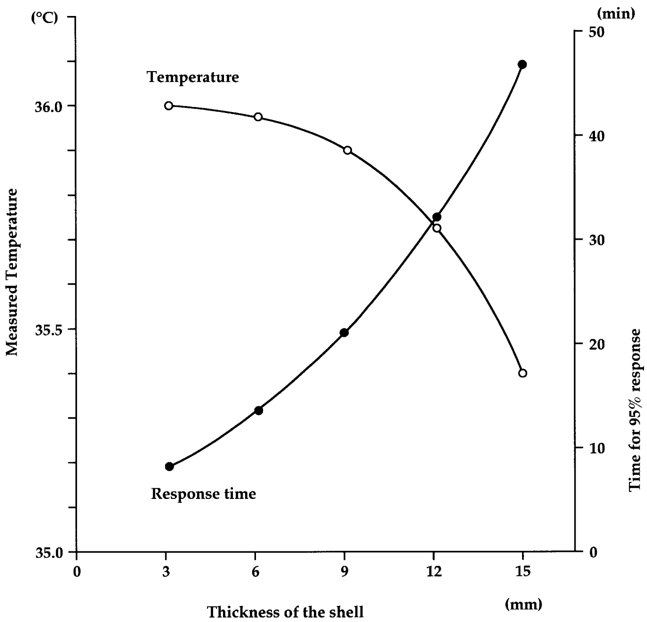

Fig. 4. Temperature at equilibrium and time required for 95% response for the shells of different thickness in core-and-shell models. Increase in the thickness of the shell to more than 9 mm caused significant increases in both the error and response time (redrawn from Togawa et al. [21] with permission)

assuming that the same-sized probe is used. When the probe is attached to the head, torso, or limbs of a human subject, the response time is usually 10–20 min. This corresponds to the case of a 9-mm-thick rubber sheet in the model experiment, and hence the same accuracy may be expected in human use.

Clinical applications

Core temperature monitoring

The deep body temperature measurement technique with servocontrolled heating was initially developed for the measurement of core temperature. It has the advantage of being noninvasive and therefore is less of an encumbrance to the subject than are conventional aural or rectal temperature measurement techniques.

For long-term clinical monitoring, the site for attachment of the probe should be standardized. Fox and Solman [9] recommended the upper sternum because it is relatively flat, has very little subcutaneous fat, and is near the large vessels of the central blood supply, which would be equilibrated with the core temperature. Ball et al. [23] compared the deep body temperatures measured by probes attached to the sternum with rectal and external auditory canal temperatures in 15 patients aged 13–75 years. The measurements of the deep body temperature were very similar to those of rectal temperature, remaining only approximately 0.2°–0.4°C below the rectal temperature reading. The measurements of the deep body temperature were also very similar (usually within 0.2°C) to those of the auditory canal temperature.

Deep body temperature measured by a probe attached to the occipital region is always higher than that measured by probes attached to other sites, even when the probe is attached to the hair. In a comparative study of rectal temperature in 15 subjects, the mean occipital reading was found to be 36.8°C [11]. The occipital region is considered a good site for core temperature monitoring, because it is close to the superior sagittal sinus, which accepts blood from the brain and includes the center for physiological temperature regulation. The probe is easily fixed on the occipital region by a hair band, and the hair does not interfere with the measurements.

The forehead is also a convenient site for probe attachment, and a study in 17 postoperative patients showed that the forehead reading remained only 0.9° ± 0.4°C below the rectal temperature. The forehead reading was also compared with blood temperature in the internal jugular vein measured with a catheter thermometer in seven patients [23]. The forehead reading was 37.8° ± 0.5°C, whereas the blood temperature was 37.7° ± 0.4°C.

In adult subjects, the abdominal region is considered an inappropriate site for core temperature monitoring because of the significant thickness of subcutaneous fat. However, this site may be used with success in infants and children. A study in 24 subjects aged 3–15 years showed that the abdominal and forehead deep temperature readings were only 0.04° ± 0.2°C and 0.2° ± 0.2°C below the rectal temperature, respectively. The abdominal deep temperature reading obtained with a probe 100 mm in diameter was only 0.2° ± 0.2°C below the rectal temperature in 22 patients.

Monitoring the blood flow of the skin

The technique of deep body temperature measurement can also be used in the field of dermatology. For example, patients with Bürger disease show higher peripheral deep temperature than do healthy volunteers. Some groups [24] have shown that this technique seems to be useful for monitoring the effectiveness of steroids in patients with skin diseases. The technique of deep body temperature measurement has also been used for monitoring appropriate blood flow in a postoperative skin flap.

Hemodynamic monitoring

The forehead, palm, and heel can be used as standard deep temperature recording sites after cardiac surgery.

Approximately 150 patients who had undergone cardiac surgery during the period from 1974 to 1976 were monitored for 10 to 270h by deep temperature measurements [23]. Forehead, palm, and heel temperatures were successfully recorded throughout the period, whereas rectal temperature recording was interrupted several times in most of the patients. In some patients who went into circulatory insufficiently, significant differences between forehead and limb deep body temperatures were observed in advance of a fall in blood pressure. Tsuji's group [25] recommended the use of deep body temperature monitoring by probes attached to the forehead and a limb for diagnosis of shock and low output syndrome (LOS). Findings that seem to indicate circulatory insufficiency are: less than 90mmHg of systolic blood pressure; over 7°C of dissociation between forehead and limb deep temperatures; and less than $1\,ml \cdot kg^{-1} \cdot h^{-1}$ of urinary output. However, care must be taken with patients who have endotoxin shock (warm shock) and patients receiving strong vasodilatory agents, because these patients do not show a dissociation between core and peripheral deep body temperatures.

Improvement in equipment

The application of a thermal insulator over a large area of the skin surface will be more effective than that over a small area in bringing the skin surface temperature close to that of the deep tissue. The disadvantage is that the probe is more cumbersome to use, and because dissipation of heat from the skin is radically changed by this measurement technique, it may actually affect the deep tissue temperature. In the case of a small probe, the temperature gradient between the center and the circumference of the probe may be significant, and the subsequent heat flow in the radial direction is not compensated by the servocontrolled heater within the probe.

To prevent radial heat flow, the use of a thick metal guard covering the probe, as shown in Fig. 5A, was proposed by Togawa et al. [11,21]. The guard makes contact with the skin surface at the circumference of the probe and tends to minimize radial heat flow. Kobayashi et al. [26] compared the results obtained by using a guard-type probe with those obtained by using a probe of the original design. The temperature distribution on the skin surface beneath the probe was calculated and measured experimentally, and the results indicated a significant reduction in the radial temperature gradient with the use of a guard-type probe. In an animal experiment, a rubber bladder with two extension tubes was inserted into the abdominal cavity of an anesthetized dog, and warm water was circulated through the bladder. The deep body thermometer probe was

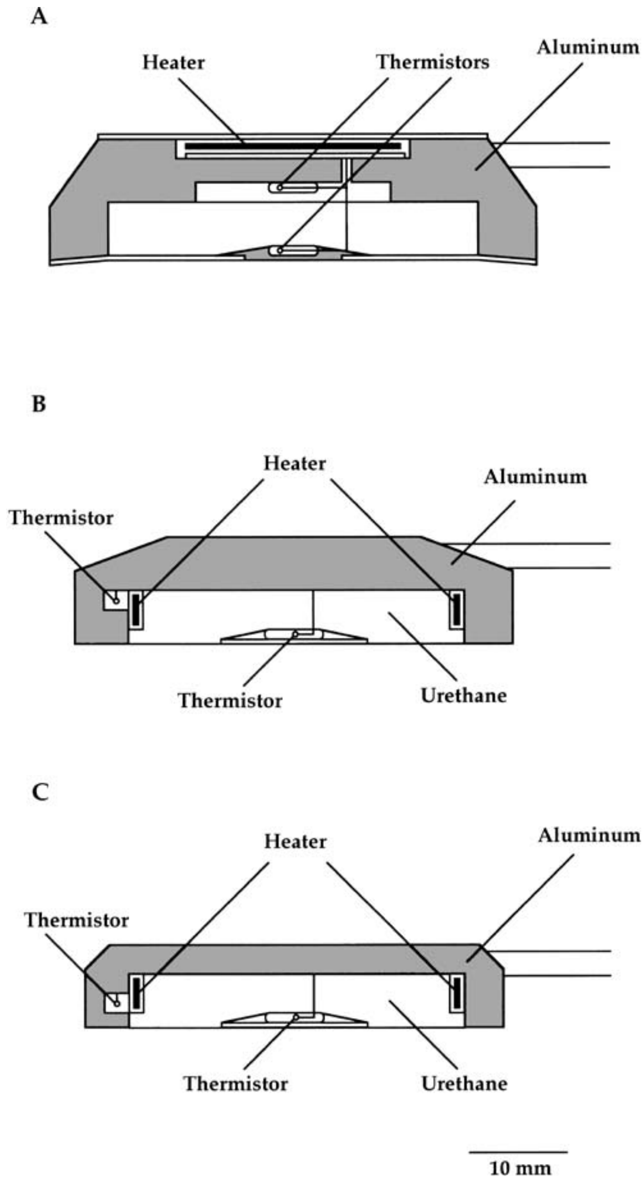

Fig. 5. Cross-sectional views of some improved probes. **A** To prevent radial heat flow, the probe is covered by a thick metal guard [11,21]. A thermal switch for protection against overheating is attached to the head. **B** An electric heater strip is attached to the entire inner circumference. A control thermistor is attached to a part of the electric heater strip. **C** For monitoring rapid deep body temperature change, the heat capacity, weight, and thickness of the probe have been reduced

placed on the abdominal wall immediately above the bladder. The temperature measured by the probe with a guard corresponded closely to that of the circulating water, but the temperature measured by the probe without a guard was 0.3°–0.9°C below the water temperature of 34.6°–41.6°C.

An illustration of the modified probe with a guard is shown in Fig. 5B. The standard probe is 45mm in diameter and 13.5 mm in thickness, and additional probes with diameters of 25 mm and 80 mm are also available. The contents of the probe are assembled in a thick aluminum cup. An electric heater strip is attached to the entire inner circumference of the aluminum cup to measure the deep body temperature more accurately. A control thermistor is attached to a part of the electric heater strip on the assumption that the temperature of the aluminum cup is maintained uniformly. The measurement thermistor is placed at the center of the probe suspended on soft urethane foam rubber. The face of the probe that is in contact with the skin is covered by a sheet of polyvinyl chloride. All probes are calibrated with an accuracy of 0.1°C (at 30°–40°C) and are interchangeable. The probe is attached to the patient with either elastic adhesive tape or double-sided adhesive tape. This type of thermometer (CTM-205; Terumo, Tokyo, Japan; Fig. 6A) has been widely used for deep temperature monitoring during surgery, especially cardiac surgery with CPB.

In a hot environment or in subjects with hypothermia, the ambient temperature is higher than the deep body temperature, and servocontrolled heating is not able to achieve a zero temperature gradient. In order to

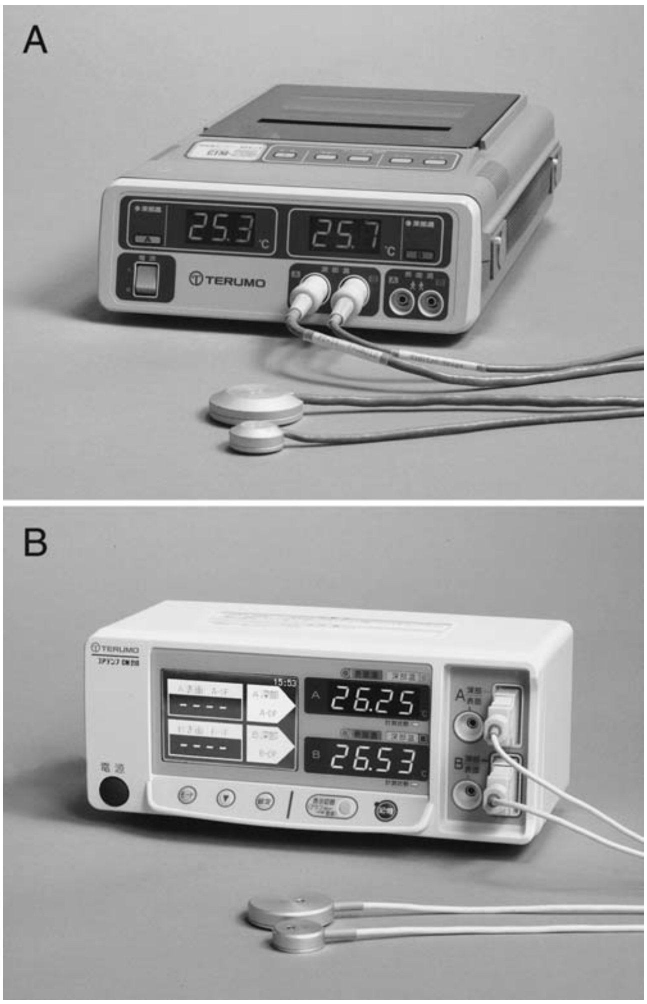

Fig. 6. Photographs of (**A**) a conventional type (CTM-205) and (**B**) a newly developed (CM-210) deep body temperature monitoring system (both made by Terumo, Tokyo, Japan). The main unit of the newly developed system CM-210 has been reduced in size and can be connected to an external vital monitor. The unit also has the color LCD that displays the temperatures measured by four probes and can store data internally for 2 days

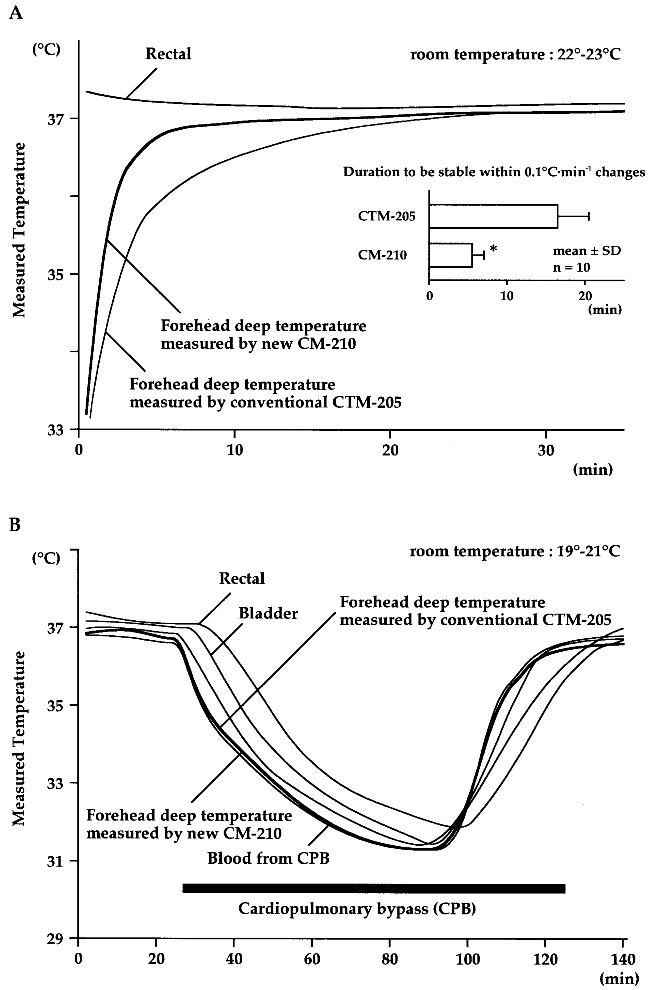

Fig. 7. Results of a study comparing the newly developed deep temperature monitor (CM-210) with a conventional one (CTM-205) [29]. **A** Initial time responses. The initial time response of the new monitor was much shorter [5.6 ± 1.2 min (mean ± SD), time within $0.1°C \cdot min^{-1}$ changes] than that of the conventional one (16.5 ± 4.2 min) (**inset**, n = 10, * $P < 0.05$ vs. conventional one). **B** Representative data of forehead deep temperature monitoring in cardiac surgery with cardiopulmonary bypass (CPB). The temperature measured by the new system was closer to the blood temperature from the CPB (r = 0.92) than was either the bladder temperature (r = 0.81) or the deep temperature measured with the conventional system (r = 0.85)

Table 1. Improved features of the newly developed deep temperature monitoring system (CM-210, Terumo, Tokyo, Japan)

Operability	
	Downsized main unit
	Color LCD displays temperatures measured by all the probes
	Connectable to external vital monitor
	Internal data storage function (1 reading·min^{-1} for 48 h)
Measurement of deep body temperature	
	Reduced heat capacity, weight, and thickness of the probes
	Adoption of initial intensive heating ("quick start")
	Display of temperature in digits of 0.01°C
Safety	
	Suitable for EMC standard IEC 60601-2-1, 1993 to reduce electromagnetic noise
	Lock-type sensor probe connector
	Increased resistance of cable against alcohol
	Improvement in detection of deteriorated wires

achieve a zero temperature gradient under these circumstances, the back of the probe must be cooled. This may be achieved by using a thermo module (a semiconductor electric cooling element), and Solman and Dalton [27] recommended that this technique be used in physiological studies of glass and steel workers and for newborn infants being nursed in incubators. This technique could also be used for body temperature monitoring during deep hypothermia. Tamura et al. [28] described a system using a thermo module for measuring blood temperature in CPB circuits. This system enables measurement of blood temperature from the outside of the tubing, even when it falls below ambient temperature.

Recently, Terumo Co. has again improved the equipment used for deep temperature monitoring (CM-210, Terumo). Figures 5C and 6B show a cross-sectional schematic view of the probe and a photograph of the new deep temperature monitor, respectively. The three main improvements in the monitor are an improvement in operability, a wider range of clinical application, and improved safety (Table 1). The reliability of the monitor seems to have been improved by the use of initial intensive heating ("quick start") and by a reduction in the diameter of the probe to 43 mm. We have tried to use this new monitor with the probe on the forehead in minor surgery under general anesthesia and in cardiac surgery with CPB [29]. As shown in Fig. 7, the initial time response was much shorter than that with conventional equipment, and the measured temperature was closer to the blood temperature from the CPB than was either the bladder temperature or the conventionally measured deep temperature. This new type of deep temperature monitor is therefore expected to be used more widely in various fields of medical care.

Conclusions

Fox and Solman [9,10] invented a noninvasive deep temperature thermometer using a zero-heat-flow method. This thermometer enables measurement of the deep body temperature indirectly from the intact skin surface. Although a conventional type of deep body temperature monitoring (CTM-205, Terumo) seems to be used at present only in the field of cardiac surgery because of its high cost and relatively slow response, it is expected that advanced deep temperature monitoring (CM-210, Terumo) that has a quick response will be used in other kinds of surgery and anesthesia in the future.

References

1. Otsuka K, Togawa T (1997) Hippocratic thermography. Physiol Meas 18:227–232
2. Ebstein E (1928) Die Entwicklung der klinischen Thermometrie. Erg Inn Med 33:407–503
3. Togawa T (1985) Body temperature measurement. Clin Phys Physiol Meas 6:83–108
4. Eichna LW, Berger AR, Rader B, Becker WH (1951) Comparison of intracardiac and intravascular temperatures with rectal temperatures in man. J Clin Invest 30:353–359
5. Ilsley AH, Rutten AJ, Runciman WB (1983) An evaluation of body temperature measurement. Anaesth Intens Care 11:31–39
6. Cooper KE, Kenyon JR (1957) A comparison of temperatures measured in the rectum, oesophagus, and on the surface of the aorta during hypothermia in man. Br J Surg 44:616–619
7. Molnar GW, Read RC (1974) Studies during open-heart surgery on the special characteristics of rectal temperature. J Appl Physiol 36:333–336
8. Vale RJ (1981) Monitoring temperature during anesthesia. Int Anesthesiol Clin 19:61–83
9. Fox RH, Solman AJ (1971) A new technique for monitoring the deep body temperature in man from the intact skin surface. J Physiol 212:8–10
10. Fox RH, Solman AJ, Isaacs R, Fry AJ, MacDonald IC (1973) A new method for monitoring deep body temperature from the skin surface. Clin Sci 44:81–86
11. Togawa T, Nemoto T, Yamazaki T, Kobayashi T (1976) A modified internal temperature measurement device. Med Biol Eng 14:361–364
12. Togawa T (1979) Non-invasive deep body temperature measurement. In: Rolfe P (ed) Non-invasive physiological measurements. Vol 1. Academic Press, London, pp 261–277
13. Togawa T, Nemoto T, Tsuji T, Suma K (1979) Deep temperature monitoring in intensive care. Resuscitation 7:53–57
14. Tsuji T (1987) Patient monitoring during and after open heart surgery by an improved deep body thermometer. Med Prog Tech 12:25–38
15. Tabor MW, Blaho DM, Schriver WR (1981) Tympanic membrane perforation: complication of tympanic thermometry during general anesthesia. Oral Surg 51:581–583
16. Wallace CT, Marks WE, Adkins WY, Mahaffey JE (1974) Perforation of the tympanic membrane, a complication of tympanic thermometry during anesthesia. Anesthesiology 41:290–291
17. Whitby JD, Dunkin LJ (1968) Temperature differences in the oesophagus, preliminary study. Br J Anaesth 40:991–995
18. Gibbons LV (1967) Body temperature monitoring in the external auditory maetus. Aerospace Med 38:671–675

19. Lipkin M, Hardy JD (1954) Measurement of some thermal properties of human tissue. J Appl Physiol 7:212–217
20. Yamakage M, Kawana S, Watanabe H, Namiki A (1993) The utility of tracheal temperature monitoring. Anesth Analg 76:795–799
21. Togawa T, Nemoto T, Tsuji T, Suma K (1976) Circulatory monitoring by an improved deep body thermometer. Digest 11th Int Conf Med Biol Eng, 40–41
22. Togawa T, Nemoto T (1973) Medical thermometer making use of zero-heat-flow method (in Japanese with English abstract). Rep Inst Med Dent Eng 7:75–83
23. Ball SG, Chalmers DM, Morgan AG, Solman AJ, Losowsky MS (1973) A clinical apraisal of transcutaneous deep body temperature. Biomedicine 18:290–294
24. McKenzie AW, Stoughton RB (1962) Method for comparing percutaneous absorption of steroids. Arch Dermatol 86:608–611
25. Tsuji T, Suma K, Togawa T, Nemoto T (1976) Deep body temperature monitoring in cardiac surgery (in Japanese with English abstract). Iyo Denshi to Seitai Kogaku 14:220–224
26. Kobayashi T, Nemoto T, Kamiya A, Togawa T (1975) Improvement of deep body thermometer for man. Ann Biomed Eng 3:181–188
27. Solman AJ, Dalton JC (1973) New thermometers for deep tissue temperature. Biomed Eng 8:432–435
28. Tamura T, Nemoto T, Togawa T (1976) Perfusion temperature monitoring transducer using zero-heat-flow method (in Japanese with English abstract). Rep Inst Med Dent Eng 10:99–102
29. Yamakage M, Iwasaki S, Namiki A (2002) Evaluation of a newly developed monitor of deep body temperature. J Anesth 16:354–357