Measuring the surface-heating of medical ultrasonic probes

Chr Kollmann¹, G Vacariu², V Fialka-Moser² and H Bergmann¹,³

¹ Dept. of Biomedical Engineering & Physics, Medical Univ. Vienna, Waehringer Guertel 18 – 20, 1090 Vienna, Austria
² Dept. of Physical Medicine & Rehabilitation, Vienna Medical School, Vienna, Austria
³ also with Ludwig-Boltzmann Institute of Nuclear Medicine, Vienna, Austria

E-mail: christian.kollmann@meduniwien.ac.at

Abstract. Due to converting losses the probe’s surface itself is heated up, especially when emitting into air. Possible temperature increases in an ensemble of 15 different diagnostic and therapeutic ultrasound probes from 7 manufacturers in the frequency range between 0.05 – 7.5 MHz have been examined. Surface temperatures were detected by means of a calibrated IR-thermographic camera using a scheme of various power and pulse settings, as well as different imaging modalities as used in clinical routine. Depending on the setup and the output power, the absolute surface temperatures of some of the probes emitting in air can be beyond 43 °C within 5-7 min.; a maximum surface temperature of 84 °C has been detected. Continuous mode or high pulse repetition frequencies on the therapeutic system side, small focused Doppler modes on the diagnostic system side combined with increased emitted acoustic intensities result in high surface temperatures. Within a worst case scenario a potential risk of negative skin changes (heat damage) or non-optimal therapeutic effects seems to be possible if a therapeutic system is used very often and if its emission continues unintentionally. In general the user should be aware that low emission intensities of e.g. 50 mWcm⁻² could already produce hot surfaces.

1. Introduction

The acoustic output of modern ultrasound systems in medical use has been increased continuously over the last three decades; in the process of converting electric power into acoustic power, due to energy losses, the probe and especially the probe’s surface itself is heated up. These losses are high, if the probe is incompletely coupled or emitting in air. In extreme situations this heating might cause damages if applied to the skin. The study will outline the measured temperatures for different probes [1].

2. Experimental set-up

The measurements were carried out for 15 modern clinically used physiotherapeutic and imaging US systems from 7 manufacturers operating in continuous-wave (CW) and/or pulsed-wave (PW) mode as well as in different imaging modalities (e.g. B-, Color- or Spectral-Doppler). The frequency range was between 0.05 – 7.5 MHz (Table 1). Three of the physiotherapeutic devices had two different probes, one with a small effective radiating area (ERA) of 0.5 – 2.5 cm² and a larger one with an ERA of 4 – 12.8 cm². In total, 41 therapeutical and 22 imaging operating modes and intensities were investigated.

All probes were carefully cleaned from gel and the surface dried before performing the measurements. An IR-thermography camera (Thermovision 900 TE /spectral response 2-5.4 μm, image size: 204 x 128 px/12 bit, FLIR Systems, North Billerica MA, USA) was positioned perpendicularly above the probe that was fixed in a rig. The optical focus of the camera was adjusted to be directly on the surface of the probe at a distance of 80 cm (spatial resolution 1.7 mrad/20° SW lens). The thermal resolution of the camera in the range of 23°C - 80°C was 0.1°C with an accuracy of 1°C.
A compensation for the ambient temperature (23 ± 2°C) and for emissitivity was done before each measurement. The IR-camera was connected to a PC allowing software-based handling of its functions (ThermoCam Researcher 2000, Flir Systems). Thermal images were acquired when the probe was operating in air for at least 5 min (max. 8 min). The surface temperature was imaged every 5 sec and stored digitally as a pseudo-color image. After recording a complete IR-image sequence, the analysis tool of the camera software was used to obtain the values within the effective area of the probe. In an off-line process the pseudo-color coded temperature values were automatically re-coded into numerical values for the total sequence and stored for further graphical analysis and plotting.

Some equipment had automatic contact detection to switch to operation mode and do not work normally in air or if coupled incompletely; in this case a thin Saran foil (25 µm, Dow Chemical) covering a film of gel was applied to the surface to match the impedance load condition for operation.

The acoustical output power of the physiotherapeutic equipment was measured using a calibrated radiation force balance (model UPM-DT-1, Ohmic Instr. Co., Easton, MD, USA) designed for measuring the US power output up to 30 Watts at resolution steps of 2 mW with a repeatability of ± 3%.

### 3. Results

#### 3.1 General aspects

Depending on the setup and the output power, the absolute surface temperatures of some of the probes emitting in air can be beyond 43°C within 5-7 min.; a maximum surface temperature of 84°C has been detected. Continuous mode or high pulse repetition frequencies on the therapeutic system side, small focused Doppler modes on the diagnostic system side combined with increased emitted acoustic intensities result in high surface temperatures. The automatic thermal control of the physiotherapeutic systems itself is working only beyond 50°C if existing. The evaluation of the acquired IR-images show clearly that for physiotherapeutic probes the total surface is heating up while on diagnostic probes only small regions are showing these hot temperatures.

#### 3.1.1 Special aspects of physiotherapeutic systems

An increase in surface temperature above room temperature was observed which ranged from 0.2°C up to 40.0°C at 5 min. One system has shown already at 4 min a maximum surface temperature increase of around 58°C using the highest intensity available for this mode and a large probe ERA.

### Table 1. Survey of the systems and modes measured

| model          | manufacturer                      | type of probe | frequency / ERA | modes  | $I_{DATA}$ [W cm$^{-2}$] |
|----------------|-----------------------------------|---------------|-----------------|--------|--------------------------|
| Impulsaphon    | Dr. Born, D                       | circular flat, 1 MHz | cw              |        | 0.5 - 2                 |
| MT 50          |                                   | 2.5 cm$^2$/ 5.0 cm$^2$ | pw              |        | 0.5 - 3                 |
| Sonoplus 590   | Enraf-Nonius B.V., NL             | circular flat, 1 MHz / 5.0 cm$^2$ | cw          |        | 0.5 - 2                 |
|                |                                   | 16/48/100     | Hz              |        | 0.5 - 3                 |
| Uniph          | Phyaction Supporta, NL            | 1:1 (cw)      | 0.5 - 2         |        |                         |
|                |                                   | 1:2/1:4/1:8   | (pw)            |        |                         |
| Phys-Assist    | Orthosonics Ltd., UK              | spherical     | 5/15/30/50      |        | mW cm$^{-2}$            |
|                |                                   | 0.046 MHz / 12.8 cm$^2$ | M/B/PD/CD      |        |                         |
| UM 9           | ATL / Philips                     | L 10-5 / P7 / L5/38 | B/PD/CD        |        |                         |
| Sequoia 512    | Acuson / Siemens                  | 15L8W / 8L5 / 6C2 | M/B/PD/CD/CDE  |        |                         |
| SSH-140A       | Toshiba                           | L 7.5 PLF / PVF 3.75 | M/B/PD/CD/CDE  |        |                         |

100 Hz : pulse 2 ms, pause 8 ms; 48 Hz : pulse 4 ms, pause 16 ms; 16 Hz: pulse 16 ms, pause 47 ms; 1:2 (PW-mode) : 1 pulse period followed by 2 pause periods (others analogue)
Figure 1. Surface temperatures of probes over time operating in CW-mode. A large probe ERA generally generates a higher surface temperature at 5 min than a smaller ERA of the same equipment operating with the same nominal intensity. Figure 2 shows the rise in temperature for the equipment operating in PW-mode. The surface temperatures reached after 5 min are lower then for CW-Mode. The increase in surface temperature can be fitted well using an exponential function (\(\Delta T=\)).
a (1-exp(-bt)), where t is the time in [sec] and a,b coefficients. For most of the operating modes the highest value of surface temperature is observed around 5 min after switching on (saturation effect). In Figures 1 & 2 the limit of surface temperature as obliged by IEC 60601-2-37 standard [2] for diagnostic imaging probes is indicated as a broken horizontal line. The standard requires that the maximum surface temperature shall not exceed 50°C when applied to patients. The limit is exceeded by 10.5 % (2/19) within 2 min resp. by 26.3 % (5/19) within 5 min of the probe’s settings investigated for operation in CW-mode. For PW–mode the temperature increase is smaller so that none of the probes reached this limit within the first 2 min but 13.6 % (3/22) within 5 min.

For CW-mode operation a linear dependence of surface temperature on intensity was found at 1 min. A surprising result was the high temperature increase of the surface temperature for the low-intensity, low-frequency equipment (Table 1, Figure 1).

Data sets from two systems could be included only in part: In both cases the maximum output settings (CW-mode & max. intensity) caused such a high absolute surface temperature (51°C resp. 82°C) within 1 min resp. 4 min operating time that the thermal control unit of the system switched off the device.

3.1.2 Special aspects of imaging systems

The increase of surface-heating for imaging probes is relative moderate and does not exceed the limit of 45°C (Figure 3). In detail, operation in pulsed-Doppler (PD) or Power Doppler-mode (CDE) results in high temperature increases compared to B- or M-mode operation. At Multihertz operation (different nominal frequencies) using the same mode the resulting surface-heating seems to change. Hot spots at the probe’s surface occur because of the special emission characteristics the probe is driven with.

![Figure 3. Surface temperatures of probes for imaging systems operating in different modes.](image)

3.2 Measurements of physiotherapeutic output power

Figures 4, 5 list the displayed and measured output powers for the various equipment settings. It is remarkable that the majority of the equipment emit considerably less power than displayed on screen. Only one probe emits more power than displayed at all operation modes. The differences range between -32% - 28%. In many cases the power data are exceeding the limit of ± 20% allowed by the international safety standard IEC 60601-2-5 for differences between effective and displayed output po-
wer [3]. The output power for the low-frequency system could not be measured because the emitted frequency was outside the working range of the radiation force balance.

**Figure 4.** Comparison of displayed and measured output power for physiotherapeutic systems operating in CW-mode for different settings.

![ Comparison of displayed and measured output power for physiotherapeutic systems operating in CW-mode for different settings. ](image)

**Figure 5.** Comparison of displayed and measured output power for physiotherapeutic systems operating in PW-mode for different settings.

![ Comparison of displayed and measured output power for physiotherapeutic systems operating in PW-mode for different settings. ](image)

### 4. Conclusion

Our results demonstrate that physiotherapeutic machines are able to heat up the treatment head surface within a few minutes to temperatures that, if applied to a patient, could cause skin irritations or local disorders of micro-circulations. A hot treatment head surface occurs normally only in a worst case situation: due to missing automatic contact detection, failures of the thermal control system and timer and/or from unintentional continuous operation caused by the user. In order to avoid these incidents, it is recommended that users should take care to terminate the emission of US power actively and that a periodical maintenance program is executed for the systems used. Furthermore it could be advisable to use a specially adapted gel pad between the patient’s skin and the treatment head if possible or to conduct the treatment under water. Finally there should be a sufficient inactive operation time for the machine to allow the treatment head surface to cool down to ambient temperature before the next patient
is being treated and to achieve a high level of uniform therapeutic heating within the aimed body depth.
The investigated diagnostic imaging systems do not exceed the limits given although special setups are close to it.
Regarding the physiotherapeutic power emission more stricter international limits of variation would be useful, as well as an obligatory thermal control unit and self-adjusting constant power emission for each system.
With the launch of low-frequency systems (<100 kHz) into routine clinical physiotherapeutic use, conventional power measurement equipment (radiation force balance) is not able to check the output; therefore new and easy-to-use methods have to be developed for these kind of systems.
Finally recurrent output controls are advised to inform the user about the status of the machine and to keep him aware that low emission intensities of e.g. 50 mWcm⁻² could already produce hot surfaces.

5. References
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