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● *Original Contribution*

THE VARIATION OF HEATING DEPTH WITH THERAPEUTIC ULTRASOUND FREQUENCY IN PHYSIOTHERAPY

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Abstract—In patient treatment, different ultrasound (US) frequencies are attributed to differences in penetration and, as an effect of that, to different heating depths in tissues. A set of 13 experiments was carried out with US frequencies of 0.86, 2 and 3 MHz. A dynamic treatment protocol and a stationary treatment protocol were used. The temperature increase patterns were thermally imaged with a 1-min interval during an insonation of 5 min. At every data point, the temperature in the reference image was subtracted from the thermal image after 1, 2, 3, 4 and 5 min. In every difference thermal image, the distance between the US applicator and the deepest point of heat increase was measured. Results show that US frequencies do not affect the depth limit for the different temperature ranges, in either the static treatment protocol or the dynamic treatment protocol. (E-mail: jhd@hib.no) © 2003 World Federation for Ultrasound in Medicine & Biology.

Key Words: **Ultrasound, Frequency, Thermal effect, Physical therapy, Therapeutic use.**

INTRODUCTION

The purpose of this article is to examine one of the biophysical effects of therapeutic ultrasound (US) as used in physical therapy, namely, the thermal effect. The focus will primarily be on the use of different frequencies to achieve heating in mammalian tissues. We investigated if different US frequencies have an effect on the heating depth in mammalian tissue. During the last decade, several new US machines have been introduced in physical therapy. Especially stressed is the importance of the choice of different US frequencies. The various manufactures quite rightly claimed and, in fact, still do, that the penetrating depth of US energy increases with a decreasing frequency. Already in 1948, Hüter (1948) described this phenomenon. Hüter stated that the higher the frequency, the greater the attenuation of US energy in structures near the surface. At a lower frequency, there is less attenuation of energy in surface tissues, allowing more energy to be available for absorption in deeper tissues (Lehmann et al 1966; McDiarmid et al 1996). Ward and Robertson (1996) claimed that the penetration depth values and rate of absorption of energy (heat production) are closely related. A small penetration depth is associated with limited transmission of energy, with rapid absorption of energy and with a higher heating rate in a relatively limited tissue depth. A high penetration depth is associated with the efficient transmission of energy and with little absorption and, consequently, limited tissue heating. So, the choice of frequency within the range of 0.8 to 3 MHz is a function of the required depth of penetration and the thermal and acoustical properties of the tissue through which the US travels. These properties are well known and generally accepted in physiotherapy (Jackins and Jamieson 1990; Lehmann and Delateur 1990; Ward and Robertson 1996; McDiarmid et al 1996; Young 1996). Traditionally, biophysical effects of US are separated into thermal and nonthermal mechanisms. This separation has been done more to classify the mechanisms, than that one class of mechanism can appear separately. Nevertheless, different physical therapy textbooks describe US therapy as a heating modality with deep-heating characteristics (Lehmann and Delateur 1990; McDiarmid et al 1996; Young 1996). In Europe, some of the health insurance companies have classified

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US therapy as a deep-heating modality. In Norway, for example, the national health insurance does not allow administration of US therapy together with another deep-heating modality in the same therapy session. Deep-heating modalities are believed to cause a rise in temperature to depths of 3 cm or more in tissue. In physiotherapy, US is used to increase selectively the temperature of periarticular structures (Lehmann et al 1968) and areas at bone-muscle interfaces (Lehmann et al 1967a, 1967b; Roebroeck et al 1998). Physical therapy makes a distinction between the ranges of temperatures considered to have a beneficial influence on human tissue. A weak heating of 1 to 2°C should, according to Lehmann (1982) and Castel (1993), result in a 13% increase of the metabolic rate for each degree Celsius. A moderate heating of 2 to 4°C should reduce muscle spasms, pain, chronic inflammation and promote blood flow, and a strong heating $>$ 4°C should decrease the viscoelastic properties of collagenous tissue (Castel 1993; Draper et al 1995). To obtain such temperatures in deeper tissue layers, a variety of technical US parameters can be varied, such as intensity, US frequency, mode of energy transfer and a static or dynamic treatment protocol.

In general, for soft tissues, the attenuation coefficient increases approximately linearly with frequency for the frequency range used in physical therapy US (ter Haar 1978). This means that the depth of penetration is inversely related to frequency (Dyson 1985). That is the reason why physiotherapists, among other technical US parameters, use US frequency to obtain the desired heating depth. With the depth of US energy penetration in mind, we developed an investigational plan to determine if frequencies have an effect on the heating depth in mammalian tissue. First, a static treatment protocol was used on three different mammalian tissues with frequencies of 0.86 MHz, 2 MHz and 3 Mhz and with all of the other technical US parameters constant, to obtain the depth limits for the different temperature ranges in the mammalian tissues caused by the used frequency. Second, a dynamic treatment protocol on two different mammalian tissues with frequencies of 2 MHz and 3 MHz was used to give the applicability of the results a more practical reality because, in clinical practice, the sound head is moved continuously. Expressed as a hypothesis: A decrease in frequency will result in a deeper penetration for the different temperature increase levels in mammalian tissues. The practical test of this hypothesis may be posed as follows: In a dead cross-section of mammalian tissue, will we observe an increased depth of the isotemperature contours, such as the 1° to 2°C contour, 2° to 4°C contour and the $>$ 4°C contour, when the frequency alone is decreased and the other settings, as indicated on the equipment, are kept constant.

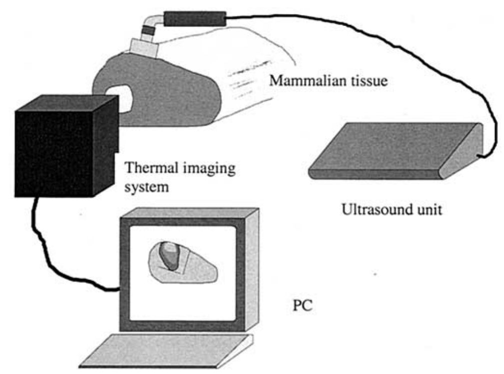

Fig. 1. Schematic diagram of the equipment used for the measurement of the heating depth in different temperature ratios with different frequencies.

MATERIALS AND METHODS

General overview of procedure

A schematic diagram of the equipment used for the measurement of the heating depth in different temperature ranges with different frequencies is shown in Fig. 1.

This study used four fresh pig cadaver specimens from diverse locations of the hind legs of one animal, reflecting similar locations that a physiotherapist can find in clinical situations. Figure 2 presents the tissue geometries of the used specimens.

The choice of pig tissue was made because, in a number of studies of US (Byl et al 1992, 1993; Forrest and Rosen 1989), it has been used and has previously been described as having a similar ratio of skin, fat, muscle and bone as human tissue. Lehmann et al (1967b) reported a greater level of heating when US was applied to dead, rather than to live, animals. They attributed this greater heating to the absence of the cooling effects of blood circulation. The height and width of the tissues were measured to use as reference for the measurements of depth in the thermal images, and the position of the transducer was marked. The tissues were cut into two halves, where the cutting face was in front of the transducer position. During the treatment, the halves were pressed to each other with a rubber band to provide good acoustic impedance match and to avoid reflections. This was interrupted every min during the session of 5-min treatment to take a thermal image.

The height and width of the tissues were measured to use as references for the measurements of depth in the thermal images. Before treatment, each tissue was shaved and thermally prepared. The thermal preparation implies that the tissues were equilibrated to 15°C in saline before each measurement. This is because, in a pilot project, temperature increases up to 22°C were found; to prevent tissue damage caused by excessive heat increase, 15°C was chosen as the equilibration tempera-

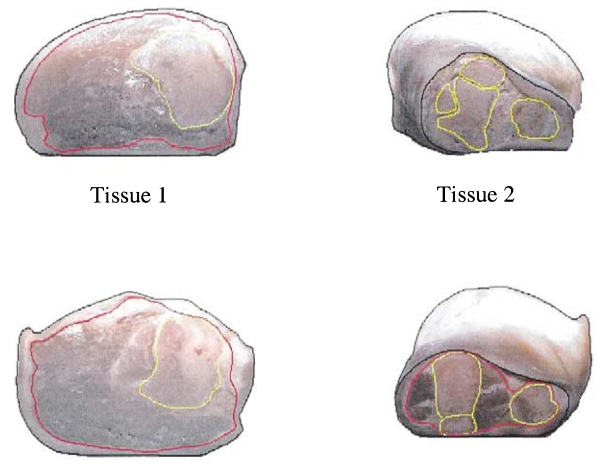

Fig. 2. The tissues and the tissue outlines. The tissues 1, 2 and 3 were used in the static treatment protocol and tissues 3 and 4 were used in the dynamic treatment protocol.

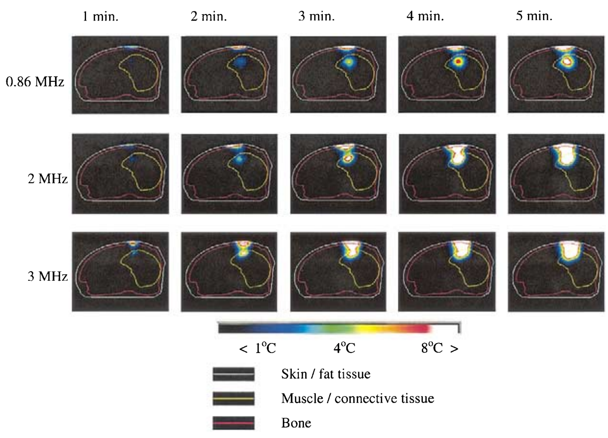

Fig. 3. Example of the thermal difference images in tissue 1 with the static treatment protocol.

ture. The experiments were conducted in a room with constant temperature of 15°C. The environmental conditions (*e.g.,* room lighting, ventilation and the presence of other heat sources in the room) were maintained to be the same for all measurements. The coupling agent was also held at a temperature of 15°C and thinly applied on the tissue surface. For each tissue, the test procedure was applied in the following predetermined order. Before starting the US unit, a thermal image was taken for reference use. The insonation with a frequency of 3 MHz and an intensity, I_{SATA}, of 2 W/cm^2 was started. Every min, a thermal image was taken until the 6th min. The standard thermal preparation of the mammalian tissue was repeated and a new treatment with a frequency of 2 MHz started. The whole procedure was repeated with the other tissues.

For the static treatment protocol, the same procedure with 3 MHz, 2 MHz and 0.86 MHz frequencies was used on the other mammalian tissues.

Ultrasound unit

For the study, a regular physiotherapeutic US unit (Chattanooga Intelect 300 US, Chattanooga Group, Bicester, Oxfordshire, UK) was used, operating at frequencies of 0.86, 2 and 3 MHz. The circular plane applicator with a radius of 1.24 cm and an effective radiation area of 5 cm^2 was connected to the unit. The technical US parameters were set as follows: continuous mode and an I_{SATA} at 2 W/cm^2 during the entire experiment. The frequencies were set at, respectively, 3, 2 and 0.86 MHz. Before each session, the frequency was checked and calibrated with a hydrophone (Precision Acoustics Ltd., Dorchester, UK) to ensure the same frequency. Furthermore, the protocol for testing US machines, as described by Pye and Milford (1994), was used before each session. This protocol contains a functional test of the treatment timer, a qualitative inspection of the beam symmetry, and measurement of acoustic output power at 2 W/cm^2 using a radiation force balance. The US applicator was, in the dynamic treatment protocol, moved in a circular technique at a rate of approximately 2 cm/s on the skin in a template, cut at precisely twice the size of the radiating area of the US applicator. For the static treatment protocol, the US applicator was fixed in a tripod with a clamp. Between the applicator and the skin of the mammalian tissue, a US transmission gel (Aquasonic 100, Parker, Fairfield, NJ) with a temperature of 15°C, was used. US gel was thinly spread on the surface of the skin. On completion of the US exposure, the mammalian tissue was cleansed of the gel.

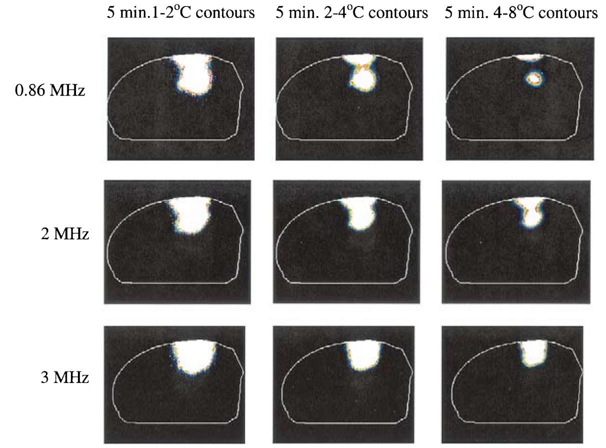

Fig. 4. Example of the different temperature contours in tissue 1 with the static treatment protocol.

Table 1. The heating depth ratios

	Deepest point of 1–2°C contour (2 MHz: 3 MHz)	Deepest point of 2–4°C contour (2 MHz: 3 MHz)	Deepest point of 4–8°C contour (2 MHz: 3 MHz)	Deepest point of 1–2°C contour (0.86 MHz: 3 MHz)	Deepest point of 2–4°C contour (0.86 MHz: 3 MHz)	Deepest point of 4–8°C contour (0.86 MHz: 3 MHz)
Tissue 1 static technique	1.02	1.06	1.15	1.10	1.14	1.26
Tissue 2 static technique	1.00	1.03	1.15	1.20	1.12	1.17
Tissue 3 static technique	1.16	1.14	1.16	1.00	1.04	1.11
Tissue 1 moving technique	1.06	0.74	0.68	–	–	–
Tissue 3 moving technique	0.97	0.76	0.70	–	–	–

Temperature analysis

To study the heat pattern in the treated tissues, we used a Compix PC2000e thermal imaging system. The system communicates with a laptop computer with a Pentium-based processor having a clock speed of 160 MHz. Because the aim was to get a detailed picture of the temperature pattern, all thermal images were compared with the reference image, being the start image without insonation. During the actual computerized comparison, every temperature data point from the reference image was "subtracted" from the thermal image. In this way, it is possible to create a difference heating pattern in the tissue as caused by the US radiation. An example of such thermal difference images is shown in Fig. 3.

In every thermal difference image, the distance between the US applicator and the deepest point of heat increase point for the different temperature levels was measured. Heating depth ratios were calculated by taking the 1-min heating depth of, respectively, the 2 MHz and 0.86 MHz divided by the 1-min heating depth of 3 MHz in the same tissue. This was done for all of the different ranges of temperatures and also for the 5-min heating depth. A total of 65 difference heating-pattern images were obtained and analyzed. The thermal image camera was at the same distance and position with regard to the mammalian tissues during the entire experiment.

RESULTS

For the static technique as well as the moving technique, the heating depth ratios are presented in Table 1.

Table 1 shows clearly that none of the measured heating depth ratios in the static technique come much above a value of 1, neither at 2 MHz nor at 0.86 MHz, which means that none of these frequencies produces heating significantly deeper than at 3 MHz. For the 2-MHz frequency, the average value was 1.1 and for the 0.86 MHz it was 1.1. In trying to come closer to the usual method of doing US treatment in clinical practice, a dynamic treatment protocol was used with 3 MHz and 2 MHz frequencies on different mammalian tissues. It is remarkable that the depth ratio for 2 MHz only in tissue 1 is just over 1 (Table 1) and, in the other tissues, the ratios are below the value 1, meaning that 2 MHz does not reach the same depth as the 3-MHz frequency in the same tissue and for the same temperature range.

For all frequencies in the same tissue, for the same temperature range and the same technique, no significant differences ($p > 0.05$) were found in heating depths in the static technique or in the moving technique.

Figure 4 gives an example of the different temperature contours in tissue 1.

The results show that different mammalian tissues give different heating depths. The different tissue geometries and, therefore, the different thermal and acoustical properties of the tissues through which the US beam travels have a cardinal influence on the depth limit for the different temperature ranges.

DISCUSSION

The purpose of this study was to determine if US frequency could be used as a specific parameter to extend the heating depth in biologic soft tissues in physiotherapy. A common given advice in physiotherapy textbooks is that the depth of penetration of the US energy and the subsequent selective tissue heating are frequency-dependent. The 3-MHz frequency is more appropriate for superficial heat and the 1-MHz frequency for deep heat (McDiarmid et al 1996; Young 1996; Low and Reed 2000). The studies of ter Haar (1978) and Wells (1977) are misunderstood. Although it is correct that that the absorption coefficient varies linearly with frequency (ter Haar 1978) and that the distance rate of energy deposition increases with decreasing frequency at a fixed intensity (Wells 1977), it is oversimplified to draw the conclusion that it is an advantage to use 1-MHz US for heating tissues at 2.5 to 5 cm depth and 3-MHz US to heat tissues at $<$ 2.5 cm depth (Gann 1991). Some later experiments have been carried out to obtain insight on the temperature changes induced by therapeutic US (Cambier et al 2001; Draper et al 1995). These experiments have in common that they used thermocouples placed at certain distances from the US transducer. The distances are based on the half value depth, meaning the distance in which the intensity in that tissue decreases by half. Draper and colleagues found small differences in the heating effects at half-value and twice the half-value. They concluded that the small differences in the heating effects were caused by the heat conduction and the close proximity of the thermocouples. Cambier and colleagues placed the probes at 1 cm, 3 cm and 5 cm from the US applicator for the 1-MHz frequency and for the 3-MHz frequency, and concluded that their results confirm the theoretical predictions of depth of penetration, if the increase in temperature is a measure for the remaining level of energy.

However, our results show that US frequency does not affect significantly the heating depth ratios for the different temperature ranges, in either the static treatment protocol or in the dynamic treatment protocol. Measuring the heating depth after some minutes had a disadvantage, in that the temperature rise not only depends on the ultrasonic intensity distribution, but also on the thermal and acoustical properties of the tissue through which the US travels. For, after the extended time of 5 min (Fig. 3), the variations in temperature were smoothed out by thermal conduction. Furthermore, the thermal difference images shows that the tissue geometry and properties have a great influence on the heating depths, especially the geometry and acoustical properties of bone in proportion to the direction of the US beam. The high absorption in bone and the low value of the specific heat of bone together with thermal conduction tends to equalize the heating depth. The finding that the temperature increases in all tissues and with both techniques has a greater and faster increase at a higher frequency than at a lower frequency, can be attributed to a greater deposition of energy at the given location. Consideration should also be given to the thermoregulation effect of blood circulation. By measuring directly the depth limits, by taking thermal images of sectional planes of fresh pig cadaver specimens, this study had a major disadvantage; it lacked the thermoregulation effect of blood circulation. This disadvantage may also be interpreted as an advantage, in that no individual differences in circulation could contaminate the results.

Although it is unwise to draw too many conclusions from the small number of results given in this paper, examination of Table 1 and Figs. 3 and 4 suggests that a decrease in frequency will not result in a deeper temperature penetration in mammalian tissues. The advice and the assumptions about heating depth commonly given in physiotherapy practice are based on the relationship between the attenuation coefficient and frequency. This advice ignores the influence of thermal parameters, such as the intensity distribution, the thermal conductivity of the tissue and the exposure time.

Our results, although not applicable for extrapolation to *in vivo* circumstances, indicates that the common assumptions about the frequency-dependence of heat penetration depth with US treatments are inadequate.

CONCLUSIONS

In summary, if one of the intentions of using therapeutic US in physiotherapy is the thermal effect, it is necessary that the physiotherapists take into account the thermal parameters. Within the limits of this study, the findings demonstrated that the thermal parameters, such as the intensity distribution, the thermal conductivity of the tissue through which the US travels and the treatment time, are parameters that play a part in the heating depth.

It is questionable if the desired heating depth can be obtained by US frequencies and reaches the desired temperature increases to achieve beneficial effects in tissue. The results have raised questions that warrant further study.

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