LUND UNIVERSITY

Effects of high intensity focused ultrasound on the intervertebral disc

Persson, Johan K

2006

I

ELSEVIER

PII: S0301-5629(02)00566-5

● *Original Contribution*

ULTRASOUND NUCLEOLYSIS: AN *IN VITRO* STUDY

Johan Persson, Björn Strömqvist, Gustavo Zanoli, Ian McCarthy and Lars Lidgren

Department of Orthopedics, Lund University Hospital, Lund, Sweden

Abstract—Thermal intradiscal therapy for chronic low back pain, using a catheter inserted into the intervertebral disc, is becoming more popular in the treatment of low back pain. The aim of this study was to investigate the possibility of heating the nucleus pulposus of the intervertebral disc with high-intensity focused ultrasound (US) or HIFU. Two specific situations were considered, invasive transducers that would be in contact with the annulus fibrosus of the disc, and noninvasive transducers that could be used externally. Theoretical simulations were performed to find the optimal parameters of US transducers and then experimental studies were done using transducers made to these specifications. These experiments confirmed that it was possible to heat the discs with HIFU. Two orthogonal transducers resulted in a superior temperature distribution than using just one transducer. It is, therefore, feasible to consider thermal treatment of the nucleus pulposus of the disc using noninvasive US. (E-mail: Johan.Persson@ort.lu.se)   © 2002 World Federation for Ultrasound in Medicine & Biology.

Key Words: Focused ultrasound, Ultrasound therapy, Disc herniation and thermal therapy.

INTRODUCTION

Disc herniation has been treated surgically since the 1930s by removing the extruded disc material and/or a part of a bulging disc (Mixter and Barr 1934). More recently, the surgical treatment has evolved into less invasive methods using microsurgical and, later, percutaneous techniques for mechanical (Maroon and Onik 1987; Postacchini 1999) or thermal (Choy et al. 1989; Quigley and Maroon 1994) removal of disc material. Recently, a new heat treatment of disc-related back pain has been presented, called IDET (intradiscal electrothermal annuloplasty treatment) (Karasek et al. 1999; Maurer 1999; Natali 1999; Saal and Saal 1999a, 1999b, 2000). In this method, the heat is distributed from a navigable catheter inserted into the disc. An alternative to surgical treatment is chemonucleolysis (Smith 1964), where the enzyme chymopapain is injected into the nucleus pulposus, the central part of the disc. The enzyme depolymerises the long proteoglycans chains in the nucleus pulposus, with subsequent loss of water-binding capacity. This reduces the volume and pressure in the nucleus pulposus and the herniated fragment, explaining the relief of sciatica in patients following chemonucleolysis. A

success rate of 75% and cost-effectiveness have been well documented (Dabezies et al. 1988; Morrison and Felts 1999). Unfortunately, it has an anaphylaxis rate estimated to be about 1%. The next step in this evolution would be a noninvasive treatment of disc herniation that, at best, would be pain-free, avoid the risk of infection and could be performed on an outpatient basis.

The overall aim of this work was to explore the possibility of using focused ultrasound (US) for the treatment of disc herniation. The main idea was to attain a local increase in temperature in the nucleus pulposus to achieve protein degradation, leading to shrinkage of the nucleus pulposus. High-intensity focused US has been used for lesioning tissue since the 1950s (Fry et al. 1954, 1950). US could be emitted through different ports, from multiple elements, focusing on the central parts of a disc. The US may be navigated by the use of an optical positioning system in combination with computerized tomography (CT) or x-ray images, diagnostic US or with a laser-based coordinate system.

The intervertebral disc consists of an outer fibrous part, the annulus fibrosus, and an inner more viscous part, the nucleus pulposus. The disc functions as a shock absorber and, if the annulus fibrosus is ruptured, even in a small area, the nucleus pulposus can herniate, which may cause a nerve compression and induce an inflammatory reaction (Olmarker et al. 1995, 1993). It is also

Address correspondence to: Johan Persson, Department of Orthopedics, Lund University Hospital, SE-22185 Lund, Sweden. E-mail: Johan.Persson@ort.lu.se
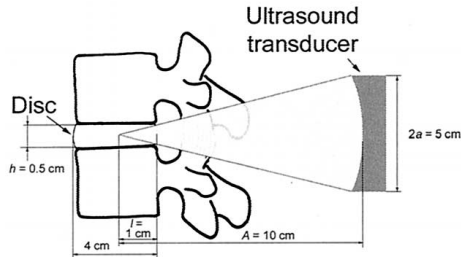
Fig. 1. Dimensions used in the optimization of frequency for the external transducers.

Table 1. Constants and variables used in the simulation

$v = 1540$ m s^{-1}	Sound velocity in tissue
$l = 1$ cm	Depth in the disc where the temperature focus should be
A	Geometrical focal distance
α	Attenuation coefficient
z	Distance from the transducer
a	Radius of the transducer
d	Diameter of the transducer
f	Frequency of the transducer
z_{max}	Distance from the transducer to the maximum intensity
I_{max}	Maximum intensity
I_0	Initial intensity ($z = 0$)

common that the disc protrudes and may compress a nerve root posteriorly, which may cause sciatica.

The absorption of US in musculoskeletal tissues increases with the apatite and protein content (Barnett et al. 1997), which implies high absorption in bone, cartilage, tendons and ligaments. Higher absorption can be expected in the annulus fibrosus (high collagen content) than in the nucleus pulposus (high water concentration). To prevent the temperature in the annulus fibrosus from exceeding a harmful level, while ensuring that the temperature in nucleus pulposus reaches sufficient levels, US could be emitted from several transducers. Thus, the fields will overlap and increase the effect in the nucleus pulposus, but keeping the intensity in the surrounding tissue, including the annulus fibrosus, low.

Both noninvasive and percutaneous methods are investigated theoretically, where mathematical modelling is used to calculate optimum characteristics of the US transducer. In an experimental section, the temperature in bovine discs *in vitro* during US exposures is studied.

MATERIALS AND METHODS

Simulations

To optimize the intensity with respect to the frequency of the transducer, the diameter of the transducer has to be calculated. Because US is reflected when it hits gas or bone, the size of the transducer and its position is limited. The maximum diameter of the US transducer (Fig. 1) is limited by the height (h) of the intervertebral disc, the length (A) through which the US has to pass from the transducer to the nucleus pulposus and the depth (l) where the temperature focus should be in the disc. The height, h, was assumed to be 0.5 cm at the dorsal side of the disc, the distance, A, 10 cm, and the depth, l, 1 cm. These assumptions restrict the diameter of an external transducer to 5 cm, and are valid when the field of the US is not diffracted and the patient has little

to normal subcutaneous fat and no disc height reduction. The anatomy in humans permits the US to be applied dorsolaterally so that the US does not pass through the intestines or spinous processes. This window of "free sight" into the disc makes it possible to place two transducers with diameters 5 cm on each side of the spine; thus, four transducers totally. Transducers intended for percutaneous treatment should have a diameter of less than 0.5 cm to make it possible for insertion through the spinal muscles with minimum discomfort for the patient.

Theoretical calculations were made to determine the optimal resonance frequencies for transducers with different focal distances and diameters. Table 1 lists the variables and constants used in the simulations. Two simulation models were used, Kossoff's model for the central axis intensity (Kossoff 1979), and ULTRASIM, a program for simulation of ultrasonic fields based on a discrete solution of the Rayleigh integral (Holm et al. 1998).

The axial intensity I, as a function of the distance z from a spherically curved transducer with diameter d and radius of curvature A, is given by (Kossoff 1979):

$$I = \left\{ K \sin\left(\frac{\pi T}{2Kz}\right) \right\}^2, \tag{1}$$

where

$$K = \frac{A}{A - z} \tag{2}$$

$$T = \frac{d^2}{4\lambda} \tag{3}$$

and λ is the wavelength. T is the transition distance from the near to the far field of an equivalent flat transducer, which means the distance from the transducer to the point where the US starts to diverge. Equation (1) reduces to:

$$I_A = \left(\frac{\pi T}{2A}\right)^2 \qquad (4)$$

at the optical focus, where the distance z is equal to the radius of curvature A. The frequency f, the velocity of sound v, and the wavelength λ are related by:

$$\lambda = \frac{v}{f}. \qquad (5)$$

Due to the attenuation, the intensity is:

$$I = I_0 \cdot 10^{\frac{-2\alpha f z}{10}}, \qquad (6)$$

where I_0 is the initial intensity from the transducer, f is the frequency in MHz, α is the attenuation coefficient in dB/(cm MHz) and z is the distance from the transducer in cm. In the attenuation coefficient, corrections both for diffraction and absorption are included, and the latter is dominant in soft tissue.

In the calculations for the noninvasive transducers, it was assumed that the attenuation in the tissue is constant from the surface of the transducer to the focus in the intervertebral disc. The intensity was optimized for different values of the attenuation coefficient. In soft tissue, 0.5 dB/cm/MHz is frequently used as a mean value for the attenuation coefficient. Because the protein concentration is higher in the disc than in soft tissue, the attenuation is higher in the disc and, therefore, simulations with different attenuation coefficients were performed ($\alpha = 0.5$, 0.7 and 1.0 dB/cm/MHz).

It was also assumed that the distance from the transducer to the center of the disc is 10 cm, the radius of the disc is 2 cm and that the maximum intensity should be situated at least 1 cm into the disc to be in the nucleus pulposus. This means that z_{max} should be at least 9 cm in the noninvasive scenario. Transducers for invasive US treatment would be applied directly onto the annulus fibrosus, which means that its intensity maximum should be at least at a distance of 1 cm from the transducer. The ULTRASIM program was used to calculate the lateral and longitudinal -3-dB beamwidth, intensity maximum on the central axis, and to make a 2-D longitudinal intensity map.

Experimental procedure

Two sets of experiments were carried out. In one, the US was generated by one or two transducers, each with a diameter of 5 cm, focal distance of 10 cm and frequency of 1.2 MHz. In the other type of experiment, testing of transducers intended for percutaneous treatment was performed. These transducers had a diameter
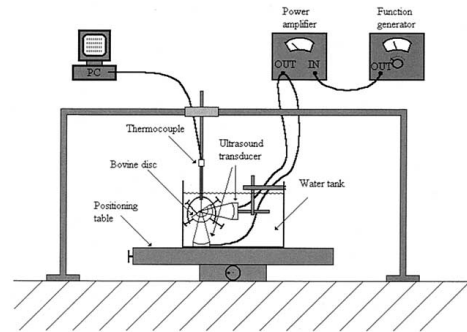

Fig. 2. Experimental setup for temperature measurements in intervertebral discs during exposure with focused US from two transducers.

of 0.5 cm, focal distance of 1 cm (7.6 MHz) or 1.5 cm (3.9 MHz).

The aim of the first set of experiments with the 1.2 MHz transducers was to investigate if it was possible to heat the nucleus with two US transducers. In the following experiments, the aim was to map the temperature both in the annulus and the nucleus. In three of these experiments, the placement of the thermocouples was controlled with X-ray.

The US transducers acoustical outputs were calibrated by means of an US power meter (model UPM-DT-1, Ohmic Instruments Co., Easton, MD) in degassed water. The temperature measurement tests were done *in vitro* and bovine discs were used. Fresh and sagittally cleaved spines were divided into segments, each consisting of half a disc and the adjacent vertebral bodies. The segments were kept frozen in plastic bags until they were thawed and used for testing. The US was generated by piezoelectric crystals made of lead zirconate titanate, type Pz 26, with maximum input power of 10 W cm^2, according to the manufacturer (Ferroperm).

The experiments were performed in a water bath filled with distilled water at 22°C. The water was degassed (kept at a low pressure during at least 2 h) to minimize the formation of air bubbles (cavitation) in front of the US transducer. The experimental set-up is presented in Fig. 2. When two US transducers were used, they were placed perpendicular to each other so that their foci coincided. The water tank was positioned in the horizontal plane by a positioning table.

The transducer was excited by a sinusoidal continuous wave, generated with a function generator (model PM 5134, Philips, first experiment series, or model TG1404, TTi, Cambridgeshire, UK, last experiment series), and then amplified *via* a power amplifier (model

ENI 2100L, ENI Inc., Rochester, NY or model 75A250, Amplifier Research, Souderton, UK). The output signals from the function generator and the power amplifier were visualized using an oscilloscope (model 2230 or model TDS 210, Tektronix, Beaverton, OR). The temperature was measured with thermocouples, diameter 0.8 mm (type K, Chromega® Alomega®) coated with stainless-steel (first set of experiments) or Teflon (second set of experiments), all with exposed junctions. The thermocouples were calibrated using ice water and boiling water. The data were transmitted to a PC, where they were converted to temperature and stored in a text file.

Initial temperature measurement experiments were made with a small piece of rubber ($1 \times 1 \times 1$ mm) on the tip of the thermocouple in the water bath. With this absorber, the position of the focal volumes of the two US transducers were determined and the foci could be positioned to coincide.

Thermocouples were inserted into the disc by means of a cannula that was retracted before the treatment started. The disc segment was placed into a holder that was mounted on a freestanding stand surrounding the water bath. In the first five experiments, it was only possible to rotate the holder around the vertical axis and to translate it vertically. In the second series of experiments, a more flexible holder was used that made translation in three directions possible and also rotation around two axes. Each US transducer was first separately and then simultaneously excited and the corresponding temperatures in the disc were registered.

In the first five experiments, one thermocouple was placed in the nucleus pulposus and the temperature was measured during US exposure. Initially, each transducer was excited separately (5 W acoustical output power) and the temperature was registered. Subsequently, the US transducers were simultaneously excited, leading to a total acoustical output power of 6 W (*i.e.,* 3 W from each transducer) and the corresponding temperature rise was measured. In the following experiments, two temperature probes were placed in the nucleus pulposus, and two in the outer annulus fibrosus. To ensure that the thermocouples were placed halfway between the vertebrae, placement was made under fluoroscopy guidance in three of the experiments (Fig. 3). The temperature probes in the annulus were situated where the fields of transducer A and B pass through the annulus (Fig. 4). The acoustical output power from the US transducers during the excitations was varied in between the experiments, ranging from 2 to 5 W from each transducer. A total of 17 separate experiments with 17 different discs were performed.

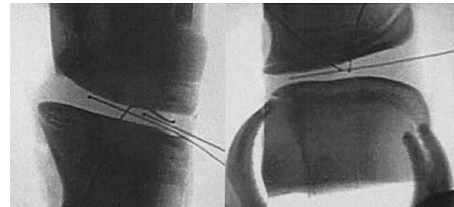The beam width of the 1.2-MHz US transducer was recorded in a water bath with degassed water and a thermocouple with a $1 \times 1 \times 1$ mm rubber absorber on the tip. This was compared with the -3-dB beam width from the USIM results.

Fig. 3. X-ray images of the thermocouple placement in the dorsal (left) and lateral views (right).

The percutaneous treatment transducers were tested *in vitro* on bovine discs. The US transducer was mounted in the bottom of the water bath with a distance of 1 mm between the transducer surface and the annulus fibrosus. The temperature was measured simultaneously at the transducer surface, the outer annulus fibrosus and 1 cm into the disc, in the nucleus pulposus.

RESULTS

Simulations

The results of the optimizations of the intensity with respect to the frequency are presented in Table 2 for three different sizes of noninvasive US transducers and one size of mini-invasive transducers. The simulations were performed using three different absorption coefficients. The
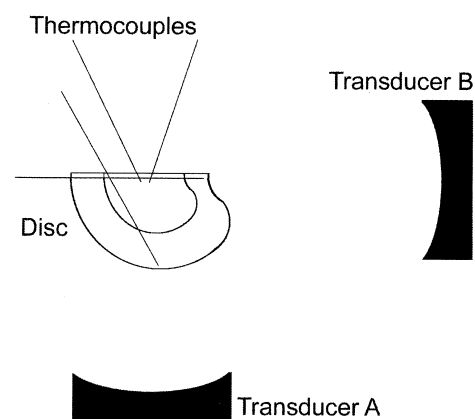

Fig. 4. Placement of the thermocouples in the disc. In the first set of experiments, only one thermocouple was placed in the disc, in the nucleus pulposus.

Table 2. Optimization results of the intensity with respect to the frequency

No.	a (cm)	A (cm)	α (dB/(cm MHz))	f (MHz)	I_{zmax}	z_{max} (cm)	z_{-3dB} (cm)
1	2.5	10	0.5	0.8	$5.7 \cdot I_0$	8.6	3.7
2	2.5	10	0.5	1.1	$5.1 \cdot I_0$	9.0	3.0
3	2.5	10	0.5	1.2	$4.7 \cdot I_0$	9.1	2.8
4	2.5	10	0.7	0.5	$3.6 \cdot I_0$	7.6	4.3
5	2.5	10	1.0	0.3	$2.5 \cdot I_0$	6.1	4.4
6	0.25	1	0.5	7.6	$5.7 \cdot I_0$	0.86	0.4
7	0.25	1	0.7	5.0	$3.6 \cdot I_0$	0.76	0.4
8	0.25	1	1.0	2.9	$2.5 \cdot I_0$	0.61	0.3
9	0.25	1.5	0.5	3.9	$2.3 \cdot I_0$	0.86	0.7
10	0.25	1.5	0.7	2.7	$1.8 \cdot I_0$	0.70	0.6
11	0.25	1.5	1.0	1.7	$1.5 \cdot I_0$	0.51	0.4
12	0.25	2.0	0.5	2.7	$1.6 \cdot I_0$	0.77	0.6
13	0.25	2.0	0.7	1.9	$1.4 \cdot I_0$	0.60	0.6
14	0.25	2.0	1.0	1.2	$1.3 \cdot I_0$	0.42	0.4

Fig. 6. Intensity (I_z/I_0) dependence of the distance from the transducer with $a = 0.25$ cm; $A = 1$ cm; $f = 7.6$ MHz; and attenuation coefficient 0.5 dB cm^{-1} MHz^{-1}. The maximum intensity was 5.7 I_0 at 0.86 cm from the transducer.

intensity maximum of the transducers with a diameter of 5 cm was calculated to be 2.5 $I_0 - 5.7$ I_0, which is satisfactory. The maximum intensity is situated too far from the geometrical focus of the 0.8-MHz US transducer, which means that the focus will be located in the annulus fibrosus. The distance to the point of maximum intensity also decreased with increasing attenuation coefficient. To increase z_{max}, the frequency was raised to 1.1 MHz, which caused a small decrease in the intensity maximum, and the position of the intensity maximum was moved 0.4 cm into the disc. Piezoelectric crystals with these parameters were chosen for the *in vitro* experiments. The resonance frequency of the transducers became 1.2 MHz after mounting the crystals into the
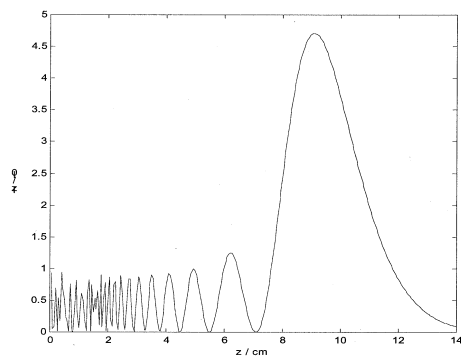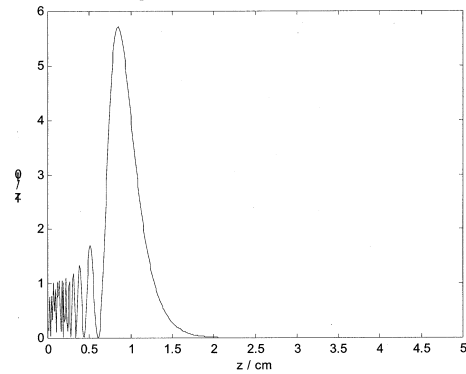
holders, thereby slightly further increasing the distance to the point of maximum intensity. The intensity profile, according to Kossoff (1979), of this transducer is shown in Fig. 5. To get an understanding of how different tissues influence the US field, simulations for one US transducer set-up ($A = 10$ cm; $a = 2.5$ cm; $f = 1.2$ MHz) were done with different absorption coefficients. The result showed that this had an extensive effect on the intensity, but the position of the focus only changed 4 mm when the absorption coefficient was changed from 0.3 to 0.8 dB/cm/MHz.

Fig. 5. Intensity (I_z/I_0) as a function of the distance from the transducer as calculated from Kossoff (1979). The parameters of the transducer were $a = 2.5$ cm; $A = 10$ cm; $f = 1.2$ MHz; and the attenuation coefficient was set to 0.5 dB cm^{-1} MHz^{-1}. The intensity maximum was calculated to 4.7 I_0 at 9.1 cm from the transducer.
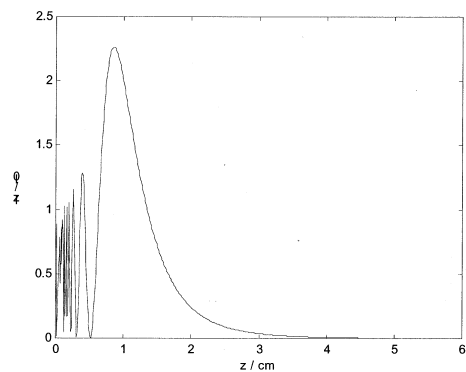

Fig. 7. The intensity distance-dependence of a transducer where $a = 0.25$ cm; $A = 1$ cm; $f = 3.9$ MHz; and $\alpha = 0.5$ dB cm^{-1} MHz^{-1}. The intensity peak, 2.3 I_0, was obtained at 0.86 cm from the transducer.

Table 3. −3-dB beam widths as calculated from USIM

No.	a (cm)	A (cm)	α (dB/(cm MHz))	f (MHz)	$z_{-3dB,longitudinal}$ (cm)	$z_{-3dB,lateral}$ (cm)
1	2.5	10	0.5	1.2	3.0	0.24
2	2.5	10	0.5	1.7	2.3	0.17
3	0.25	1	0.5	7.6	0.42	0.01
4	0.25	1.5	0.5	3.9	0.78	0.06

The calculations for the smaller transducers, with diameter of 0.5 cm, show that focal lengths of more than 1.5 cm do not yield sufficiently high-intensity maxima ($I_{max} \leq 1.6 I_0$). Decreasing the focal length from 1.5 cm to 1 cm approximately doubled the maximum intensity. The problem is the position of the maximum, which is situated 0.9 cm from the transducer. This will probably mean that the annulus fibrosus will be heated to a greater extent than if the focus were situated farther into the disc. For the *in vitro* measurements, transducers with diameter of 0.5 cm and focal lengths of 1.0 and 1.5 cm with frequencies of 7.6 and 3.9 MHz, respectively, were chosen. The intensity profiles of these transducers can be seen in Fig. 6 and Fig. 7.

Regarding the size of the foci the −3-dB longitudinal beam widths calculated from Kossoff's formula are shown in Table 2 as $z_{-3 dB}$. These results agree well with the results from USIM (Table 3), which also gives the lateral beam width. For the 1.2-MHz transducer, the longitudinal −3-dB beam width is 2.8 cm according to Kossoff (1979) and USIM gives a value of 3 cm. The results from USIM show that the lateral beam width should be in the order of mm or part of 1 mm. The calculation model in USIM gives a slightly different figure of the optimal frequency according to the intensity. Instead of 1.1 MHz, which Kossoff's model suggested, USIM gives an optimum frequency of 1.7 MHz.

Experiments

The results of the measured temperature in front of a 1.2-MHz US transducer in degassed water are shown in
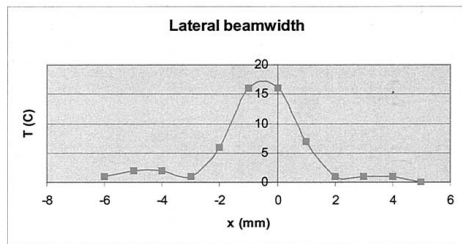

Fig. 8. The lateral temperature distribution in water, measured 100 mm in front of the 1.2-MHz transducer.
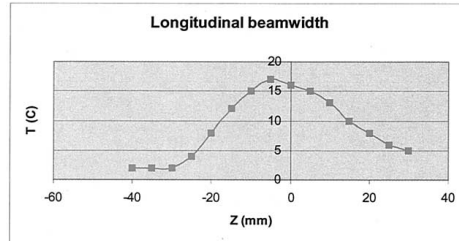

Fig. 9. The longitudinal temperature distribution measured at the central axis of the 1.2-MHz transducer.

Figs. 8 and 9. The lateral half temperature maximum was 2.6 mm and the corresponding longitudinal value was 38 mm.

The increase of the temperature in the bovine discs after 120 s of exposure to continuous US from one or two transducers is presented in Table 4 and Table 5. In Table 5, the result is shown as the differences in mean temperatures between annulus and nucleus after single and double US transducer exposure.

When the invasive transducers were applied to the disc, the temperature at the US transducer surface increased to 80°C or more during continuous US treatment at an output power of 1 W from the transducer. It is likely that heat conduction from the transducer surface contributed to the temperature increase in the annulus. The temperature rise in the nucleus pulposus was 13°C (frequency 3.9 MHz) and 27°C (frequency 7.6 MHz).

DISCUSSION

These experiments have demonstrated the feasibility of heating the intervertebral disc using HIFU. The first

Table 4. Temperature increase in the nucleus pulposus during US exposure

	Mean temperature increase after single US transducer treatment (°C)	Temperature increase after double US transducer treatment (°C)
	14.8	13.7
	14.6	12.3
	5.2	5.2
	13.5	10.5
	16.8	13.9
Mean	11.1	13.0
SD	3.6	4.5

The average of the temperature increase originating from the separate exposures (output power of 5 W) with US transducer 1 and 2 is shown in the left column. The right column shows the temperature increase when both US transducers simultaneously where emitting US, 3 W each.

Table 5. The mean of the temperature increase in annulus fibrosus subtracted from the mean of the temperature increase in nucleus pulposus

	Single US transducer treatment (°C)	Double US transducer treatment (°C)
	4.6	2.1
	5.2	1.2
	8.7	1.7
	3.0	1.4
	2.3	1.7
	10.5	6.9
	3.6	0.4
	2.1	−1.7
	−2.1	−9.0
Mean	4.2	0.5
SD	3.7	4.2

The temperature was recorded after 120 s of US treatment from single (3 W output power) or double (2 W from each transducer) US transducers.

series of experiments demonstrated a temperature increase in the nucleus pulposus, which is regarded as the target tissue. Relatively low powers were used, and the mean temperature increase was 11.1°C with two orthogonal transducers operating at 3 W each. It is clearly possible to achieve much greater temperatures within the nucleus using higher power from the transducers.

The second series of experiments addressed the issue of heating in the adjacent annulus fibrosus. Use of two orthogonal transducers resulted in the same temperature rise in the nucleus but reduced the temperature increase in the annulus. Within the constraints of the anatomy of the lumbar spine, an array of transducers could be used that would reduce heating of tissues surrounding the nucleus. An additional way to minimize the temperature in the tissue between the transducer and the focus could be to pulse the US. The burst length should be on the order of 1 s with a pulse period of 10 s. In this way, surrounding tissues with significant blood flow will have time to cool in between each burst more effectively than the avascular nucleus. Because the main aim of this work has been to position the focus of the US to the nucleus pulposus, it was most convenient to use continuous US. Severson et al. (1997) showed that pulsed US raised the temperature in the disc in an *in vitro* experiment.

In this *in vitro* model, water was used between the US transducer and the disc. In the *in vivo* situation, there is muscle, fat and skin tissue in front of the disc that the US has to pass through. Barkman et al. (1999, 1998) have shown that it is possible to obtain tissue lesioning at deep locations when the US has passed through different types of tissue, such as skin, fat and muscle tissue. The lesions were well discriminated from the surrounding

tissue. They performed tissue lesioning using HIFU *in vitro* on pig muscle and *in vivo* on a pig using both a single US transducer (focal distance of 50 mm, frequency 1 MHz) and a treatment head containing 7 separate focused US transducers (focal distance 100 mm, frequency 0.5 MHz).

The high temperature at the surface of the percutaneous transducer shows that some type of cooling of the crystal is necessary in our application. If the distance between the transducer and the surface of the disc is increased on the order of a few mm, saline flow cooling of the crystal will be possible. This could be distributed through one or more saline channels in the cables to the transducer, similar to that used by Lafon et al. (2000).

The energy of the US must be limited to ensure that no damage occurs to the nerves or soft tissue in front of the focus. This means that the temperature should not exceed 44°C except in the focus, which must be positioned inside the disc. Ongoing experiments measure the temperature in different parts of the disc and the surrounding tissue, with the aim of determining the input power so that the temperature remains below 44°C outside the disc. The heat in the disc will spread by heat conduction. Because bone is a good thermal insulator, the heat will not spread into the vertebrae to a great extent. However, the heat will flow to the outer part of the disc and, when it reaches the annulus pulposus, posterior annular fissures may be welded together (Saal and Saal 1999b). The vascular system outside the disc will have a cooling effect. Thus, potential injury outside the disc associated with heat conduction from the nucleus pulposus should not be a problem.

In a promising new method for treatment of disc-related back pain called IDET (intradiscal electrothermal annuloplasty), heat is used to coagulate nerve tissue, shrink collagen fibers and cauterize granulation tissue in patients with disc degeneration (Karasek et al. 1999; Maurer 1999; Natali 1999; Saal and Saal 1999a, 1999b, 2000). Under local anaesthesia and fluoroscopic control, a trocar is inserted into the disc. A navigable catheter with a temperature-controlled thermal resistive coil, 50 mm long, is inserted so that the heating segment is situated in the disc. The heating starts at a coil temperature of 65°C and increases incrementally to 90°C, and it lasts for about 1 h.

If the transducer surface is spherical, the transducer has a fixed focus, it is also possible to achieve a flexible focus; by means of a phased array (Fan and Hynynen 1995). The reason to replace a fixed with a flexible focus is to enable treatment on people with different body weights and anatomy with the same equipment.

For hyperthermia treatment, information is available on how to plan, monitor and follow the treatment with magnetic resonance imaging (MRI) (Bohris et al. 1999;

McDannold et al. 1998, 1999; Morocz et al. 1998; Vimeux et al. 1999). MRI makes it possible to monitor the temperature change in tissue noninvasively in real-time, although the drawback is that it is an extremely resource-consuming method. It is disputed whether or not diagnostic US can be used to monitor the temperature noninvasively (Gertner et al. 1998; Steger et al. 1992). Gertner et al. (1998) have studied echogenicity as a function of the temperature in a soft tissue such as liver *in vitro* and found that it is dependent on variations in the attenuation. For temperatures up to 40°C, the attenuation decreases, which is thought to be a thermal effect, but, above 40°C, the attenuation increases, which is explained by the irreversible change in the tissue structure. This effect may be used to confirm lesioning noninvasively in US nucleolysis.

It is important that energy can be safely focused into the nucleus pulposus. This could be achieved by a combination of known navigation methods. Another example of how the US could be navigated is to match a pretreatment CT-image of the treatment volume to the position of the patient in the treatment room reference system with regards to a laser beam coordinate system or an optical positioning system. A similar procedure, where the US is navigated by the use of double X-ray images and an optical positioning system, could also be used. Ideally, these navigation methods would be combined with temperature-monitoring techniques described above.

SUMMARY

The theoretical simulations of the 5-cm transducer shows that a good intensity gain can be reached with one transducer with 10 cm focal distance and 1.1 MHz frequency. Theoretically, this gain can be increased four-fold if foci from four transducers coincide which in our case could give intensity amplification in the focus up to 20 times the initial intensity, in front of the transducer surface.

The experiments with these transducers show that it is possible to heat the nucleus pulposus in bovine discs in *in vitro* experiments above to 44°C. When one US transducer was used, the heating was greater in the annulus, but the temperature in the annulus was significantly reduced when using two orthogonal transducers. Improving the positioning system of the transducers, distribution of the US from two to four transducers, and use of pulsed US will increase the temperature in the nucleus compared with the annulus and the intermediate tissue.

The smaller US transducers, aimed for percutaneous treatment, reached extremely high temperatures at the surface during treatment. This caused temperature rises in the disc due to heat conduction, which indicates that a chilling irrigation system in the transducer will be necessary. (Hynynen et al. 1998).

REFERENCES

Barkman CA, Almquist LO, Kirkhorn T, Holmer NG. Thermotherapy: Feasibility study using a single focussed ultrasound transducer. Int J Hyperthermia 1999;15:63–76.

Barkman CA, Kirkhorn T, Almquist LO, Holmer NG. Measurements of the thermal focus of an experimental focused ultrasound thermotherapy system. Int J Hyperthermia 1998;14:383–393.

Barnett SB, Rott HD, ter Haar GR, Ziskin MC, Maeda K. The sensitivity of biological tissue to ultrasound. Ultrasound Med Biol 1997; 23:805–812.

Bohris C, Schreiber WG, Jenne J, et al. Quantitative MR temperature monitoring of high-intensity focused ultrasound therapy. Magn Reson Imaging 1999;17:603–610.

Choy DSJ, Ascher PW, Liebler W, et al. Percutaneous laser decompression of intervertebral disc. Laser Med Surg News Advanc 1989; October 25.

Dabezies EJ, Langford K, Morris J, Shields CB, Wilkinson HA. Safety and efficacy of chymopapain (Discase) in the treatment of sciatica due to a herniated nucleus pulposus. Results of a randomized, double-blind study. Spine 1988;13:561–565.

Fan X, Hynynen K. Control of the necrosed tissue volume during noninvasive ultrasound surgery using a 16-element phased array. Med Phys 1995;22:297–306.

Fry JF, Mosberg MS, Barnard JW, Fry FJ. Production of focal destructive lesions in the central nervous system with ultrasound. J Neurosurg 1954;11:471–478.

Fry WJ, Wulff VJ, Tucker D, Fry FJ. Physical factors involved in ultrasonically induced changes in living systems: I. Identification of non-temperature effects. J Acoust Soc Am 1950;22:867–876.

Gertner MR, Worthington AE, Wilson BC, Sherar MD. Ultrasound imaging of thermal therapy in in vitro liver. Ultrasound Med Biol 1998;24:1023–1032.

Holm S, Teigen F, Ödegaard L, et al. ULTRASIM, User's Manual ver 2.1, Program for simulation of ultrasonic fields. Oslo: Departments of informatics, 1998.

Hynynen K, Wolf D, Wolf G, Jolesz F. MRI evaluation of thermal ablation of tumors with focused ultrasound. J Magn Reson Imaging 1998;8:91–100.

Karasek M, Karasek D, Bogduk N. A controlled trial of the efficacy of intra-discal electrothermal treatment for internal disc disruption. North American Spine Society 14th Annual Meeting, Chicago, 1999; 76–77.

Kossoff G. Analysis of focusing action of spherically curved transducers. Ultrasound Med Biol 1979;5:359–365.

Lafon C, Theillere Y, Prat F, et al. Development of an interstitial ultrasound applicator for endoscopic procedures: Animal experimentation. Ultrasound Med Biol 2000;26:669–675.

Maroon JC, Onik G. Percutaneous automated discectomy: A new method for lumbar disc removal. Technical note. J Neurosurg 1987;66:143–146.

Maurer P. Thermal lumbar disc annuloplasty: Initial clinical result. North American Spine Society 14th Annual Meeting, Chicago, 1999; 73–74.

McDannold NJ, Jolesz FA, Hynynen KH. Determination of the optimal delay between sonications during focused ultrasound surgery in rabbits by using MR imaging to monitor thermal buildup in vivo. Radiology 1999;211:419–426.

Mixter W, Barr J. Rupture of intervertebral disc with involvement of the spinal canal. N Engl J Med 1934;211:210.

Morocz IA, Hynynen K, Gudbjartsson H, et al. Brain edema development after MRI-guided focused ultrasound treatment. J Magn Reson Imaging 1998;8:136–142.

Morrison P, Felts M. A brief history of tissue modulation in minimally invasive spine surgery. North American Spine Society 14th Annual Meeting, Chicago, 1999; 1–10.

Natali C. Good news for bad backs! Intradiscal electrothermal therapy (IDET) a new treatment for painful disc degeneration. Ortopaed Prod News 1999;38–37.

Olmarker K, Blomquist J, Strömberg J, et al. Inflammatogenic properties of nucleus pulposus. Spine 1995;20:665–669.

Olmarker K, Rydevik B, Nordborg C. Autologous nucleus pulposus induces neurophysiologic and histologic changes in porcine cauda equina nerve roots. Spine 1993;18:1425–1432.

Postacchini F. Management of herniation of the lumbar disc. J Bone Joint Surg Br 1999;81:567–576.

Quigley MR, Maroon JC. Laser discectomy: A review. Spine 1994;19: 53–56.

Saal JA, Saal JS. Intradiscal electrothermal annuloplasty (IDET) treatment for chronic multi-level discogenic pain: Prospective 1 year follow-up outcome study. North Am Spine Society 14th annual meeting, Chicago, 1999a; 78–79.

Saal JS Saal JA. Intradiscal electrothermal annuloplasty (IDET) for chronic disc disease: Outcome assessment with minimum one year follow-up. North Am Spine Society 14th annual meeting, Chicago, 1999b; October/November: 75–76.

Saal JS, Saal JA. Management of chronic discogenic low back pain with a thermal intradiscal catheter. A preliminary report. Spine 2000;25:382–388.

Severson J, Albu Barkman C, Strömqvist B, Lidgren L. Kan fokuserat ultraljud användas för att behandla diskbråck? [Can ultrasound be used to treat disc herniation?] Lund: Department of Biomedical Engineering, 1997.

Smith L. Enzyme dissolution in nucleus pulposus in humans. JAMA 1964;187:137.

Steger AC, Shorvon P, Walmsley K, et al. Ultrasound features of low power interstitial laser hyperthermia. Clin Radiol 1992;46:88–93.

Vimeux FC, De Zwart JA, Palussiere J, et al. Real-time control of focused ultrasound heating based on rapid MR thermometry. Invest Radiol 1999;34:190–193.