AN APPLICATION OF DIAGNOSTIC ULTRASOUND TO DETERMINE THERAPEUTIC ATTENUATION AND TEMPERATURE IN PHANTOM

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M.Eng. (BIOMEDICAL ENGINEERING)

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ABSTRACT

Ultrasonic intensities during a therapeutic application of 1 MHz ultrasound are calculated from the data of reflected amplitudes due to 3.75 MHz diagnostic ultrasound exposure in a constructed tissue-equivalent phantom. This tissue phantom is a mixture of 180 g gelatin, 350 g non-dairy cream, 15 g floral cell, 50 g sodium benzoate, and 1 liter of water. The measured average density of the tissue phantom is 1.11 g/cm^3. The ultrasound attenuation in the tissue phantom calculated from echo patterns due to diagnostic ultrasound exposures is 0.434 dB/cm/MHz with a velocity of 1836 m/s. It is found that the attenuation coefficient does not vary with the liquid or solid state of the tissue phantom. The echo patterns due to diagnostic ultrasound exposures indicate that calculated therapeutic ultrasound intensities are only valid for a plane wave application in which this tissue phantom occurs at the depth of 4.3 cm and beyond. The measured temperature distribution in the tissue phantom shows the 2 states of temperature rise. The initial state is in about the first few minutes after therapeutic ultrasound is applied when the temperature increases rapidly without the effect of conduction in the tissue phantom. After that, the rate of temperature rise decreases due to conduction in the tissue phantom; therefore, the temperature slowly increases until it seems to reach a plateau at resting state. After that the measured temperature rise in the solid tissue phantom is compared with the calculated temperature rise from the equation of linear bio-heat transfer. It is most interesting that the measured and the calculated temperatures are in reasonable agreement at the depth where the plane wave of therapeutic ultrasound is applied. The study of temperature distribution in the two layers of the tissue phantom and bone shows that there is a more pronounced rate of increase in the two layers of phantom than in tissue phantom alone. After therapeutic application for 5 minutes, the temperature rise at the bone surface rapidly increases.

KEY WORDS: TISSUE-EQUIVALENT PHANTOM / THERAPEUTIC ULTRASOUND PHANTOM / THERAPEUTIC ULTRASOUND ATTENUATION / TEMPERATURE DISTRIBUTION / STATIONARY TECHNIQUE

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การประยุกต์ใช้เครื่องอัลตราซาวด์ทางการวินิจฉัยในการหาค่าการลดทอนอัลตราซาวน์และอุณหภูมิที่เกิดขึ้นในเนื้อเยื่อจำลองจากการใช้อัลตราซาวด์ในการรักษา (AN APPLICATION OF DIAGNOSTIC ULTRASOUND TO DETERMINE THERAPEUTIC ATTENUATION AND TEMPERATURE IN PHANTOM)

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บทคัดย่อ

ความเข้มของอัลตราซาวด์ที่ใช้ในการรักษาที่มีขนาดความถี่ 1 MHz นั้นสามารถประยุกต์หาได้จากแอมพลิจูดของการสะท้อนกลับของคลื่นจากเนื้อเยื่อจำลองด้วยอัลตราซาวด์ทางการวินิจฉัยขนาดความถี่ 3.75 เมกะเฮิร์ตซ โดยเนื้อเยื่อจำลองนี้ ส่วนผสมประกอบด้วยเจลาติน 180 กรัม ครีมเทียม 350 กรัม ฟลอราเซล 15 กรัม โซเดียมเบนโซเอต 50 กรัม และน้ำ 1 ลิตร ความหนาแน่นของส่วนผสมเท่ากับ 1.11 g/cm^3 โดยเนื้อเยื่อจำลองนี้มีค่าการลดทอนของคลื่นอัลตราซาวด์เท่ากับ 0.434 dB/cm/MHz ที่ความเร็วของการส่งผ่านคลื่นเสียงเท่ากับ 1836 m/s ในการศึกษาพบว่า อัตราการลดทอนของคลื่นไม่ขึ้นอยู่กับสถานะของเนื้อเยื่อจำลองว่ามีสถานะเป็นของเหลวหรือของแข็ง จากผลการสะท้อนของคลื่นที่วัดได้จากอัลตราซาวด์ทางการวินิจฉัยพบว่า ค่าความเข้มของอัลตราซาวด์ที่คำนวณได้จะนำมาใช้ได้ตั้งแต่ระดับความลึก 4.3 cm เป็นต้นไป เนื่องจากคลื่นที่ส่งลงไปจากเริ่มมีลักษณะเป็นระนาบเดียว (Plane wave) ในการศึกษาความเปลี่ยนแปลงของอุณหภูมิในเนื้อเยื่อจำลองพบว่าการเพิ่มของอุณหภูมิจะแบ่งได้เป็นสองช่วงคือ ช่วงแรก อุณหภูมิจะมีการเปลี่ยนแปลงจากอุณหภูมิเริ่มต้นอย่างรวดเร็ว ซึ่งเกิดจากการให้ความร้อนด้วยอัลตราซาวด์ทางการรักษา หลังจากนั้นจะเข้าสู่ช่วงที่สองคือ อัตราการเพิ่มขึ้นของอุณหภูมิจะลดลง ซึ่งเป็นผลจากคุณสมบัติการนำความร้อนของเนื้อเยื่อจำลอง โดยอุณหภูมิที่วัดได้จะมีอัตราการเพิ่มลดลงจนกระทั่งเข้าสู่สภาวะที่ไม่มีการเปลี่ยนของอุณหภูมิในที่สุด จากการเปรียบเทียบอัตราการเปลี่ยนแปลงของอุณหภูมิที่ได้จากการคำนวณในการทดลองในเนื้อเยื่อจำลองชนิดแข็งกับอุณหภูมิที่วัดได้จริงจากสมการ linear bio-heat transfer พบว่า อุณหภูมิที่ได้จะมีค่าใกล้เคียงกันที่ระดับความลึกของการวัดที่คลื่นอัลตราซาวด์เป็นระนาบเดียว ในการศึกษาการเปลี่ยนแปลงของอุณหภูมิในเนื้อเยื่อจำลองชนิดสองเนื้อเยื่อซึ่งประกอบด้วยชั้นของเนื้อเยื่อและชั้นของกระดูก พบว่าอัตราการเปลี่ยนแปลงของอุณหภูมิจะเพิ่มสูงกว่ากรณีที่ให้ความร้อนกับเนื้อเยื่อจำลองที่มีชั้นเนื้อเยื่อเพียงอย่างเดียว ซึ่งหลังจากให้ความร้อนด้วยอัลตราซาวน์ทางการรักษาผ่านไป 5 นาที อุณหภูมิที่วัดได้เหนือผิวกระดูกจะสูงขึ้นอย่างรวดเร็ว

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CHAPTER I
INTRODUCTION

1.1 Introduction

The early use of ultrasound in medicine was largely confined to its therapeutic application in a form of diathermy, utilized its heating and disruptive effects on animal tissues (1). It was not until the early 1940s that attempts were made to employ ultrasound as a diagnostic tool. (2)

Reports from the United States in 1973 stated that for a population of 500,000, ultrasound diathermy were carried out at a rate of 3,651 per month compared to 1,917 microwave diathermy and 1,468 shortwave diathermy per month (3,4). The advantages of ultrasound diathermy are the efficient transfer of ultrasonic energy to tissue (5, 6), deeper penetration (7), greater absorption in tissue of high protein content (8), selective heating at the tissue interfaces (9), and the biological effects of non-thermal mechanisms (10).

When a continuous ultrasound beam is applied to a tissue, the variety of physiological responses to the temperature elevation occurs. These reactions include an increase or decrease in the extensibility of the collagen tissue, neuromuscular activities, the pain threshold, blood flow, membrane permeability, metabolism and enzymatic activities of tissue (11). These responses are occurred when the tissue temperature is in the approximate range of 40 to 45 °C. Moritz and Henriques (12) showed that at the skin temperatures above 44 °C tissue would be damaged, depending on the length of exposure time and the level of temperature raised. The thermal pain is also evoked at 45 °C which is closed to the maximum temperature of therapeutic use. Therefore, the temperature level of the tissues during ultrasound application should be determined.

As ultrasound travels through the tissues, part of its energy is progressively absorbed and converted into heat (13). The amount of heat deposited depends upon the absorption characteristics of the tissue and the amount of ultrasonic energy passing through that tissue (14). The absorption characteristics of a tissue characterized by attenuation coefficient depending on both frequency of the ultrasound and types of the tissue within beam (15, 16). A frequency of 3 MHz of ultrasound is used for the superficial wound treatment whereas 1 MHz ultrasound is used for the deeper tissues (15, 17). The typical frequency of the most ultrasonic therapeutic devices is 1 MHz (4, 18). The attenuation coefficient of 1 MHz ultrasound is about 0.06-0.16 Np/cm for subcutaneous fat and 1.3-3.0 Np/cm for bone (19). The variations in the attenuation properties result in the selectively heating of the tissues. Furthermore, the presence of shear wave in high dense tissue which can highly absorbed at the boundary of the tissue with different acoustic impedance (20) causes the heating at the interfaces of the tissues more pronouncedly than the tissue itself (21). As a result, the nature of the tissue to absorb ultrasound must be considered as an internal factor that could affect the heating patterns in the tissue. The external factors that the therapist can manage are the selection of the frequency and the determination of the total energy.

The total energy is defined by the product of the intensity in Watt/cm^2 and the exposure time (5). This definition is also used by the American Institute of Ultrasound in Medicine (AIUM) to indicate the effects of ultrasound on biological tissue. For the product of intensity and exposure time less than 50 J/ cm^2, no substantial bioeffects have been demonstrated in mammalian tissue in vivo (22). Guideline of the intensity setting for therapeutic purpose is empirically based on the clinical experiences (5) being approximately 0.25 W/cm^2 to 3.0 W/cm^2 for continuous ultrasound (1). For the exposure time, it depends on the area of treatment and the techniques used. By stroking technique Lehmann et al. (23) used the exposure time about 5-15 min. whereas Oakley (24) recommended 2 min. for an area of 1.5 times of the face size of transducer. By stationary technique, Boonsinsuk et al. (25) used 1.8±0.48 min. The exposure time for therapeutic ultrasound is clinically indicated by the patient sensation (5) ranging from comfortable to the first occurrence of pain (23) and to the maximum pain tolerance level (26). Palawiwatt et al. (27) reported the first occurrence of pain responded to 1 W/cm^2 stationary ultrasound 13.1±4.7 seconds. The maximum pain tolerance duration responded to stationary ultrasound at the intensity 0.5 W/cm^2, 1.0 W/cm^2, and 1.5 W/cm^2 reported by Chantaraviroj (28) was 15.05, 26.12, and 86.48 seconds, respectively. Such variations in intensity and exposure time determination are implied the lack of information upon the effects of ultrasound in the living tissue related to the dosage of ultrasound. Although Reid and Cummings (28) had raised the question in 1973 about the optimal dosage calculation that could be applied to a given clinical situation for a maximum desired response, there was only a few studies concerned the effects of the intensity or the exposure time on tissue temperature. Lehmann et al. (29) compared the temperature produced by 1.0 W/cm^2 to1.5 W/cm^2 ultrasound using the first occurrence of pain to judge the exposure time. The results showed the controversial effect of the intensity without mentioned to the total energy. In the group with soft tissue thickness less than 8 cm, the temperature at the first occurrence of pain obtained from 1.5 W/cm^2 was lower. This result implied that the temperature change can be in part due to the exposure time and the thickness of the tissue.

The problem that caused the rarely used of the stationary technique is the temperature control (31). Despite of this problem the advantages of the stationary technique are the controllable total energy and incident angle (32), the specific volume to be sonicated in the small and sharply localized area (5), reduction of exposure time (25), and the easy performing with the proper contact to avoid reflection at the air-tissue interfaces (5). The therapeutic ultrasound head should be applied perpendicular to the surface because thermal effects are noted to be the greatest at 80 degree and 90 degree angles (33).

Patient exposure or the total energy can be minimized by (i) testing patient skin sensation prior to application of ultrasound (if patients have sensory paralysis and unable to differentiate between hot and cold, an alternative type of treatment should be given as the patients applied when anesthetized areas are involved), (ii) keeping the transducer moving slowly over the treatment region to minimize the risk of hot spots, (iii) reducing the ultrasound intensity level, if a mild tingling sensation or pain is felt in the treatment region (such a sensation may be an indication of overheating within the treatment region, and significant damage to the tissue can occur if the exposure is allowed to continue) (30).

When ultrasound beam is passing through tissue composed of blood vessels, nerves, muscle, fat, etc., that overly irregularly, it is hard to assure whether the effect of ultrasound is on the target. Therefore, this study designs to determine the therapeutic ultrasound dosages by using the pulse-echo data from diagnostic ultrasound. The echo amplitude can indicate the absorbed and reflected properties of tissues, and can be related to the intensity at the depth of echo presented. A tissue equivalent phantom simulated tissue is designed and constructed and used for all measurements. It is also designed to calculate the temperature from the echo data measured in phantom by using the linear bio-heat transfer equation, and then compare with the direct temperature measurements in the same phantom.

CHAPTER II
LITERATURE REVIEW

2.1 Therapeutic ultrasound (34)

It is a medical treatment by means of heating and disruptive effects developed by ultrasonic properties in tissues. Treatment intensity is defined in Watt unit per radiating area of the applicator and is from 0.25 W/cm^2. The primary effects of ultrasound on tissue are thermal and mechanical effects. During treatment, ultrasound energy is transferred and absorbed via a contact medium (water soluble gel) to the treatment volume and converted into heat by means of thermal effect. The ultrasonic vibrations can also produce a micromassage by means of mechanical effect. With ultrasound, the excellent penetration features of massage are attained especially when the pain in the treatment area prevents manual massage. Ultrasound can increase the permeability of the cell membrane and metabolism. It is reported that ultrasonic treatment can reduce pain and muscle spasms, as well as slag products and scar tissue formation. The level of ultrasound treatment intensity is selected according to the thickness of subcutaneous fatty tissue of the patient and the distance from the surface of the body to the injured area.

Rule of thumb to select the intensity:

Low intensity up to 0.25 W/cm^2, is used to treat small joints (foot, finger), arthrosis, epicondylitis, ganglia, and scar tissue.

Medium intensity 0.25 – 0.50 W/cm^2, is the most common used.

High intensity 0.50 – 1.0 W/cm^2, is used when a thick fatty tissue is presented.

Dosage. Dose indicating the total energy of ultrasound is the product of intensity and.

The exposure time depends on the tissue volume to be treated. Lehmann fixed the maximum duration of treatment at 15 minutes. This refers to a treated area of 75-100 W/cm^2, which he considers the maximum area that can reasonably be treated. The effective radiating area (ERA) of the treatment head is important in the respect.

The intensity is expressed in W/ cm^2. For continuous beam, < 0.3, 0.3-1.2, and 1.2-3.0 W/ cm^2 are classified as low, medium, and high intensity, respectively. With continuous and pulsed ultrasound at high intensity a sensation of heat may be felt. Only a mild sensation of warmth is acceptable. If, as a result of treatment, fatigue, vertigo, headache, and/or other (autonomic nervous) reactions develop, subsequent treatment should be given at a lower intensity. If the Arndt-Schultz rule is kept in mind it will be clear that in general the intensity of the ultrasound energy applied should be low.

2.2 Diagnostic ultrasound (35)

Diagnostic ultrasound is the use of high frequency sound waves (3-7 MHz) to view organs and tissues inside the body. The diagnostic transducer generates short ultrasound pulses by the piezoelectric effect. As the beam propagates through the patient it encounters anatomical interfaces. At each tissue interface, some of the beam is reflected and some is transmitted according to the difference of acoustic impedances (Figure 2.1). The reflected waves (echoes) return to the transducer where they deform the piezoelectric crystal and produce an electrical signal. . The echo signal is generally weak and must be amplified and processed for display (Figure 2.2 and 2.3).

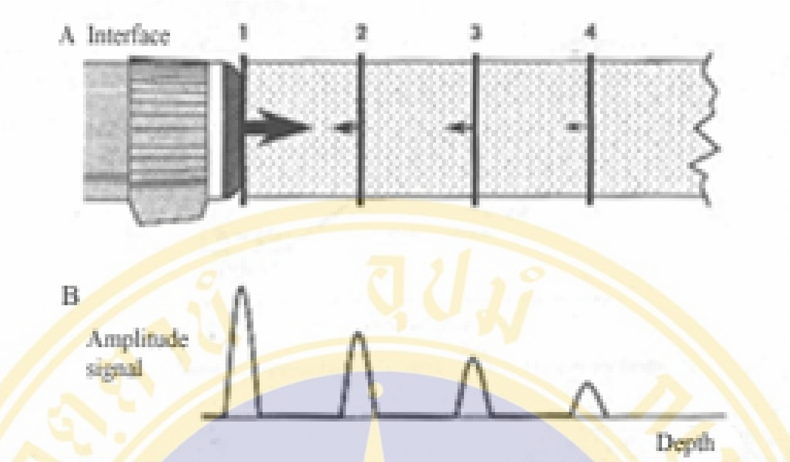


Figure 2.1 A, An ultrasound beam is incident on a series of identical interfaces.

B, The amplitude of the echo from each interface decreases with depth due to attenuation. (35)

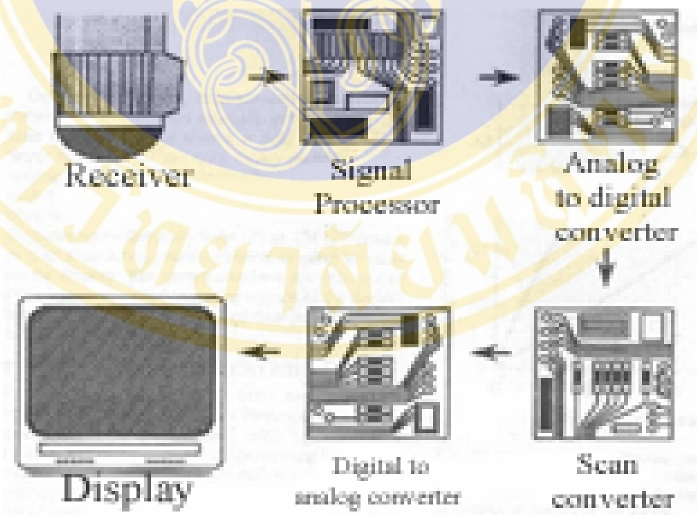


Figure 2.2 The image processing of echo signal. (35)

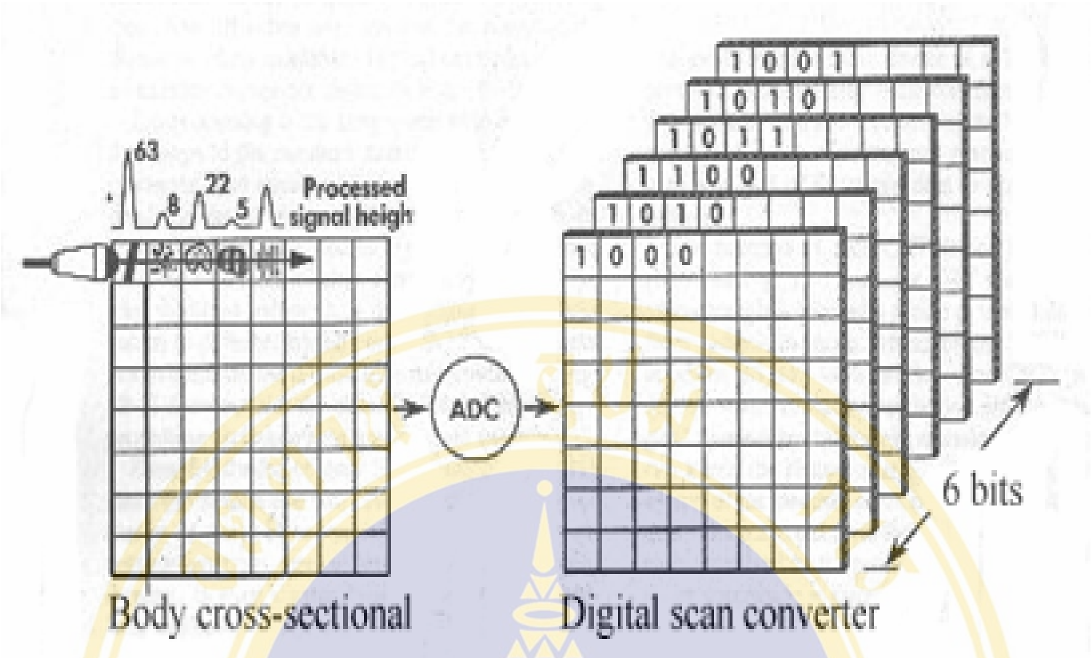


Figure 2.3 Transfer of information from the echo amplitude to the corresponding location in the digital scan converter. The echo amplitude signals are converted to binary numbers in analog to digital converter and sorted in 6-bit deep memory (35).

2.3 Mechanical effect (36)

The first effect to occur in body tissue as a result of ultrasound is a mechanical vibration. Sound vibrations require an elastically deformable medium for their propagation. The mechanical vibrations cause compression and expansion in the tissues at the same frequency as the ultrasound, leading to pressure variations in the tissue. Therefore the mechanical effect is also called “micromassage”. Due to the reflection of the sound beam at the tissue interfaces the intensity may be increased, so that the greatest pressure variations would occur at the boundary of two different tissues. Therefore one may assume the most pronounced therapeutic effects to occur at the boundaries. The pressure differences have consequently change in volume of the body cells of the order of 0.02%, permeability of the cell and tissue membranes, and improve exchange of metabolic product

Micromassage is great therapeutic importance because all effects of ultrasound therapy are caused by it. The effect occurs with both continuous and pulsed ultrasound, depending on the intensity used for treatment.

2.4 Thermal effect

Micromassage in the tissues can lead to generation of frictional heat. The amount of heat generated differs for the various tissues. It depends on a number of factors, some of which can be adjusted, e.g. the form of ultrasound (continuous or pulsed), the intensity and the duration of treatment. In addition, the absorption coefficient plays an important part. Lehmann (31) stated that the temperature in muscle tissues increases by 0.07 °c per second for continuous ultrasound of 1 W/cm^2. This value has been calculated for a muscle phantom, without the regulating effects of the bloodstream. Therefore, this appears to indicate the value of the maximum increase in temperature in muscle tissue. In a therapeutic situation involving ischemia such a marked rise in temperature would increase and lead to unfavorable effects. In a study on the medial side of the knee of pigs Lehmann (31) showed that an increase in temperature in the soft parts is relatively small in comparisons with the deeper articular structures. For continuous ultrasound at 1.5 W/ cm^2 of 5 minutes with a transducer plate of 12.5 cm^2 the average increase in temperature of the capsule is 6.3 °C, and the soft tissues is 3.3 °C. The medial part of the meniscus shows an average increase of 8.2 °C, whereas in the bony tissue an increase in temperature of 9.3 °C occurs.

The reflection and interference phenomena can lead to an increase in intensity, and therefore, heat can be more pronounced such as at bony tissue, cartilage and tendons. Heat is marked in the periostium and may lead to periostial pain.

The ultrasound beam is almost parallel, therefore, the area where the thermal effect occurs will approximate that of the treatment head. If heat is expected to have a favorable effect on the healing of lesions in the tissues, ultrasound therapy in continuous form should be used. It is important to ensure that the patient at most senses a small thermal effect. At a high intensity (more than 2 W/cm^2) with the continuous form of ultrasound, a marked increase in the blood circulation can be seen. It is noted that a lower intensity can also promote blood circulation. The significance of heat as a part of ultrasound therapy has been variously assessed. Many diseases are accompanied by disturbed circulation. The body is often incapable of dissipating the heat generated by ultrasound. This leads to an increase in temperature that may have an adverse effect on the disease. In case of an acute injury, for example, a sprain of the ankle, the heat generated (in combination with the mechanical irritation) may have an adverse effect on the recovering blood vessels. Bleeding may easily recur. Therefore it is advisable to wait a couple of days before local ultrasound therapy is started in such cases.

Lehmann (31) has shown that a rise in temperature is an important factor in the development of some physiological processes.

Bickford and Duff (37) sonicated 4 forearms with 0.8 MHz continuous ultrasound at 3.5 W/ cm^2 intensity for about 10-15 min. The temperature increased was found to be 1.8 °C at the skin, 0.6 °C at subcutaneous tissue, 1.8 °C at the muscle tissue 1.5 cm. depth, and 2.1 °C at the muscle tissue 3 cm. depth.

Lehmann (29) measured the temperature distribution in 20 volunteers, which divided into 2 groups according to the soft tissue thickness over the femur with the median value of 8 cm. A 1 MHz continuous ultrasound with 12.5 cm^2 radiating surface was applied by stroking technique using mineral oil as a coupling media. The mean temperature obtained from the tissue in front of the bone from the group of soft tissue thickness less than 8 cm. When sonicated by 1.0 W/ cm^2 for 9.22 min. the temperature was 41.73 °C and when sonicated by 1.5 W/ cm^2 for 2.4 min. was 40.61 °C. For the group of soft tissue thickness greater than 8 cm, the mean temperature of tissue in front of the bone sonicated by 1.0 W/ cm^2 for 15 min. and 1.5 W/ cm^2 for 7.96 min. were 41.53 °C and 42.06 °C, respectively.

Lele and Parker (38) investigated the temperature distribution in the tissue equivalent phantom; the cut thigh muscle of beef, and the dog muscle in vivo by using 4-6 thermocouples placed in various planes through the volume of interest. After irradiation by 0.9 and 1.8 MHz, continuous wave ultrasound, the peak temperature was attained over a very small volume and the axial location of the hot zone shift towards the surface at the higher frequency of ultrasound. Only in the case of dog muscle, which had the bone inside, there was a second hot region in the deeper tissue.

Kramer et al. (39) investigated only the subcutaneous tissue temperature change after 5 min. sonication by continuous wave ultrasound. For the placebo, the intensity 0.5 W/ cm^2, 1.0 W/ cm^2, and 1.5 W/ cm^2 the subcutaneous tissue temperature were decreased by 2.4 °C, 1.3 °C, and 0.1 °C respectively. Whereas after sonication with the 1.5, 2.0, and 2.5 W/ cm^2 intensity the subcutaneous tissue temperature were increased 0.7, 1.01, and 2.4 °C, respectively. They concluded that the decrease in subcutaneous tissue temperature was due to the cooling effect of the coupling media.

Enwemeka et al. (40) reported that the highest temperature changes in human thigh tissue 1, 2, and 3 cm. below the skin were 0.66 °C, 0.82 °C, and 0.91 °C respectively. These tissue temperature changes were observed at 10 minutes after sonication of 1 MHz, 1.0 W/ cm^2 continuous ultrasound.

According to these studies it may be concluded that each study had the specific intensity appropriated for each tissue that means any kind of tissue has different properties and each person has own tissue properties, such as reflection, refraction, acoustic property, blood circulation, etc., therefore, the appropriate intensity in one study may not be effective with another study.

2.5 Biologic effects

The effects of ultrasound therapy are all the result of micromassage (mechanical effect). Depending on the form, continuous or pulsed ultrasound, the micromassage results in predominance of either thermal or other effects. The following biologic effects can be seen as a physiological response to the mechanical and thermal effects mentioned.

- Promotion of blood circulation
- Muscle relaxation
- Increased membrane permeability
- Increased regenerative power of tissues
- Effect on peripheral nerves
- Reduction of pain
- Other effects

2.6 Other effects

Other effects have been observed as a consequence of ultrasound application. At presence, however the therapeutic significant cannot as their therapeutic significance is not clear, while many of the effects in this category are known to have a negative influence.

Tissue damage. Although the use of pulsed ultrasound can reduce the thermal effect it but because of a high temporal may cause a marked mechanical peak loading of the tissue. This may cause cavitation in the tissues and lead to tissue damage. Although the output of equipment in current use is such that this phenomenon can occur or not at all, it is wise to adjust the intensity such that the patient dose not sense a painful excitation.

Stasis of blood cells. The stasis of blood cells in the blood vessels running parallel to the ultrasound beam after application of ultrasound to chick embryos. The minimum energy at which this phenomenon still occurred was 0.5 W/ cm^2 of continuous ultrasound. This phenomenon is generally reversible. Continuous movement of the treatment head certainly appears sufficient to eliminate this phenomenon.

Other side effects believed that result from overdose are reduction of the blood vessel level, fatigue, nervousness, irritation, anorexia, constipation, and tendency to catch cold.

2.7 Tissue phantom

Tissue phantom is made of the materials that are equivalent to human tissue in acoustic properties, such as velocity and attenuation. There are two types of materials used for the construction of the phantoms, gel based or rubber/latex based. For example the RMI model 415@ is a rubber based phantoms and used for quality control measurements for diagnostic instruments. It has the speed of sound 1540 m/s and attenuation 0.5 to 0.7 dB/cm/MHz which is equivalent to liver tissue.

An ideal materials for the construction of ultrasound phantoms should have the same ranges of speed, attenuation coefficients, and scattering coefficients as real tissue. These parameters should be stable in the room temperatures and ease of storage. Speeds of sound in mammalian soft tissues vary over a fairly small range with an average value of approximately 1540m/s. It is thought to be approximately 1470m/s in fat and it ranges from 1570m/s in liver to 1600m/s along the direction of the fibers in muscle. Amplitude attenuation coefficients appear to vary over the range from 0.4dB/cm to about 2dB/cm at a frequency of 1 MHz in these tissues.

Perspex was chosen to be a rib cage phantom this because its acoustical impedance is similar to bone, and it is easy to get mold. All the organs were housed in the rib cage and sealed with silicon rubber, which simulated skin in terms of acoustical properties.

Wilfred et al. (42) manufactured a two layers tissue phantom for aberration study. The phantom was used for simulating fat and muscle layers in skin. The fat was simulated using PVC, which has a speed of sound 1395 m/s, attenuation coefficient 6 dB/cm at 4 MHz, and acoustic impedance 1.35 kg/m ^{2}s. The muscle was simulated using 65% water, 35% glycerol mixture by mass, which has speed of sound, attenuation coefficient 0.5 dB/cm at 4 MHz, and acoustic impedance 1.72 kg/m ^{2}s. Agar was used to solidify the muscle layer, and silica was used to, produce a speckle pattern of scattering tissue in the image.

Hynynen et al. (43) constructed the phantom for the temperature distribution studies which was prepared in a perspex box, 9×9×12 cm. A piece of sound absorbing carpet was placed in the bottom of the box and the top was covered with thin plastic film. The box was filled carefully with liquid gelatin-graphite mixture, having an attenuation coefficient of 10 Np/m, and rotated during congealing.

Ter Haar et al. (44) studied about a comparison of ultrasonic irradiation and RF inductive heating for clinical localized hyperthermia applications, designed a homogeneous tissue equivalent phantom used to demonstrate ultrasonic heating patterns. It was a modified gelatin model, rectangular in cross-section and of dimensions 28×15×6 cm. The attenuation coefficient is 0.46 Np/cm at 1 MHz. The RF phantom composed of water based gel, in cylindrical cross-section with a diameter of 20 cm, and a 1 cm surface layer of fat equivalent gel above a 4 cm layer of muscle equivalent gel.

Lele et al. (38) studied temperature distributions in tissues during local hyperthermia in the tissue equivalent phantom. It was a gelatin-albumin-agar mixture, 15 cm in diameter and up to 15 cm in height and was mounted on a thick aluminum plate to simulate presence of bone at that depth.

All the phantoms mentioned are mostly solid-like form, which are not suitable for this study because the temperature sensor probe cannot be inserted. Then the phantom in this study should be liquid-liked form. The materials acoustic properties near real tissue such as olive oil is , ethanol Amide, and concrete may be selected to represent the layer of fatty tissue, muscle, and bone respectively. The acoustic impedance of olive oil is 1.39×10^{6}-kg/m ^{2}s (45) compared to fatty tissue 1.34 - 1.40 $\times10^{6}$ kg/m ^{2}s (35,36,46), ethanol amide is 1.75×10^{6} kg/m ^{2}s (45) compared to muscle 1.60 - 1.71×10^{6} kg/m ^{2}s (35,36,46), and concrete is 6.9-10.4 $\times10^{6}$ kg/m ^{2}s (45) compared to bone 6.3 – 7.8 $\times10^{6}$ kg/m ^{2}s (35,36,46).

2.8 Temperature rises within phantom.

Jago et al. (47) studied the comparison by the calculated temperature rises from the tissue-mimicking model. This model consists of three layers as the phantom and uses all parameters from the phantom, such as absorption coefficient, the size of model, attenuation coefficient.

Most numerical models of tissue heating make use of a linear bioheat transfer equation.

$$\nabla^2 T - \frac{1}{K}\frac{\partial T}{\partial t} - \frac{T}{L^2} + \frac{q_v}{K} = 0 \qquad \text{……….. (2.1)}$$

T = temperature rise above the ambient level (°C)

q_v = heat source strength (the rate of heat production per unit volume, W/m^3)

K = thermal conductivity

L = perfusion length

t = time

The perfusion length L is a measure of the distance over which heat generated at one point has an appreciable effect on the temperature rise at another point.

The steady-state solution (i.e., for infinite t) for the temperature rise at a point a distance r from a point heat source of strength q_v is:

$$T = \frac{q_v}{4\pi}\frac{dV}{Kr}\exp\left(-\frac{r}{L}\right) \qquad \text{…………(2.2)}$$

It can be shown that, at any point in an acoustic field, provided the shear viscosity is negligible, the heat source function is given by:

$$q_v = \frac{2\alpha_a p_{TA}^2}{\rho c} \qquad \text{…………(2.3)}$$

p_{TA}^2 = temporal average of the square of the acoustic pressure

α_a = tissue amplitude absorption coefficient

ρc = acoustic impedance = density ρ multiplied by acoustic speed c

Jago et al. (48) tried to compare the AIUM/NEMA thermal indices with calculated temperature rises due to diagnostic ultrasound exposures for a simple third-trimester pregnancy model. The model, shown in Fig.4, assumes a 1.0-cm. thick layer, representing abdominal wall, with an attenuation of 0.5 dB/cm MHz (assume due to absorption only), a nonattenuating liquid layer representing maternal urine or amniotic fluid, and an absorbing bone target layer, representing perhaps fetal spine or skull.

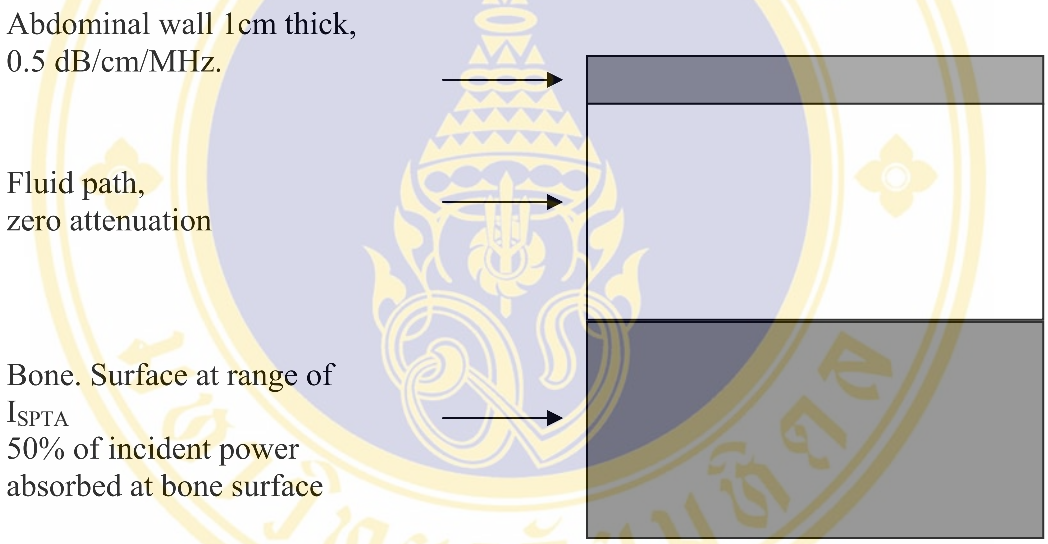


Figure 2.4 Schematic diagram of the third-trimester tissue model.

CHAPTER III
MATERIALS AND METHOD

3.1 Materials

3.1.1 Diagnostic Ultrasound Instrument

Toshiba Sonolayer SSA-250A (Japan) is the diagnostic ultrasound instrument used in this study equipped with a transducer of 3.75 MHz. In all measurements the ultrasonic beam was operated at non-focus mode and the overall gain was set up at 78.

Figure 3.1 Toshiba Sonolayer SSA-250A diagnostic ultrasound instrument

3.1.2 Therapeutic Ultrasound Instrument

Enraf Nonius Sonopuls 190 is the therapeutic ultrasound instrument used in this study. Both of continuous and pulsed ultrasonic beams can be generated at a frequency of 1 MHz. The set up is shown in appendix 1.

Figure 3.2 Enraf Nonius Sonopuls 190 therapeutic ultrasound instrument

3.1.3 Thermometry

A multi-channel Telethermometer YSI 400 series (Figure. 3.3) is used for all measurements of temperature in tissue- equivalent phantom. In this study the three thermistor probes type YSI 402 with the diameter of 3.2 mm. (Figure. 3.4) were used. An accuracy ±0.2°c and ±0.1°C can be expected at the temperature range 0°C to 60°C (°F to °F) and 32°C to 42°C (°F to °F) respectively.

Figure 3.3 Multi-channel Telethermometer YSI 400 series

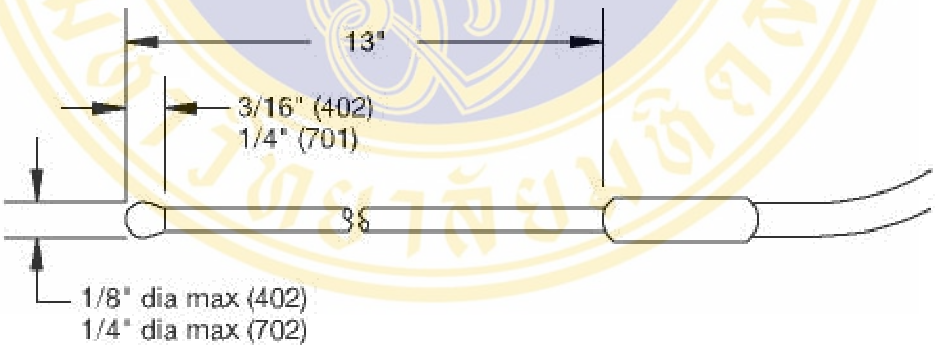


Figure 3.4 The dimension of YSI 402 rectal probe

All thermistor probes were calibrated to be assured that they can read the same temperature from 20°C to 50°C. The calibration had been made by putting each thermistor probe into water baths set at the temperature of 25°C, 30°C, and 35°C.

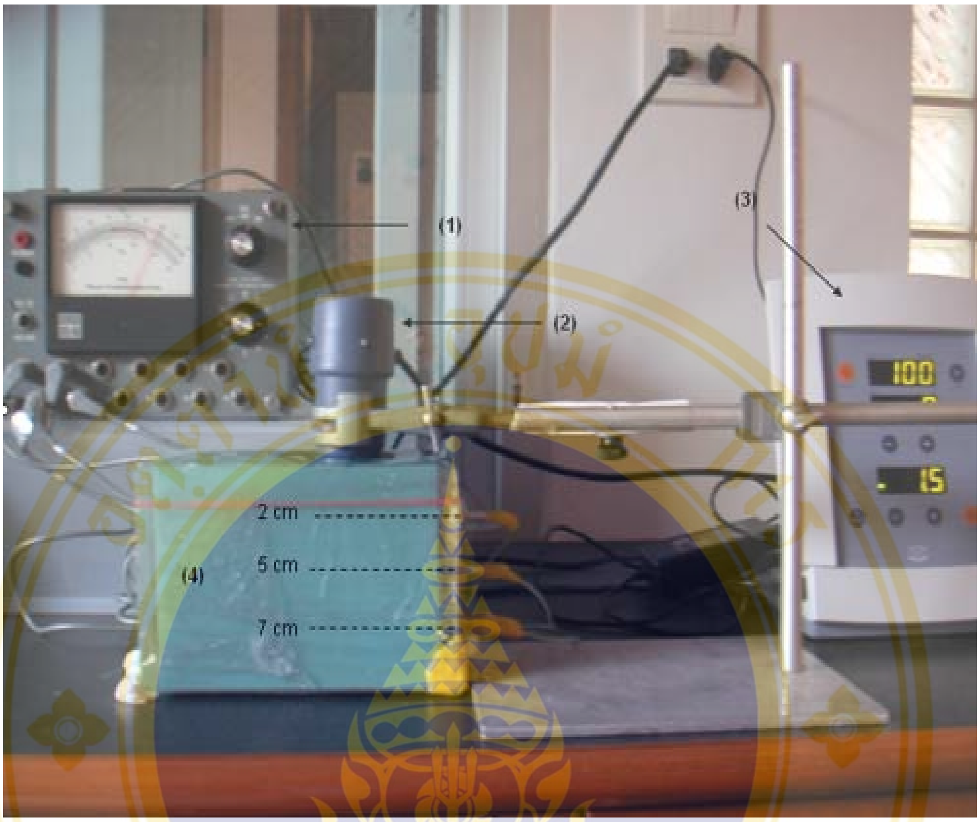


Figure 3.5 The set up for temperature measurement in phantom.

(1) Multi-channel Telethermometer

(2) Therapeutic ultrasound probe

(3) Therapeutic ultrasound instrument

(4) The depths of measurements in phantom

3.2 Preparation of Phantom

3.2.1. Composition of Tissue equivalent phantom

A tissue-equivalent phantom was designed and constructed. In terms of a tissue equivalent material means that it must has the equivalence of attenuation property and ultrasonic velocity. According to the literature review, the attenuation coefficient of human tissue is in the range of 0.3-2.0 dB/cm/MHz at the average velocity of 1540 m/s (41).

At the first place, ethanol amide was chosen to substitute the muscle tissue because their acoustic impedance equivalent. After testing an echo image it was found that there was no scattering pattern at all. This study needed the amplitude of echo signal to calculate the intensity ratio and the acoustic properties.

A custom built of liver tissue equivalent phantom for quality control of diagnostic ultrasound instrument (50) was then selected to develop further. The phantom was an elastic liked material made from the mixture of gelatin, non-dairy cream (coffee mate), floral cell (oasis flora foam), sodium benzoate, and water. The phantom attenuation coefficient is quoted to be 0.102dB/cm/MHz.

The gelatin was chosen due to its elastic property and the closed tissue density. The floral cells were expected to construct the scattering pattern of echo image. The non-dairy cream was mixed in order to make the uniform suspending of the floral cell in the mixture. Sodium benzoate was used for preventing the fermentation in the phantom. (49)

The appropriate composition of the tissue phantom was found by adjusting each content such as increase the quantity of gelatin as shown in table 3.1 for 8 different types of mixtures. The method of mixing the composition appears in the Appendix 2.

Table 3.1 The eight types of the mixtures of tissue phantom which are different in the quality of each composition.

Composition	Mixture number							
	1	2	3	4	5	6	7	8
Gelatin (g/cm^3 water)	500/1200	170/400	170/400	300/400	180/400	170/400	180/400	180/300
Coffee mate (g/cm^3 water)	250/600	85/200	160/200	85/200	300/200	85/200	300/400	350/500
Floral cell (g/cm^3 water)	25/600	8/200	13/200	8/200	8/200	8/200	8/100	15/100
Sodium benzoate (g/cm^3 water)	150/600	50/200	70/200	50/200	50/200	50/100	50/100	50/100
cement powder (g/cm^3 water)	0	0	0	0	0	150/100	0	0

The eight mixtures were investigated the attenuation property by applying the diagnostic ultrasound and measuring the echo signal in A-mode display. The attenuation coefficient by was calculated by the equation,

$$\frac{A}{A_0} = e^{-\alpha x} \qquad \text{.............................(3.1)}$$

where A, A_0 = the amplitudes of echo in cm measured along the slope of the attenuation corresponding to the depth x in the mixture

α = attenuation coefficient at diagnostic ultrasound frequency (Np/cm/MHz)

x = depth at the point of interest (cm)

3.3 Ultrasonic Properties of phantom

3.3.1 Density

The density of tissue phantom (ρ) was calculated by ρ = mV where the mass (m) in g/cm^3 and the volume (V) in cm^2 were measured from the definite size of mixture investigated.

3.3.2 Test for attenuation coefficient

To evaluate the attenuation coefficient of the interested mixture, it was done by mixing and pouring the mixture into the 13 containers. Each container had a thin aluminum plate with the same size as the container placed at 1 cm different depth respectively, as shown in Figure 3.6. The echo signal at the aluminum interface in A-mode display of each container was recorded. The amplitude of all echo signals were measured, recorded and plot against the depth of aluminum plate in logarithmic scale. The attenuation coefficients of the interested mixtures were calculated by means of the equation 3.1.

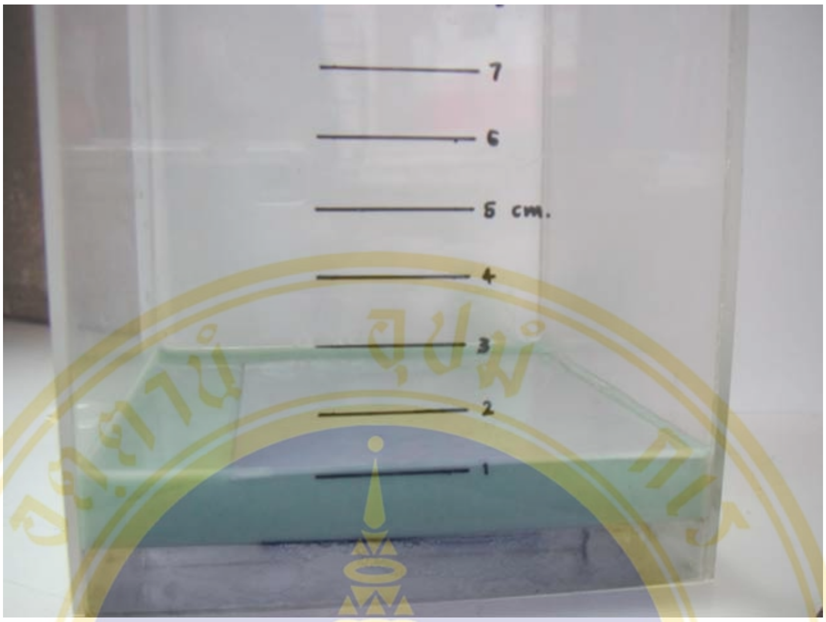
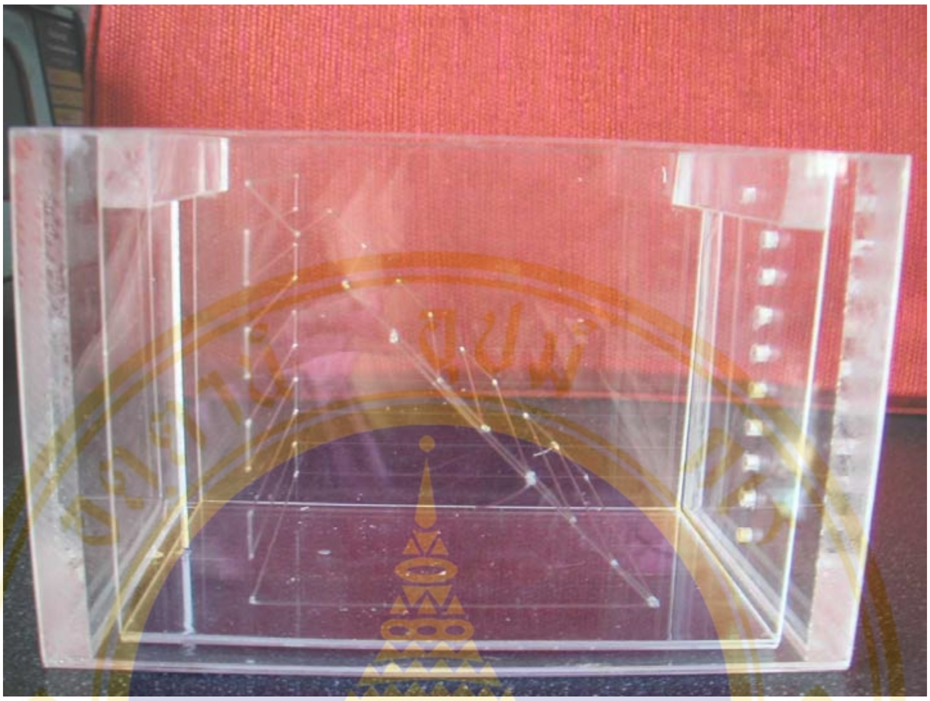


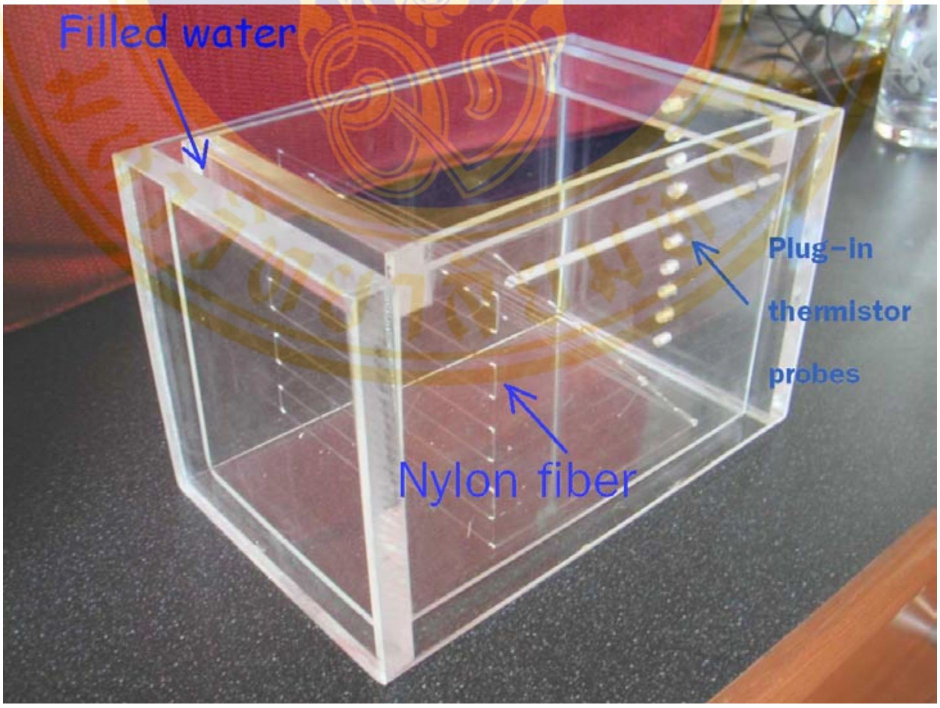
Figure 3.6 Aluminum plate placed at 12 cm depth from the top of phantom surface.

3.3.3 Relationship between profile of echo signal and the state of tissue phantom.

The echo signals were investigated and recorded during the temperature of tissue phantom was changed by heating. A designed phantom having a set of nylon fiber (0.75 mm in diameter) lived vertically at 1 cm apart from top to bottom (Figure 3.7) was used to fill up with the solid state tissue material. The phantom was immersed in a water bath. The diagnostic ultrasound head was fixed firmly perpendicular to the phantom surface and covered the group of nylon fibers. The change of echo profiles before and after heating were investigated and recorded every 5 minutes until the tissue phantom was completely melted. The echo profiles were then compared in terms of the slope changed.



a) The phantom box



b) Side view

Figure 3.7 (a) The phantom box.,
(b) The position of nylon fibers

3.3.4 Temperature distribution in liquid stated and solid stated phantoms

To measure the temperature distribution the phantom was set up as shown in Figure 3.4. From the phantom in Figure 3.6, three thermistor probes were inserted at the depths of 3, 6, and 9 cm from the surface (Figure 3.8). A therapeutic ultrasound head of 1 MHz was fixed firmly in contact with the phantom surface. The temperature distributions were measured separately in both liquid and solid stated tissue phantoms. The therapeutic ultrasonic intensity was set at 1.5 W/cm^2 for all measurements. The temperatures were recorded at every 2 minutes for 20 minutes duration.

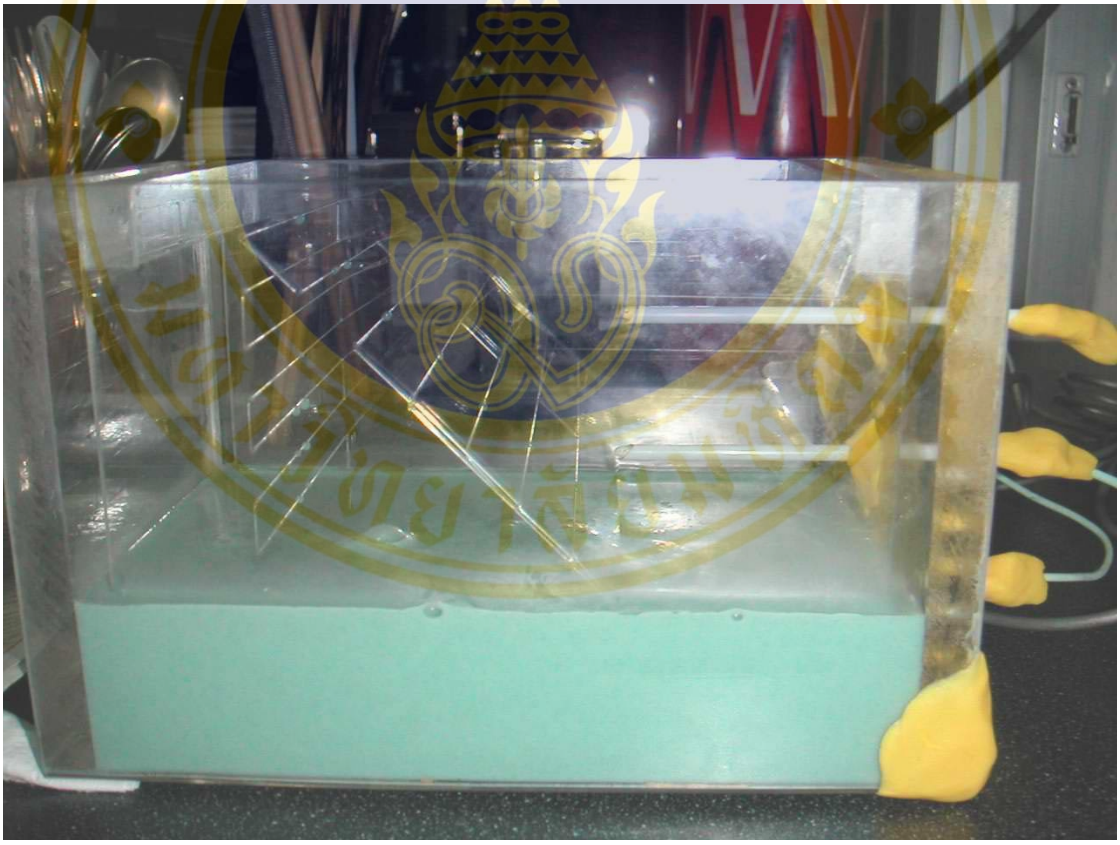


Figure 3.8 The insertion of 3 thermistors into the liquid state phantom

3.3.4 Comparison of the measured temperatures in phantom with the calculated temperatures.

The temperature raised in phantom during therapeutic ultrasound can be calculated from the bio-heat transfer equation.

$$Q'''_t - Q'''_c = \rho c \frac{dT}{dt} \quad \text{..............(3.2)}$$

where Q'''_t = heat per unit volume by therapeutic ultrasound

Q'''_c = heat per unit volume by conduction

$\frac{dT}{dt}$ = the rate of temperature raised

ρ = density of the tissue phantom

c = heat capacity of tissue phantom

CHAPTER IV
RESULTS

4.1 Ultrasonic properties of phantom

4.1.1 Attenuation coefficient

It is found that the attenuation coefficient of tissue phantom varies with the composition of the mixture as shown in Table 4.1. After measuring the echo signal of each mixture, the attenuation coefficient is calculated by the slope of echo pattern as shown in Figure 4.1.

Table 4.1 The mixtures of tissue phantom

Composition	Mixture number							
	1	2	3	4	5	6	7	8
Gelatin (g) /water(cm^3))	500/ 1200	170/ 400	170/ 400	300/ 400	180/ 400	170/ 400	180/ 400	180/ 300
Coffee mate (g) /water(cm^3)	250/ 600	85/ 200	160/ 200	85/ 200	300/ 200	85/ 200	300/ 400	350/ 500
Floral cell (g) /water (cm^3)	25/ 600	8/ 200	13/ 200	8/ 200	8/ 200	8/ 200	8/ 100	15/ 100
Sodium benzoate (g) /water (cm^3)	150/ 600	50/ 200	70/ 200	50/ 200	50/ 200	50/ 100	50/ 100	50/ 100
Cement powder (g) /water (cm^3)	0	0	0	0	0	150/ 100	0	0
Attenuation coefficient (dB/cm/MHz)	0.194	0.091	0.235	0.272	0.313	0.222	0.313	0.4

The attenuation coefficient is calculated from the equation,

$$\frac{A}{A_0} = e^{-\alpha x} \quad \text{.......................................(4.1)}$$

Therefore, $$\alpha = -\frac{\ln \frac{A}{A_0}}{x} \quad \text{....................................(4.2)}$$

Where A_0 and A = amplitude of echo signal in cm. at the distance along slope of echo pattern x cm., and α is the attenuation coefficient in Np/cm/MHz.

For example, if we substitute the data in Figure 4.1 into the equation (4.2);

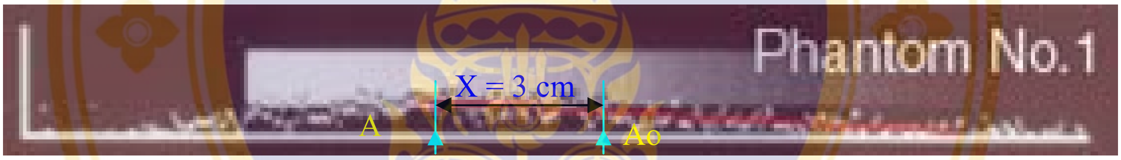


Figure 4.1 Echo pattern of mixture No.1 in Table 4.1 and parameters from the echo signals

$$\alpha = -\frac{\ln \frac{0.7}{0.9_0}}{3}$$

= 0.083 × 8.686 / 3.75MHz.

= 0.194 dB/cm/MHz.

The variations of echo patterns in the same tissue phantom with temperature in phantom changed are shown in Figure 4.2. Resulting from the calculation of attenuation coefficients, it is found that the attenuation coefficient changed from solid state to liquid state phantom varies from 0.26 to 0.28 dB/cm/MHz. The data of calculations are shown in Table 4.2.

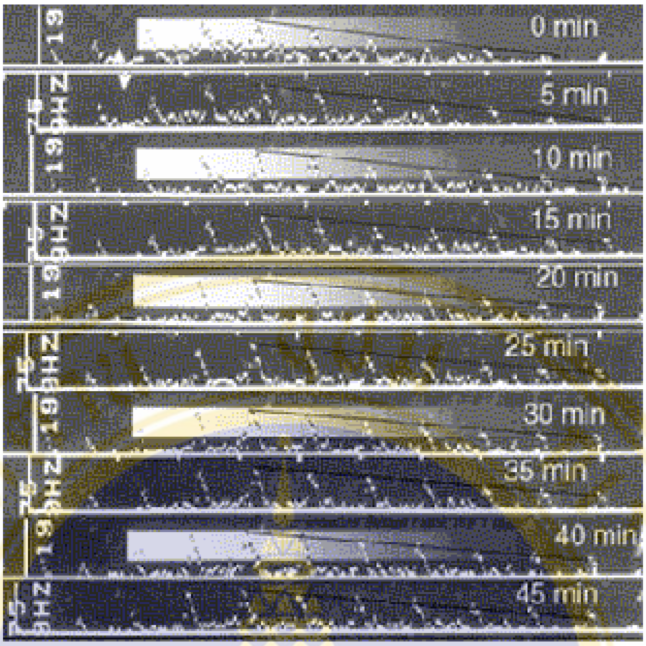


Figure 4.2 The patterns of echo signals in the same tissue phantom measured at every 5 minutes during heating started from solid state phantom. It is noted that the tissue phantom is completely melt after 45 minutes of heating. (The original images are shown in Appendix 3.)

Table 4.2 The echo signal measured values for calculation and comparison of the attenuation coefficient (α, dB/cm/MHz) every 5 minutes during heating.

Time (min)	A(cm)	Ao(cm)	X(cm)	∝(dB/cm/MHz)
0	1	0.7	3	0.28
5	1	0.7	3	0.28
10	1	0.7	3	0.28
15	1	0.7	3	0.28
20	1.05	0.75	3	0.26
25	1	0.7	3	0.28
30	1.05	0.75	3	0.26
35	1	0.75	3	0.26
40	1	0.7	3	0.28
45	1	0.7	3	0.28

From using echo pattern for calculation of attenuation coefficient as described early, the random error may occur from the reverberation of the echoes between the nylon fibers. It is therefore, designed to use the echo from a strong reflecting material (aluminum plate) at the different depths in separated phantom instead. The echo signals from aluminum reflector at different depths are shown in Figure 4.3.

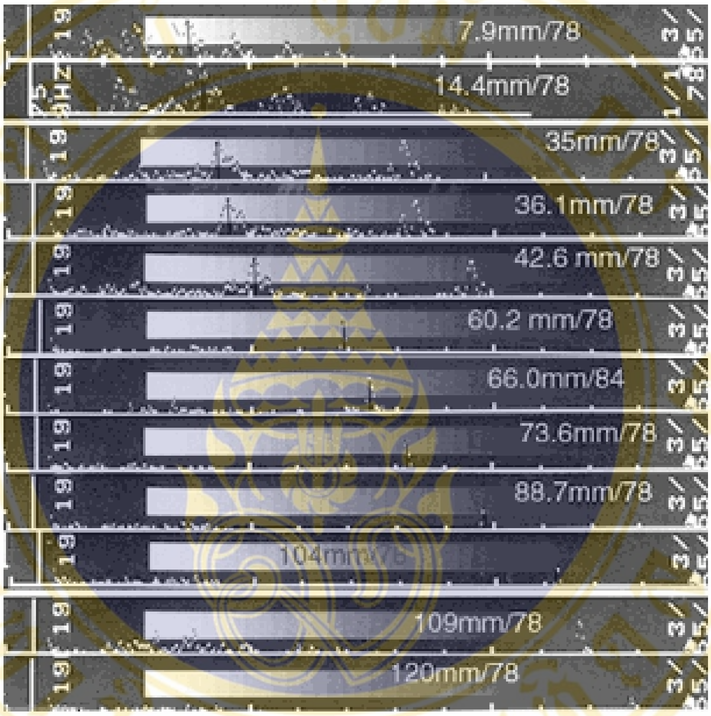


Figure 4.3 Representation the amplitude of echo signal at different depths measure from 10 separated phantom of the same tissue material.

The amplitude values of echo signals (D) corresponding with depths (H), are shown in Table 4.3 and also plotted in Figure 4.4. The decreasing of echo amplitudes after the peak at 42.6 mm. depth results from the ultrasonic attenuation in tissue phantom. (The original images are shown in Appendix 4.)

Table 4.3 The amplitude (cm) was measured at different level in 10 phantoms, D is distance from phantom surface and H is amplitude height of echo signal.

Phantom number	D (cm)	H (cm)
1	0.79	0.9
2	1.44	1
3	3.5	1.1
4	3.6	1.15
5	4.26	1.3
6	6.02	0.85
7	7.36	0.75
8	8.87	0.5
9	10.4	0.4
10	12	0.3

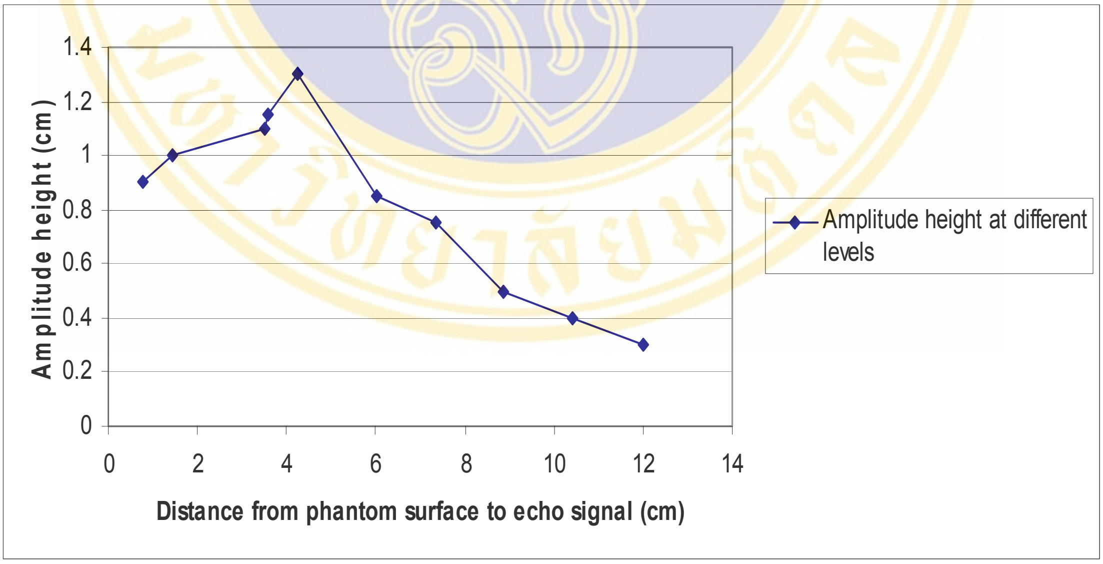


Figure 4.4 Representation of amplitude signal (D, cm) at different levels of 10 phantoms which are the same mixture.

The attenuation of echo amplitudes in tissue phantom is fitted with the equation $y = 2.8e^{-0.1877x}$ with a correlation R^2 = 0.9912, where y is the echo amplitude at the depth x from the phantom surface.

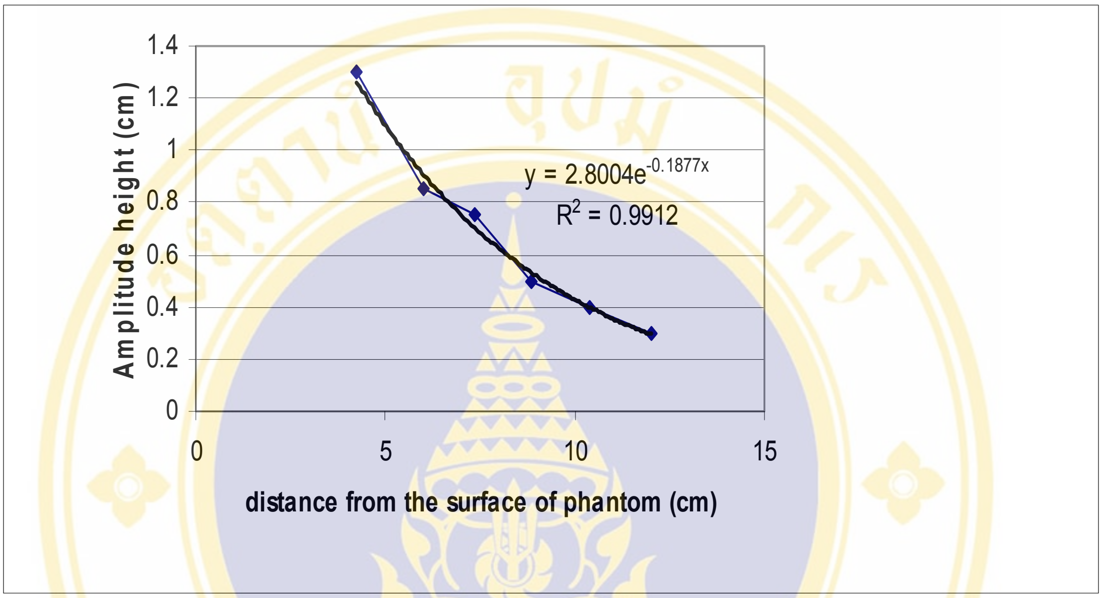


Figure 4.5 Representation of the exponential graph, exponential equation, and R^2 of mixture No.8 phantom.

From the fitted exponential equation, it can be said that the attenuation coefficient of the tissue phantom is 0.1877 Np/cm/MHz or 0.43 dB/cm/MHz.

4.1.2 Density

The density of the tissue phantom can be calculated from: ρ = m/V where m is the mass in gram of tissue equivalent material of volume V in cm^3. The densities of five different size of tissue phantom are shown in Table 4.4. The average density is 1.11 g/cm^3.

Table 4.4 The densities of five different mass and volume of tissue phantoms

Measured parameters	Number of example				
	1	2	3	4	5
mass (g)	342	560	109	508	374
volume (cm^3)	300	500	100	450	350
density(g/cm^3)	1.14	1.12	1.09	1.13	1.07

4.2 Temperature distribution in liquid state phantom and solid state phantom

4.2.1 Temperature distribution in liquid state phantom

The temperature distribution was recorded at the depth 3, 6, and 9 cm. represented as T_1, T_2, and T_3 respectively. The temperature data are shown in Table 4.5 and plotted in Figure 4.6. The rapid temperature raised at the first few minutes of ultrasound and then slowly with depths can be seen for all depths investigated.

Table 4.5 The measured temperatures at the depth 3 cm (T_1), 6 cm (T_2), and 9 cm (T_3) in liquid tissue phantom during therapeutic heat with ultrasound 1.5 W/cm^2 for 20 minutes.

Time (minutes)	T1(°F)	T2(°F)	T3(°F)
0	87.1	87.5	87.5
2	107	99	93
4	107.5	99.3	94
6	109.5	99.8	94.2
8	111.5	101.2	94.3
10	113.7	102.7	94.8
12	116.5	103.8	95.2
14	118.2	104.6	95.7
16	119.8	105.3	96.2
18	121.5	106.2	97.1
20	122	107	98

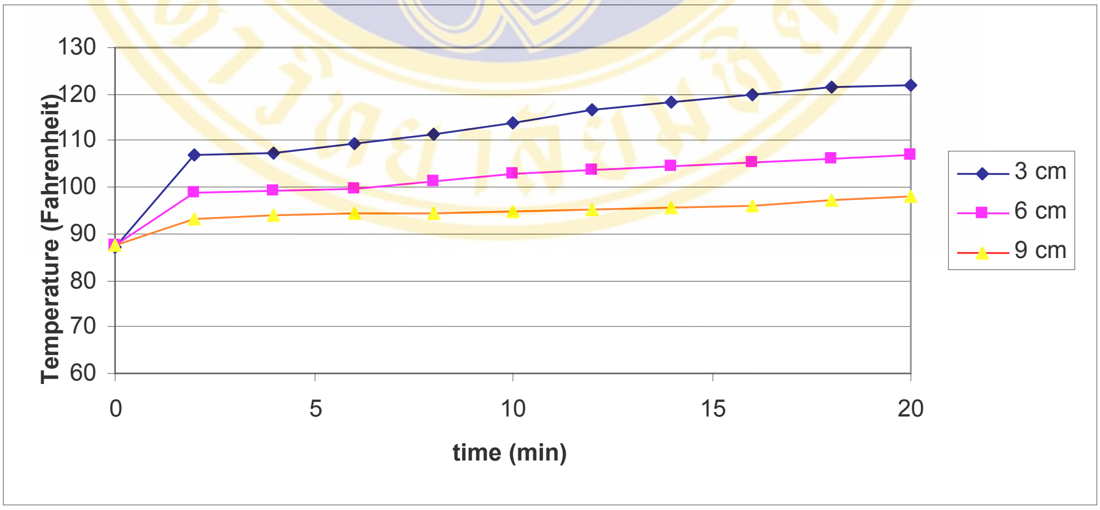


Figure 4.6 The temperature rises in liquid tissue phantom at the depth of therapeutic heat with ultrasound 1.5 W/cm^2 for 20 minutes.

4.2.2 Temperature distribution in solid state phantom

The temperature distribution was recorded at the same set up as described in liquid phantom. The temperature distribution is shown in Table 4.6 and Figure 4.7. The rapid temperature rises at the first few minutes of ultrasound can be seen all depths investigated expected at 9 cm. At the depth 9 cm the temperature change can not be seen.

Table 4.6 The measurement temperatures at the depth 3 cm (T_1), 6 cm (T_2), and 9 cm (T_3) in solid tissue phantom during therapeutic heat with ultrasound 1.5 W/cm^2 for 20 minutes.

Time(minutes)	T1(°F)	T2(°F)	T3(°F)
0	73	70.2	70.2
1	85	77	71.3
2	87	80	70
3	90	83.2	71.3
4	105	85	72.5
5	107	86.6	72.5
6	109	88.5	72.6
7	111.3	98	72.6
8	113.7	102	72.6
9	115.2	103.7	72.6
10	116	104.8	72.8
11	118.1	106.8	72.8
12	120	109.3	72.8
13	-	110.2	72.8
14	-	111.8	72.8
15	-	112.4	72.9
16	-	114.3	72.9
17	-	115	73.1
18	-	116.6	73.1
19	-	118	73.1
20	-	118.5	73.2

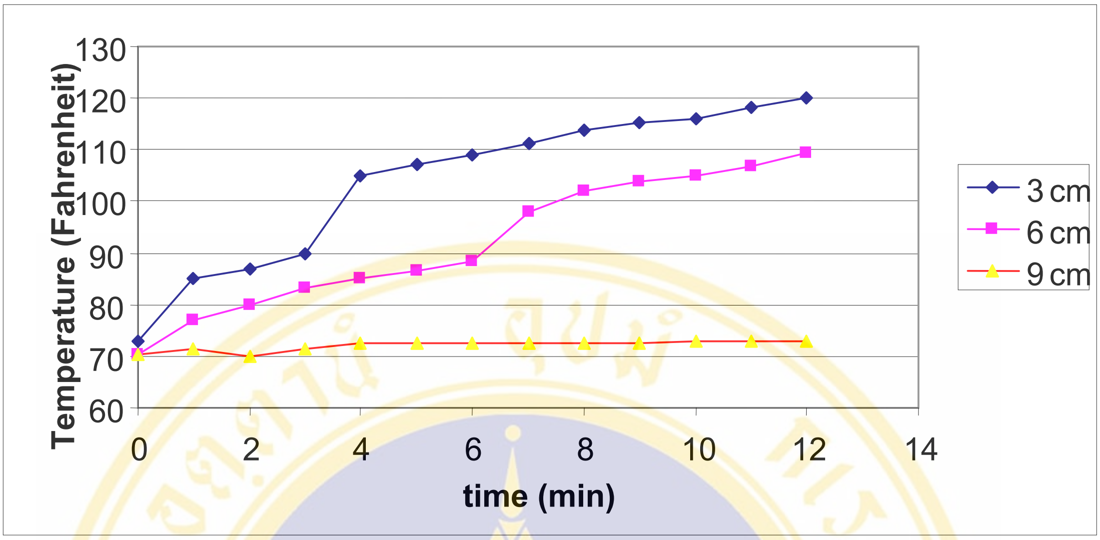


Figure 4.7 The temperature rises in solid tissue phantom at the depth of therapeutic heat with ultrasound 1.5 W/cm^2 for 20 minutes.

The start up temperature before heating of both stated phantoms are not equal. It is about 87.1-.87.5 Fahrenheit in liquid state phantom, and 70-73 Fahrenheit in solid state phantom. Temperature rise after heating for 2 minutes of liquid state phantom and solid state phantom is very high, as an initial state it had no conduction effect in tissue phantom. After heating for 12 minutes the temperature of both phantoms tends to have a resting state. At 9 cm level of both stated phantoms, the temperature would not be significantly changed. It can be said that there is less effect of therapeutic ultrasound at this depth.

4.3 Heat conduction in tissue phantom

From the same measurement set up as described in the temperature distribution in the liquid and solid state phantom, after therapeutic heat for 20 minutes then the ultrasound is stopped and the temperature is continued to record until the temperature of tissue phantom is closed to the ambient temperature. The recorded temperatures at 3, 6, and 9 cm depths are shown in Table 4.7 and 4.8 and plotted in Figure 4.8 and 4.9 for the liquid and solid state phantom respectively.

Table 4.7 The recorded temperatures in liquid state phantom during heating with ultrasound 1.5 W/cm^2 for 20 minutes and after heating.

Time (min)	T1 at 3 cm (□F)	T2 at 6 cm (°F)	T3 at 9cm (°F)
0	87.1	87.5	87.5
2	107	99	93
4	107.5	99.3	94
6	109.5	99.8	94.2
8	111.5	101.2	94.3
10	113.7	102.7	94.8
12	116.5	103.8	95.2
14	118.2	104.6	95.7
16	119.8	105.3	96.2
18	121.5	106.2	97.1
20	122	106.4	98
22	98	95	95.2
24	96.3	94.1	94
26	95.2	93.3	93.1
28	94.7	92.7	92.5
30	94.5	92.2	92.1
32	94.1	92	91.8
34	94	91.8	91.5
36	94	91.7	91.2
38	93.7	91.5	91.2
40	93.4	91.2	91
42	93.4	91.2	90.7
44	93.1	91.2	90.7
46	93	91	90.5
48	93	91	90.5
50	92.8	91	90.3
52	92.7	91	90.2
54	92.5	90.9	90.2
56	92.3	90.8	90.1
58	92.2	90.8	90.1
60	92.2	90.8	90.1
62	92	90.8	90
64	92	90.8	90
66	91.8	90.5	89.9
68	91.7	90.5	89.9
70	91.5	90.5	89.8
72	91.5	90.5	89.8
74	91.5	90.5	89.8
76	91.5	90.5	89.8
78	91.3	90.5	89.8
80	91.3	90.5	89.8
82	91.3	90.5	89.8
84	91.1	90.3	89.8
86	91.1	90.3	89.8

Table 4.8 The recorded temperatures in solid state phantom during heating with ultrasound 1.5 W/cm^2 for 20 minutes and after heating.

Time (min)	T1 at 3 cm	T2 at 6 cm (°F)	T3 at 9cm (°F)
0	73	70.2	70.2
2	87	83	70
4	98	85	72.5
6	107	98.5	72.5
8	109.5	102	72.6
10	113.5	104.8	72.8
12	122	108.7	72.8
14	90	83.2	72.5
16	87.1	81	72.5
18	85	79.3	72.5
20	83.8	78.5	72.5
22	82.6	77.7	72.5
24	81.8	77	72.8
26	81	76.8	72.8
28	80.3	76.2	72.8
30	80	76.1	72.8
32	79.6	75.8	72.8
34	78.8	75.5	73
36	78.8	75.5	73
38	78.3	75.3	73
40	78	75.2	73
42	77.9	75.1	73.2
44	77.8	75	73.2
46	77.6	75	73.2
48	77.6	75	73.3
50	77.5	75	73.5
56	77.3	75	73.7
62	77.2	75.2	74
70	77.2	75.2	74.2
80	77.2	75.5	74.5
114	77.2	76.8	76.2

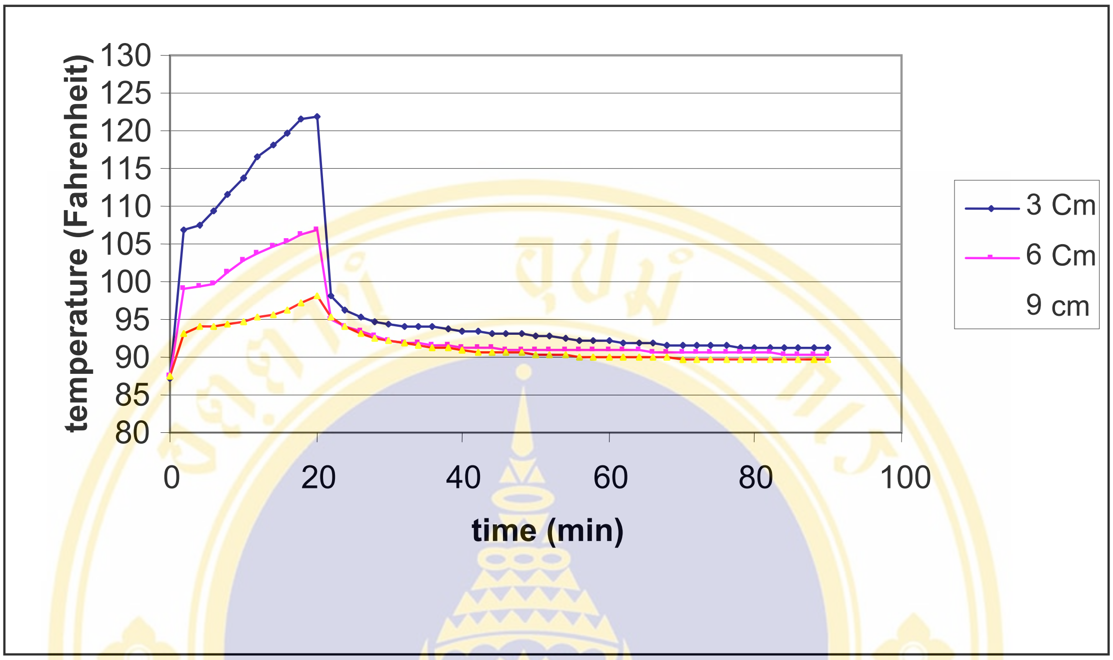


Figure 4.8 Temperature distribution of liquid state phantom at 3, 6, and 9 cm depths measured during heating with 20 minutes of therapeutic ultrasound and after heating reaching the ambient temperature.

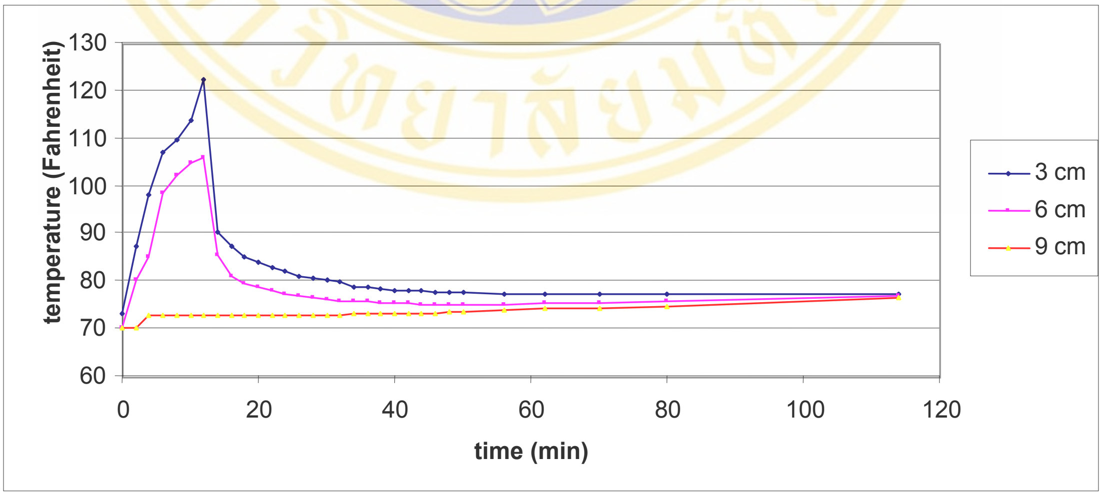


Figure 4.9 Temperature distribution of solid state phantom at 3, 6, and 9 cm depths measured during heating with 20 minutes of therapeutic ultrasound and after heating reaching the ambient temperature.

It can be seen from Figure 4.8 that at initial state (about 2 minutes after sonication) which has less effect of conduction, the temperature raised may due to ultrasound only. After that rate of temperature rise is less because of the conduction effect. If there is no conduction effect, the temperature would be increase at the same rate as initial state all the time. However, the measured temperatures at this depth after 2 minutes shown in Figure 4.8 indicate the lower temperatures. Therefore, it should be caused by the conduction effect.

4.4 Temperature distribution in 2 layers phantom heated with 1.5 W/cm^2 therapeutic ultrasound.

The two layers phantom consisted of the tissue phantom and fresh bone. The bone layer surface is set up at 7 cm depth. The thermistors are located at the depth of 2 cm (T_1), 5 cm (T_2), and 7 cm (T_3) as shown in Figure 4.10. The temperature distributions before and during therapeutic application are shown in Table 4.0 and plotted in Figure 4.11 and 4.12,

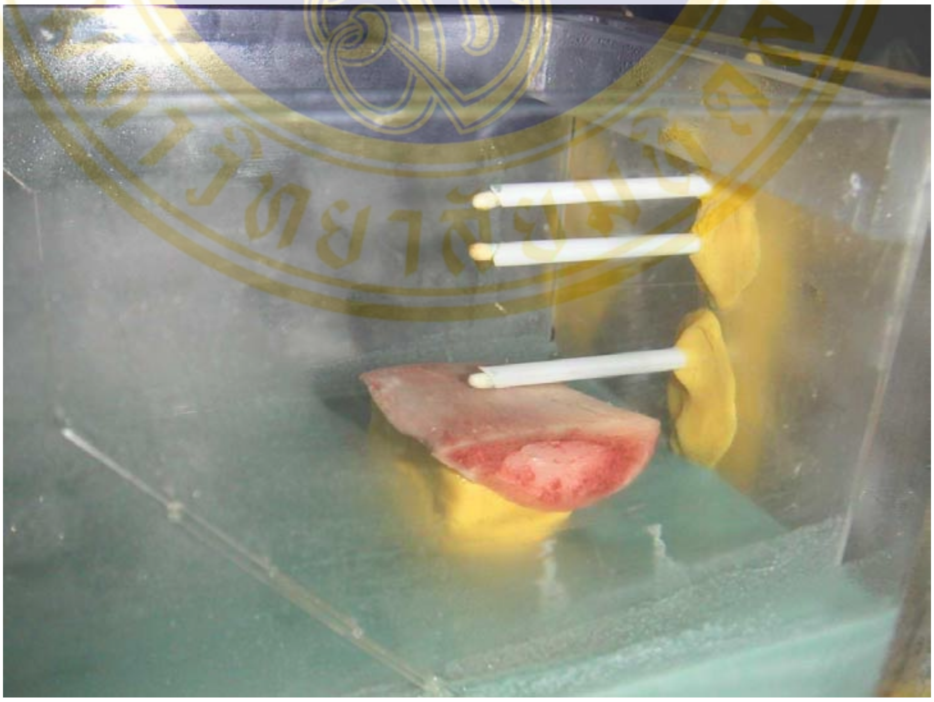


Figure 4.10 Thermistors located at the depth of 2, 5, and 7 cm (above bone surface).

Table 4.9 The temperature distribution in 2 layers phantom measured before and during with 1.5 W/cm^2 therapeutic application for 20 minutes at the depths of 2, 5, and 7 cm. The measurement is repeated with different started tissue phantom temperature.

Time (min)	Experiment No.1			Experiment No.2		
	T1 at 2 cm (°F)	T2 at 5 cm (°F)	T3 at 7 cm (°F)	T1 at 2 cm (°F)	T2 at 5 cm (°F)	T3 at 7 cm (°F)
0	70	73.2	73	71.3	71.8	70.8
1	70	73.5	73.5	79	85.5	71.5
2	70.5	74.5	74	82	85	71.8
3	71.1	76.5	74	80.7	88.5	72.5
4	72.3	78.7	75.4	81.2	91	72.5
5	73	84.5	76.8	84	95	73.5
6	73.8	83.1	78.8	85.5	95.5	74
7	74	81.5	79	85.5	99	75
8	74.5	87	82.5	89	99.8	75.2
9	76	93	84.5	91	95	75.8
10	77.5	94	90.5	91.3	101.2	76.1
11	74.5	92.3	88.6	94	97.2	76.6
12	75	94.6	91	95	104.5	76.5
13	76.7	96	93.4	98	105.2	77.5
14	75.6	97.5	95	98.2	108.5	77.8
15	76	110	102.5	98	105.2	79.2
16	76.2	98.5	96.5	97.5	108.5	80
17	76	97.5	94	99.2	111	81.3
18	75.5	96.2	92.7	99.2	111.5	82
19	76	97	95	99.8	112	81.7
20	76.8	99.5	102	100.2	113	83

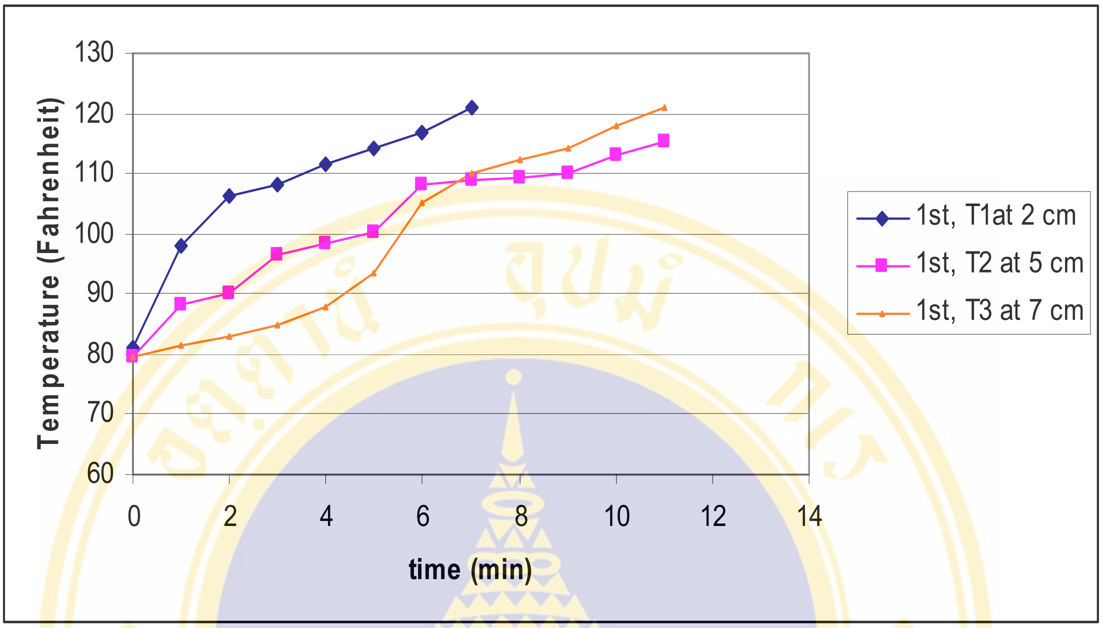


Figure 4.11 The temperature distribution at the depths of 2, 5, and 7 cm in the 2 layers tissue phantom during heating with 1.5 W/cm^2 therapeutic ultrasound.

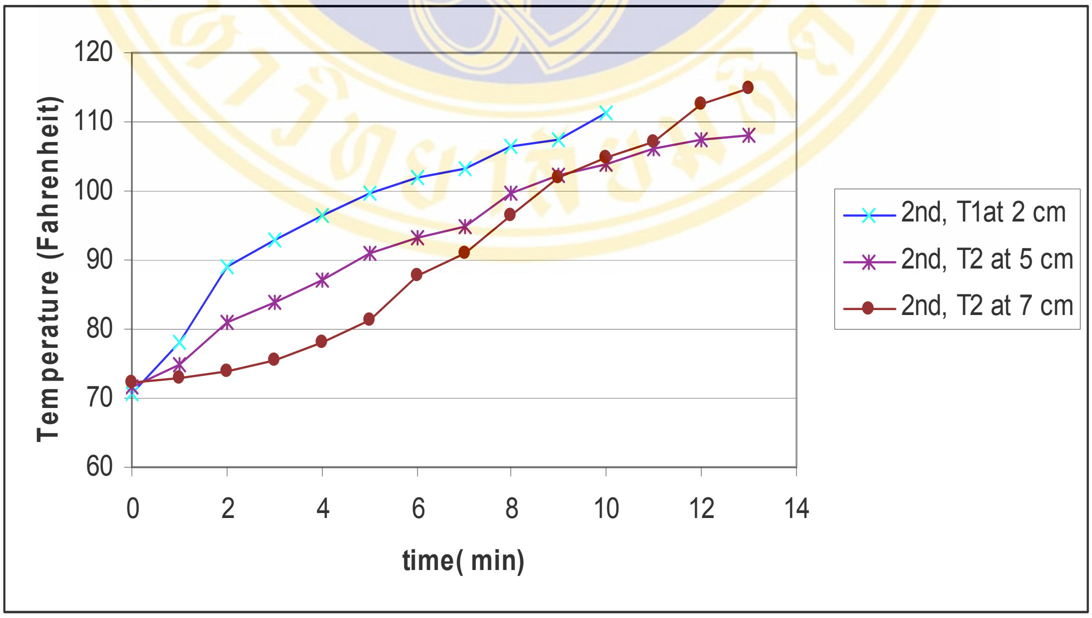


Figure 4.12 The temperature distribution resulting from the repeated measurements in the two layers phantom with different started phantom temperature.

As a result from Figure 4.11 and 4.12, the similar trend of temperature distribution can be seen. It one compares the temperature distribution to the experiment of heat conduction in 4.3, the trend of the graphs look similar expected for the 7 cm depth (the bone surface). After therapeutic application for 5 minutes, the temperature raised at the bone surface is more pronounced. It means that there is the effect of accumulated heat in bone transferred to tissue layer.

4.5 Calculation of therapeutic ultrasound intensity and temperature distribution during therapeutic ultrasound application in the tissue phantom.

The therapeutic ultrasound intensity in the tissue phantom can be calculated from the attenuation coefficient by the equation,

$$\frac{I}{I_0} = e^{-2\alpha x} \quad \text{.............................. (4.3)}$$

where I = intensity at point of interest (W/cm^2)

I_0= intensity at the phantom surface (1.5 W/cm^2)

$\propto$= attenuation coefficient resulting from 4.1.1 (0.05 Np/cm/MHz)

x= distance between the phantom surface and the point of interest (cm)

Equation (4.3) is valid only for the plane wave ultrasound.

From the pattern of echo amplitudes shown in Figure 4.4 indicates interference effect up to the depth about 4.3 cm in tissue phantom. Therefore, equation (4.3) can be applied for the tissue phantom after the depth of 4.3 cm.

In this study, the intensity of the therapeutic ultrasound is used at 1.5 W/cm^2. From Figure 4.7 there is no temperature raised at the depth 9 cm, therefore, the ultrasound may not effect at this depth and beyond.

From the limitation described above, the temperature distribution at the depth 6 cm are calculated and compared with the measurement with the thermistor.

The temperature (T_c) at the depth 6 cm at the therapeutic time (t) can be calculated from the bio-heat equation (3.2). For example, to calculate the temperature at the depth 6 cm at therapeutic time 4 minutes;

From Equation (3.2); $$Q'''_t - Q'''_c = \rho c \frac{dT}{dt}$$

t = 4 min; $$2\alpha I_{at\,6\,cm} - \rho c(\frac{\Delta T}{dt}) = \rho c(\frac{dT}{dt})$$

where $I_{at\ 6\ cm}$ is calculated according to (4.3)

Substitution;

$$2(0.05)(0.82) - 1.11(1.3)(\frac{41.33-35}{2\times 60}) = 1.11(1.3)(\frac{T_{at\,4\,min} - 37.22}{2\times 60})$$

$$0.082 - 0.075 = 0.012\ T_{at\ 4\ min.} - 0.012(37.22)$$

$$T_{at\ 4\ min} = 37.83\ °C$$

$$= 100.094\ °F$$

Table 4.10, 4.11 and Figure 4.13, 4.14 shows the comparison of temperature distributions at the depth 6 cm at various therapeutic times between calculated and measured temperatures in the liquid and solid state phantoms respectively. A closed agreement between measured and calculated temperature distributions can be seen in both liquid and solid phantom at the depth 6 cm.

Table 4.10 Results of measured and calculated temperatures in liquid state phantom at therapeutic times up to 20 minutes.

Time (min)	T2 (°F at 6 cm) measured	T2 (°F at 6 cm) Calculated
0	87.5	87.5
2	99	99
4	99.3	100.094
6	99.8	101.156
8	101.2	102.2
10	102.7	103.24
12	103.8	104.29
14	104.6	105.33
16	105.3	106.38
18	106.2	107.42
20	106.4	108.46

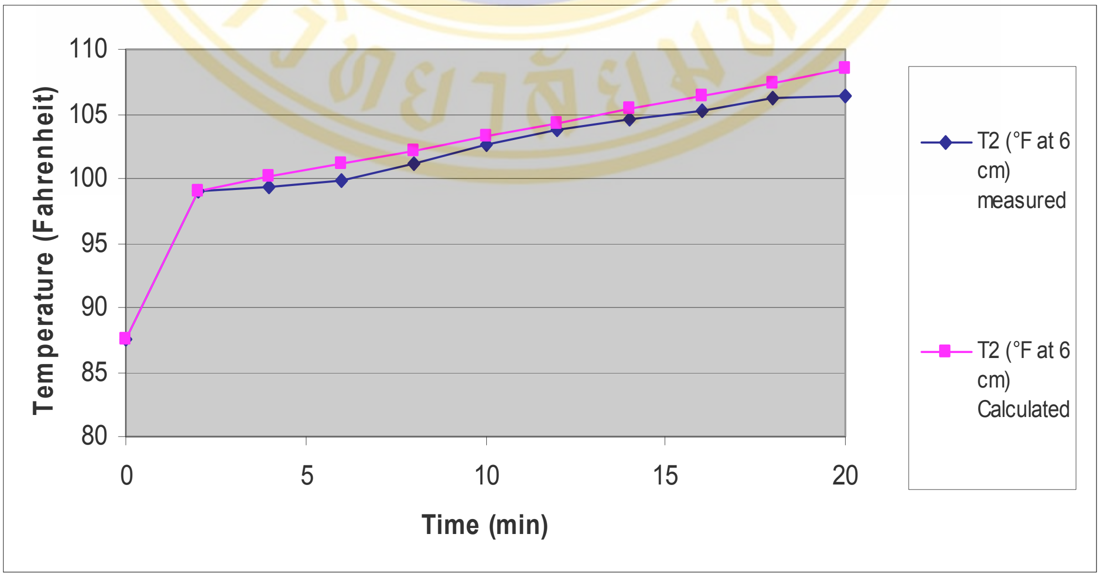


Figure 4.13 A comparison of measured and calculated temperatures in liquid state phantom at various therapeutic times.

Table 4.11 Results of measured and calculated temperatures in solid state phantom at therapeutic times up to 20 minutes.

Time (min)	T2 (°F at 6 cm) measured	T2 (°F at 6 cm) calculated
0	70.2	70.2
2	80	80
4	85	85
6	98.5	98.5
8	102	104.1
10	104.8	109.7
12	108.7	115.3

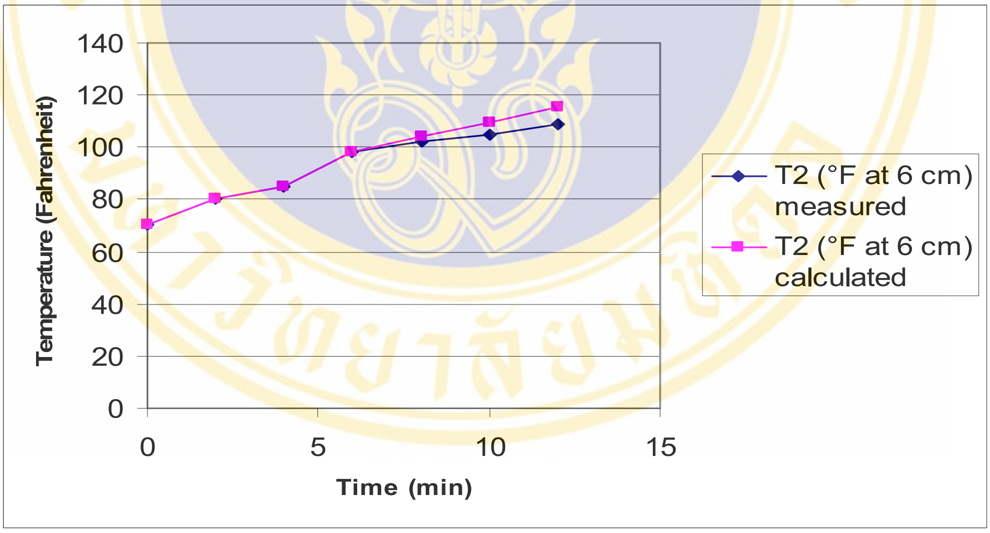


Figure 4.14 A comparison of measured and calculated temperatures in solid state phantom at various therapeutic times.

CHAPTER V
DISCUSSION

5.1 Tissue equivalent phantom

Tissue equivalent mixture consists of 180 g. gelatin, 350 g. coffee mate, 15 g floral cell, 50 g Sodium benzoate, and 1 liter water. The attenuation coefficient is 0.434 dB/cm/MHz. It is found that the quantity of gelatin and coffee-mate is direct proportional to the attenuation coefficient. However, when the ratio of gelatin to water is high, it is very hard to dissolve. Therefore, if one would like to make a tissue phantom with higher attenuation coefficient, some other contents would be considered.

Two methods are used to calculate the attenuation coefficient of the tissue material. One is calculated from the amplitude ratio pf echo signals along a line of echo profile in tissue phantom. The other is calculated from the amplitude of echo from strong echo material (aluminum plate). Since the ultrasound intensity and amplitude is attenuated exponentially in a medium, the relationship between amplitude and depth in medium can be plot graphically in logarithmic scale. It is found that the attenuation curve in tissue phantom is $y = 2.8004e^{-0.1877x}$, with R square = 0.9912. Therefore, the attenuation coefficient is 0.1877 Np/cm/3.75 MHz or 0.434 dB/cm/MHz. It would be said that both methods can be to obtain the attenuation coefficient in a tissue phantom.

Tissue equivalent phantom is normally in the solid state at room temperature. However, when the therapeutic ultrasound is applied, the state of phantom is changed gradually to liquid state. A study of comparison the trend of temperature in liquid and solid state phantom indicates that temperature raised at the first 3 cm depth of each phantom is rapidly increase. After that the temperature is slowly increased with depth until it tends to reach the plateau at about 9-12 cm depth. It seems that there are 2 states; an initial state at about 2-3 minutes after therapeutic application, in which the rate of temperature conduction in tissue phantom has no effect. Then after, when the conduction effect is more pronounced, the temperature tends to reach the resting state.

5.2 Calculation of therapeutic ultrasound intensity at depth of interest

From the attenuation coefficient (α), the intensity of therapeutic ultrasound can be calculated by the equation $\frac{I}{I_0} = e^{-2\alpha x}$ where I is the intensity at a depth of interest (W/cm^2) in phantom, I_0 is the reference intensity (1.5 W/cm^2), and x is the distance at the depth of reference and the depth of interest (cm). It is found that the therapeutic intensity at 2, 3, 5, 6, 7, and 9 cm are 1.23, 1.11, 0.91, 0.82, 0.74, and 0.61 W/cm^2 respectively.

It is found that the calculation intensities at the depth above 6 cm are chosen because there was the interference by diagnostic ultrasound at 1 to 5 cm, the intensity from calculation and the experiment may not be the same, and 9 cm level had a little effect of ultrasound, then this study considered the temperature at only 6 cm to compare.

According to Figure 4.13 and 4.14, we assumed that the conduction is only at 2 minutes and 6 minutes after stopping the sonication, but the temperature did not decrease to ambient temperature. The temperature distribution in liquid phantom for measurement did not follow the calculation, and because of the scattering property of the component the temperature rise may be caused by the impact of molecules inside the phantom.

For two layers phantom which consisted of tissue layer and bone layer (at 7 cm), it was found that there is temperature raised above the bone layer because of the accumulation of heat at bone surface and transfer to surrounding tissue.

CHAPTER VI
CONCLUSION

The tissue equivalent material consists of 180 g. gelatin, 350 g. coffee mate, 15 g floral cell, 50 g Sodium benzoate, and 1 liter of water. The measure attenuation coefficient is 0.434 dB/cm/MHz at density of 1.11 g/cm^3. It can be said that the attenuation coefficient is not changed with the liquid or solid state of tissue phantom.

From the echo profile, it can be seen that the amplitude of echo is increased as increasing the depth until reached the plateau at 6 cm and then gradually decreased with increasing depth. This due to the interference occurring at the depths closed to the surface of tissue phantom. Therefore, the amplitude ratio at the depths beyond 6 cm should be used for calculation of attenuation coefficient.

The calculated therapeutic intensities at 2, 3, 5, 6, 7, and 9 cm are 1.23 W/cm^2, 1.11 W/cm^2, 0.91 W/cm^2, 0.82 W/cm^2, 0.74 W/cm^2, 0.74 W/cm^2, and 0.61 W/cm^2 respectively at 1.5 W/cm^2 applications.

Because of the range and sensitivity of Telethermometer is too low, we cannot measure the fine values and it takes time to let it stable before measure. The temperature measure should have higher sensitivity or it should have the automatic record thermometer for more reliable result.

The study of temperature distribution shows that the temperature raised in phantom has 2 states. The initial state is in about the first 2-3 minutes where the temperature raised rapidly without the effect of conduction in tissue phantom. After that the rate of temperature raised decreases until it reaches a plateau at resting state.

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APPENDIX

APPENDIX

Appendix 1

The methodology to set up Enraf Nonius Sonopuls 190 therapeutic ultrasound.

1) Switch on the unit using On/Off button.

2) Select the desired duty cycle (continuous ultrasound is 100%, pulsed ultrasound is 5, 10, 20, 50, and 80%). The selection is indicated in the display

3) Select the treatment time. The set time is indicated on the display

4) Select display in watt or W/cm^2. Set the intensity/power.

5) Position the treatment head. Apply the aquasonic gel.

6) Starting the treatment when adequate contact is made, the timer will start (flashing dot)

Appendix 2

The seven steps to make the human tissue phantom are:

1) Measure of each mixture as shown in the table.

2) Boiling water and dividing into 4 portions as shown in the table.

3) Put the gelatin into the hot water and stir until the gelatin dissolved and sweep the bobble out.

4) In a mixing bowl, dissolve the coffee mate (non-daily cream) in the hot water and done with the floral cell powder.

5) Dissolve the sodium benzoate in the normal water. (It cannot dissolve in the hot water)

6) Mix all together and stir until the mixture is well blended.

7) Pour the mixture into the prepared container. Leave the container in the refrigerator until the mixture set.

Appendix 3

The original image of echo signals in the same tissue phantom measured at every 5 minutes during heating started from solid state phantom

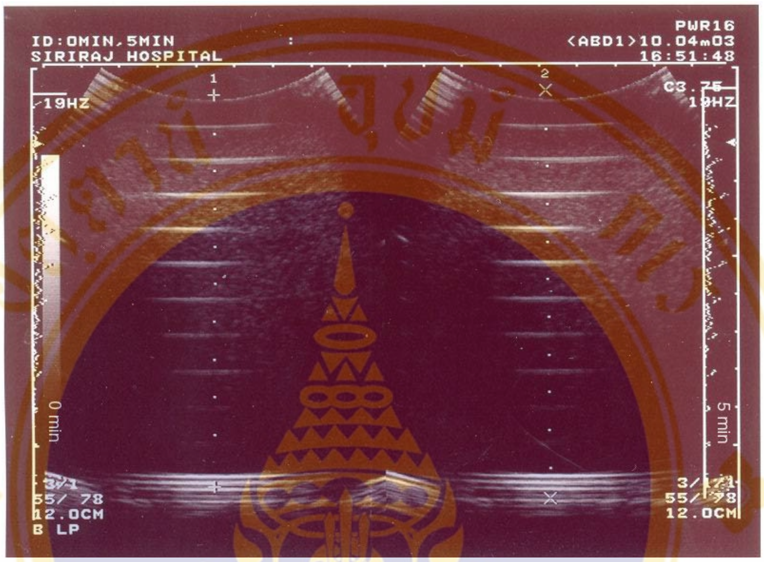

Figure 1 The original image of echo signal before heating and 5 minutes after heating.

Figure 2 The original image of echo signal after 10 and 15 minutes heating.

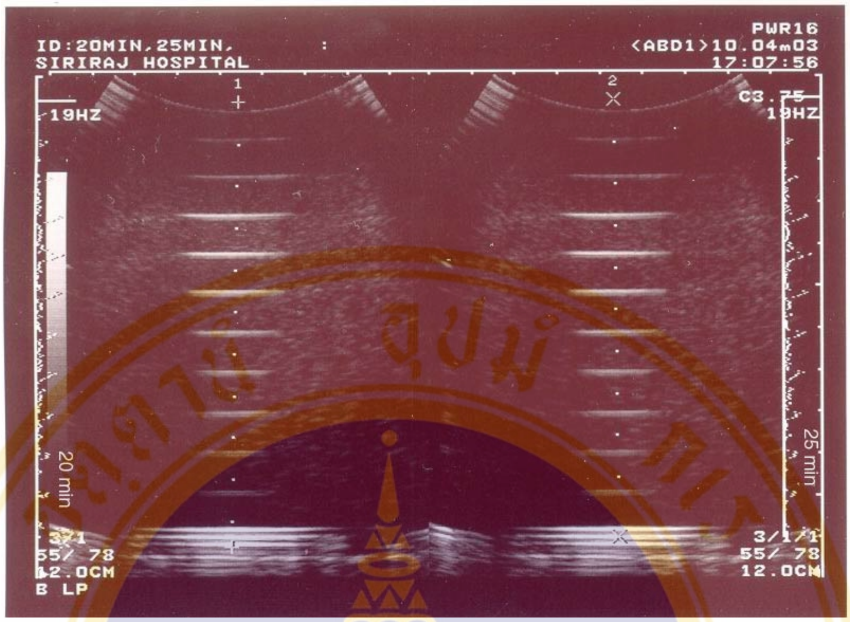

Figure 3 The original image of echo signal after 20 and 25 minutes heating.

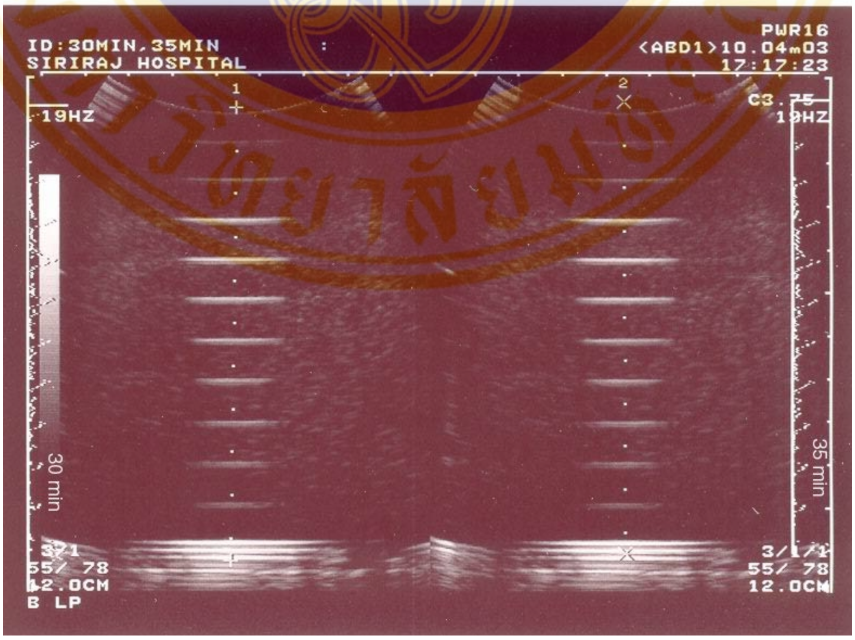

Figure 4 The original image of echo signal after 30 and 35 minutes heating.

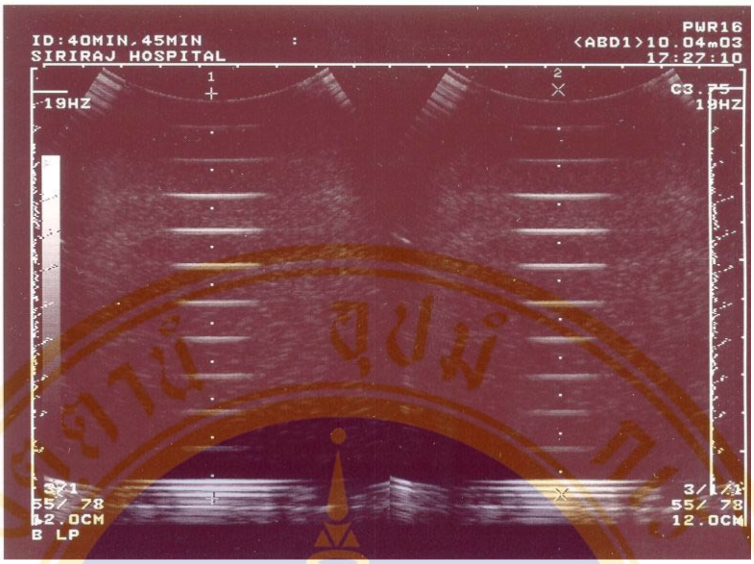

Figure 5 The original image of echo signal after 40 and 45 minutes heating.

Appendix 4

The original image of the amplitude of echo signal at different depths measure from 10 separated phantom of the same tissue material.

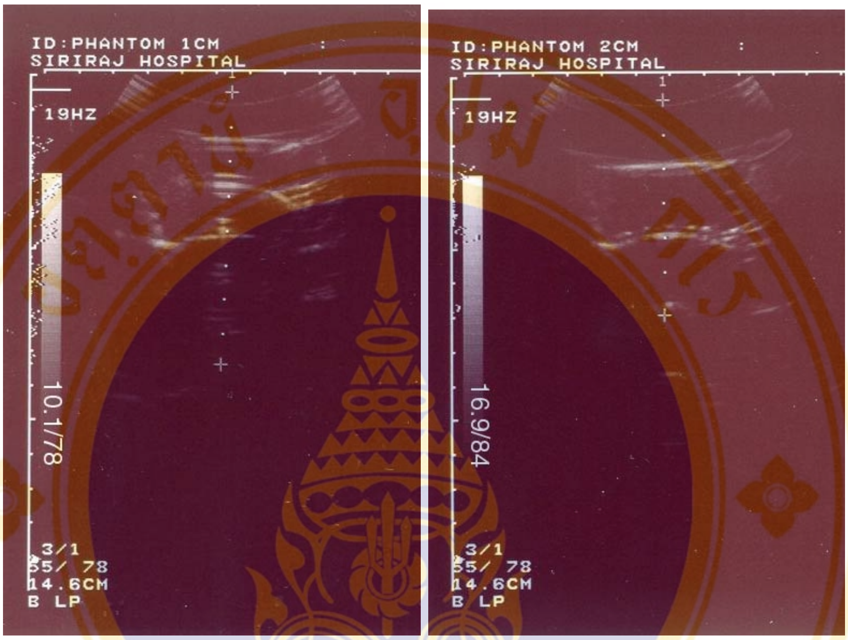

Figure 1 The original image of phantom at 1 and 2 cm depth

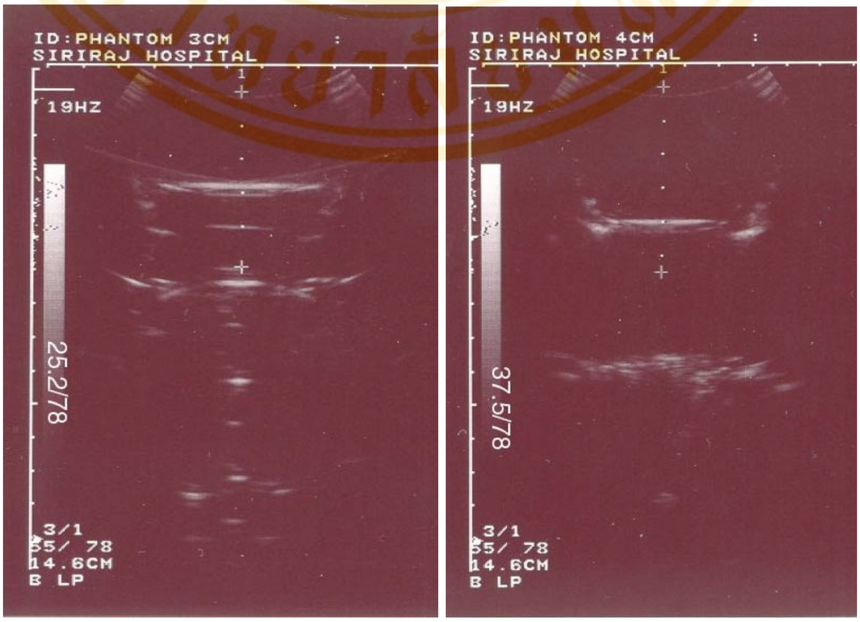

Figure 2 The original image of phantom at 3 and 4 cm depth

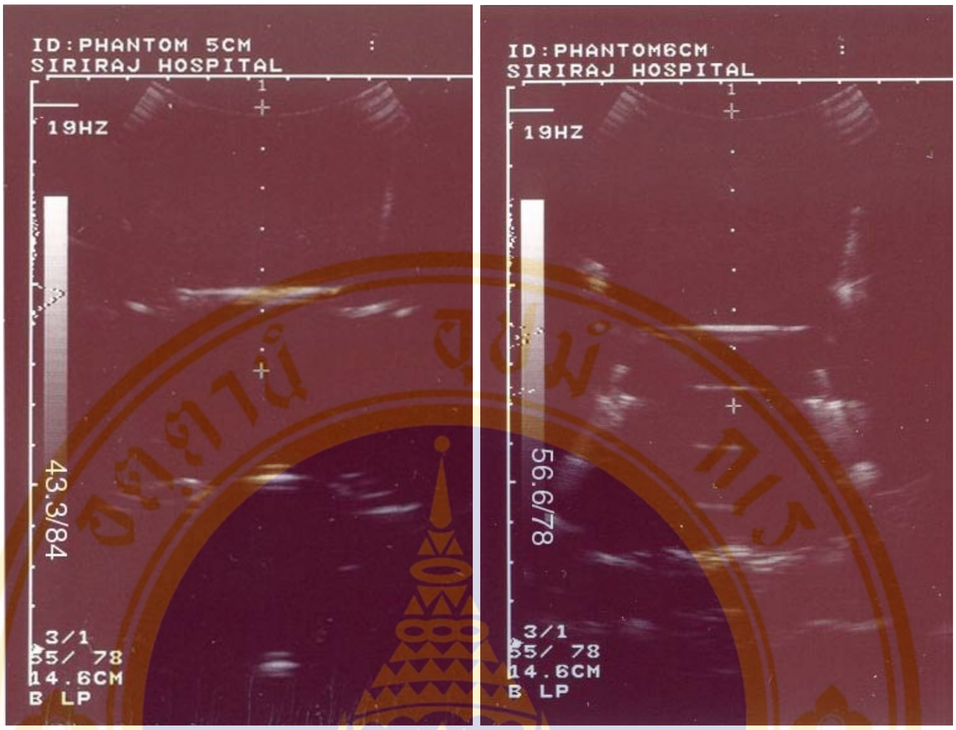

Figure 3 The original image of phantom at 5 and 6 cm depth

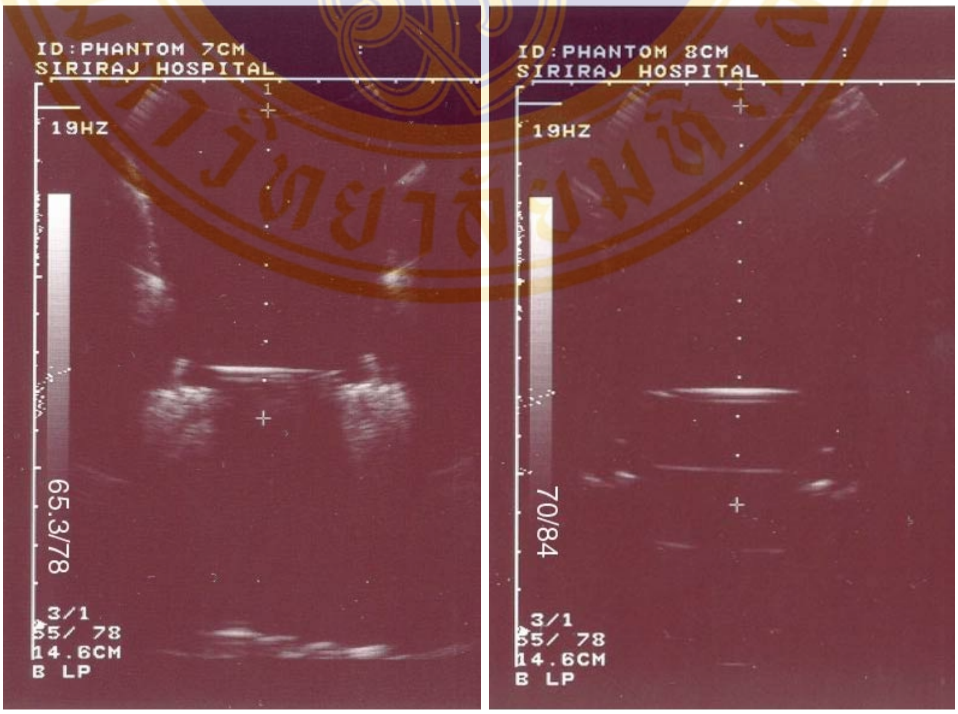

Figure 4 The original image of phantom at 7 and 8 cm depth

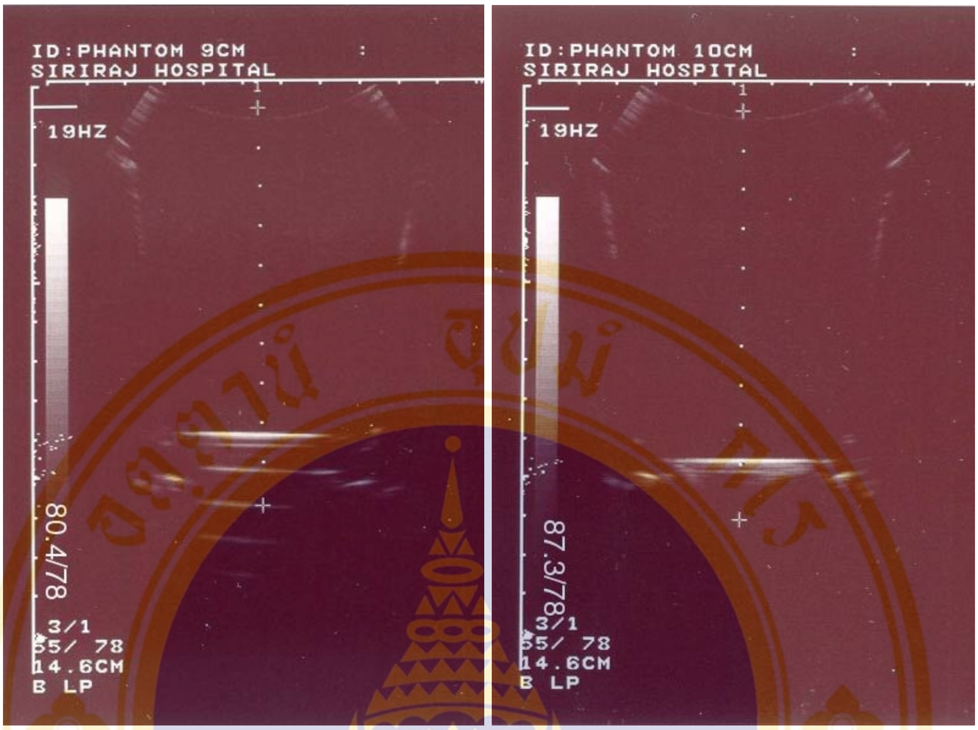

Figure 5 The original image of phantom at 9 and 10 cm depth

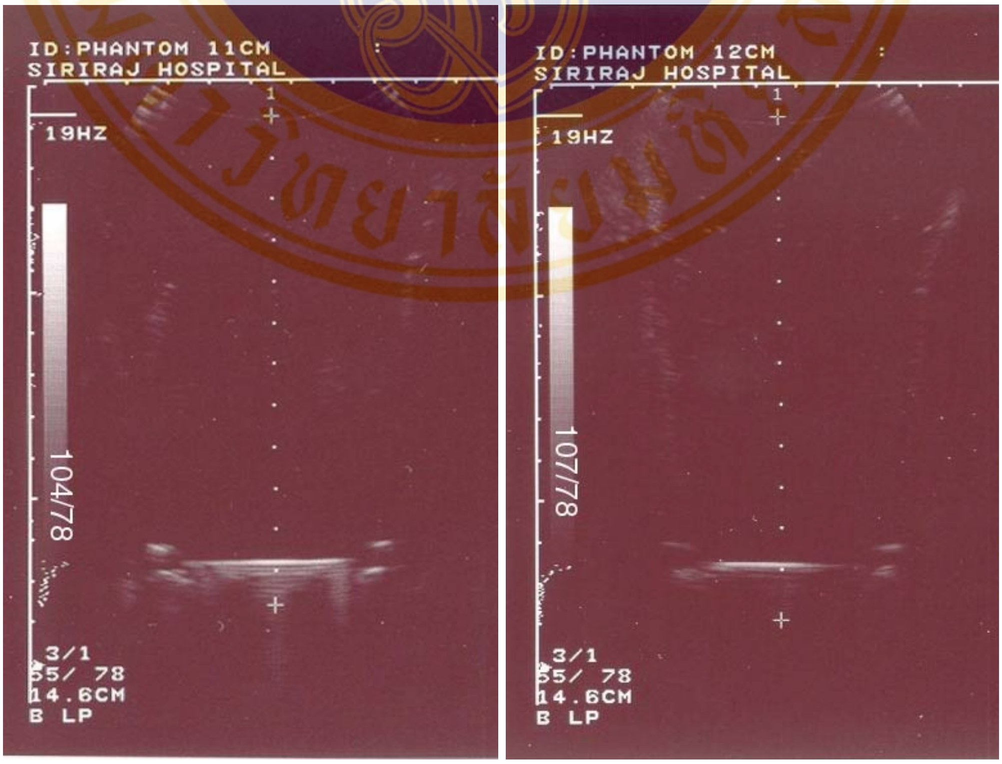

Figure 6 The original image of phantom at 11 and 12 cm depth

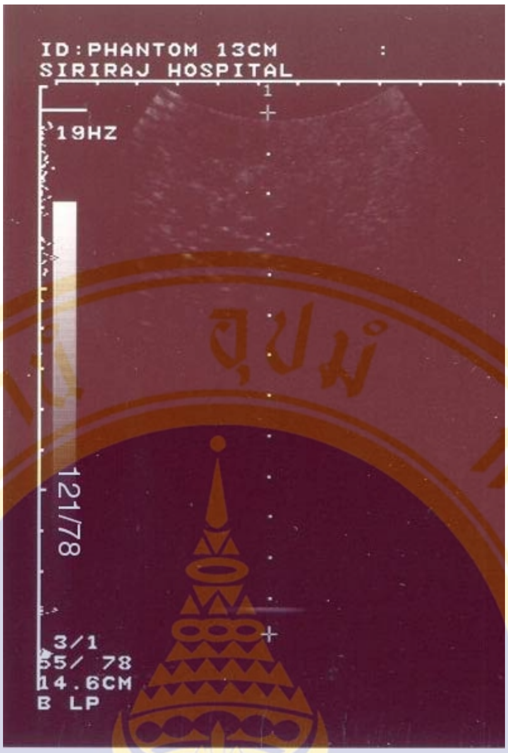

Figure 7 The original image of phantom at 13

BIOGRAPHY

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