Fiber Optic Probe Hydrophone for HIFU field Measurements

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I am grateful to the Almighty God for showering His countless blessings upon me. It is indeed a moment of great pleasure & honor to be graduating from Yonsei University, one of the most prestigious universities in South Korea. I am so deeply thankful to Prof Seo Jongbum for giving me the opportunity to travel thousands of miles to pursue a Masters course. Prof Seo Jongbum has indeed proven to be a remarkable mentor who continuously provided superb guidance & supervision throughout my studies. He continuously motivated & inspired me to reach my maximum potential to complete my Master degree. I am also thankful to my thesis supervision committee members Prof Yoon Daesung and Prof Lee Sangwoo for their contributions and guidance in my thesis.

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Abstract

Fiber optic probe hydrophone for HIFU field measurements

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Fiber Optic Probe Hydrophone (FOPH) has been developed in this research for measurements of HIFU fields and shockwaves. The FOPH offers a unique advantage over other PVDF Hydrophones for measuring higher pressure levels and shockwaves because of its frequency independence and the fact that the sensor tip is repairable if damaged by cavitation. The FOPH also offers small detector size which eliminates the need of spatial averaging. It also provides greater immunity to electromagnetic interferences. The proposed system contains a high power pigtailed laser diode module (1.5W@850nm) connected to a 2x2 arm 3dB optical coupler via FC-FC Adaptor. The optical coupler used in the system employs a 100/140μm multimode

optical fiber as the sensing element. A silicon photodetector (0.55A/W@850nm)receives the modulated light signal and converts it into voltage output proportional to the applied acoustic pressure, which is further amplified by using a 40 dB amplifier. Three transducers having resonance frequencies of 1.1 MHz, 2.2 MHz and 5.6MHz have been used for calibration of the FOPH by comparing its output with the output of a calibrated 0.2 mm PVDF membrane hydrophone. The sensitivity was determined as 3.6±0.1mV/MPa. Sensitivity was found to be linear and frequency independent. Pressure waveforms measured by FOPH showed agreement with those measured by membrane hydrophone for all transducers. The FOPH system was then tested with two ballistic shockwave devices to measure the pressure of shockwaves of their transducer probes. Shockwaves generated by the ballistic sources were recorded. Positive acoustic pressures in the range of 4 MPa to 20 MPa and negative pressures in the range of -3MPa to -15 MPa were measured.

Chapter 1

Introduction

High intensity focused ultrasound (HIFU) is a rapidly growing field. HIFU devices are used for non-invasive surgery and cancer therapy. HIFU is an entirely non-invasive process which uses external transducer to focus ultrasound beam at a target inside the body. The aim of HIFU is to deliver extracorporeal focused ultrasound energy to a well-defined target volume through intact skin and there by induce coagulation of the tumor without causing damage to the surrounding areas [6]. By focusing acoustic energy into a small volume, HIFU can produce thermal ablation and tissue necrosis [25]. Modern HIFU devices operate at high focal intensity levels from 1000W/cm^2 to 25000 W/cm^2 and are highly focused to a size of millimeter range [6]. HIFU measurements are different from the diagnostic medical ultrasound because the output pressure levels are very high and closer to the damage threshold of healthy tissues [25]. Pressure and intensity calculations of HIFU depend on the temporal waveform measured by the hydrophones [18]. The over-treatment or under-treatment of the affected area is clearly undesirable; therefore, accurate characterization of HIFU fields is crucial for planning and optimization of HIFU treatments [16].

Shockwaves have long been used for the therapeutic purposes in the medical field [5]. The shockwaves are high energy acoustic pulses, which are generated by electromagnetic, electrohydraulic or pneumatic mechanisms [3, 35]. The extracorporeal energy generated by the shockwave devices is delivered to the tissues inside the body for the treatment purposes [7]. The energy of the shockwaves needs to be properly determined for the treatment planning and optimization.

The hydrophones are most commonly used for measuring the output pressure levels produced by HIFU and acoustic fields [3]. The hydrophones are usually specified in terms of sensitivity, frequency response, effective detector size and level of robustness [18]. The measurement of the HIFU fields needs the hydrophones with adequate sensitivity, broad bandwidth, small detector size and higher robustness to withstand the high pressures [1-3]. The bandwidth and size limitations may lead to errors in the field measurements; therefore, in order to minimize the spatial averaging, the element size should be smaller than the acoustic wavelength [16, 18]. Robustness is a very fundamental characteristic of the hydrophone in the high power measurements [8] as cavitation is likely to occur during the HIFU measurements. The phenomenon of cavitation relates to the formation and collapse of microbubbles. Microbubbles are generated due to the rarefactional pressure of the acoustic waves. Cavitation is of two types, stable and inertial. Stable cavitation means the periodic growth and oscillations of the bubbles, while inertial cavitation means the violent growth and collapse of the bubbles. Inertial cavitation usually happens at higher pressures, and it can cause irreparable damages to the surface of the hydrophones.

Polyvinylidene fluoride (PVDF) membrane or needle hydrophones fulfill most of these requirements and are often used for measuring the HIFU fields at low intensity levels. The needle hydrophones consist of a disc-shaped active element at the tip of a needle like structure [8]. The needle hydrophones have limited bandwidth and irregular frequency response. Their element size varies from 0.2mm to 1mm and their robustness and sensitivity decrease with a decrease in the element size; which makes them more delicate and their susceptibility to cavitation damage increase due to the reduction in their active element size [8]. Also, the needle hydrophones have to be replaced in case of damages caused by cavitation. The membrane hydrophones are generally accepted as a standard for the acoustic pressure measurements in the medical ultrasound fields below the threshold of cavitation [8]. They have a broad bandwidth and a higher sensitivity but the high pressures can damage them irreparably. Therefore, to avoid the cavitation, the measurements are usually carried out at low pressure amplitudes and then the results are extrapolated linearly [8, 25], which may not provide accurate information and satisfactory results. Also, considerable difficulties are associated with manufacturing of the PVDF membrane hydrophones having a small sensor area [3]. The sensitivity of the membrane hydrophones decreases with a decrease in the detector size and they become more delicate and fragile [8] which make them more vulnerable in high pressure fields. The ability of PVDF hydrophones (membrane and needle) to measure negative pressure is limited, and their sensitivity is frequency dependent [8, 23].

Fiber optic techniques have been investigated by the researchers to develop a small size robust hydrophone which can withstand high pressures in the acoustic field measurements [1, 8]. Based on the working mechanism, they can be classified into phase modulated, wavelength modulated and intensity modulated [11]. Phase modulation methods involving interferometry have higher sensitivity; but they are more complex in designs [8]. Fabry-Perot interferometer uses phase modulation mechanism; which involves the deposition of a thin polymer film at the fiber tip. Ultrasonic field induces a change in the thickness of the tip and an interferometer is used to detect these acoustically induced displacements [15]. The problem arises if the fiber tip is damaged, because it has to be recoated with the dielectric and the calibration has to be performed again [1, 15]. Wavelength modulated optical fiber sensors use the mechanism of acoustically induced shift in the wavelength of optical signal; which results in frequency modulation of the signal which is then detected by an FM detector; however, these sensors have bandwidth in the range of few kHz which is a serious limitation [11].

The simplest fiber optic sensor mechanism is the intensity modulated fiber optic probe hydrophone (FOPH) system [1]. The FOPH system provides an optimal solution to meet the above mentioned requirements of HIFU field measurements. The FOPH system was first used by Staudenraus and Eisenmenger [3] for acoustic field measurements. The FOPH system detects pressure induced changes using Fresnel reflection phenomenon, and it is more simplistic in construction. The FOPH offers several advantages over the PVDF hydrophones, which include the ability of the FOPH to register shockwaves and high pressure HIFU fields, robustness, small detector size, broader directionality, enhanced spatial resolutions and greater immunity to electromagnetic interference [1-6]. The FOPH system theoretically provides frequency independent sensitivity which is limited only due to the associated electronics of the system [3]. The uniform response of the FOPH system with respect to the frequency of acoustic source has been verified up till 40MHz [33].

The core area of optical fiber forms the detector area, which can be in the range of 50-200 μm for multimode fiber, thus practically eliminating the need for spatial averaging corrections in HIFU fields [3]. Additionally, strong adhesion of water on the glass keeps the fiber wet and reduces the event of cavitation at fiber tip [24]. Furthermore, the FOPH sensor tip can be easily repaired by cleaving, if damaged by cavitation, which can cause irreparable damage to the other PVDF hydrophones [20-24]. Therefore, FOPH is more desirable for measurements in HIFU fields.

This research focuses on the simplest method of the FOPH system using Fresnel reflection phenomenon, which is straightforward in construction but has a fundamental disadvantage of lower sensitivity [18]. Our purpose is the development of the FOPH system having a higher sensitivity by using the combination of a high power laser diode and a silicon photodetector along with a 40 dB pre-amplifier for HIFU measurements. The high power laser diode used in this system will ensure that adequate intensity of the laser light is transmitted to the fiber tip, which will increase the reflected intensity and hence will improve the sensitivity of the system.

The system was calibrated by comparing the signals of the FOPH with a calibrated membrane hydrophone to determine the sensitivity. Three ultrasonic transducers having different center frequencies were used for calibration. Once the sensitivity and the frequency response were established, the FOPH was then used to measure the acoustic shockwaves from two ballistic shockwave therapy devices.

1.1 Theoretical Background

The FOPH device measures pressures of an acoustic field by using piezo-optic effect [1] utilizing Fresnel reflection formula. Piezo-optic effect relates to the change in the refractive index of the medium due to the presence of ultrasound waves in that medium. The optical reflectance at fiber endface is linked with the pressure amplitude via an index of refraction and density relationship. The refractive index of the water changes due to a change in density with the dynamic effect of the pressure field [1-6]. Light from a laser module is launched into the optical fiber which is positioned in the acoustic field inside water. The applied acoustic field changes the density of water, which in turn changes the reflectivity and hence the light that is reflected back at the fiber endface is proportional to the applied acoustic pressure [1-6]. A photodetector can be used to monitor the modulated light intensity to obtain a time varying voltage output. Due to the low compressibility of the optical fiber, the change in its density is neglected. The reflectivity 'R' for normal incidence of laser light at the fiber endface and host water is given by the Fresnel Formula given in equation (1.1).

$$R = [\frac{n_c - n_w}{n_c + n_w}]^2 \qquad (1.1)$$

where n_c is the refractive index of fiber optic and n_w is the refractive index of water. In case of transmission from fiber (n_c=1.485) to water (n_w=1.329), the value of R is 0.3%. The change in the refractive index of water varies linearly with the applied pressure in the range of 0.01 to 100 MPa. The relationship of the refractive index n with density ρ can be obtained by using Gladstone-Dale model in the equation (1.2) and the values are valid up to a pressure level of 500 MPa [3].

$$\frac{n(\rho)-1}{\rho} = const = \frac{n_{0-1}}{\rho_0} \qquad (1.2)$$

A change in the acoustic pressure changes the density, which in-turn changes the refractive index and reflectivity; hence the change in reflectivity is proportional to the change in applied acoustic pressure. The change in reflectivity modulates the reflected light according to the applied acoustic pressure. Figure 1.1 shows that the linear relationship of acoustic pressure and change in reflectivity (ΔR) calculated by Lewin et al[4].

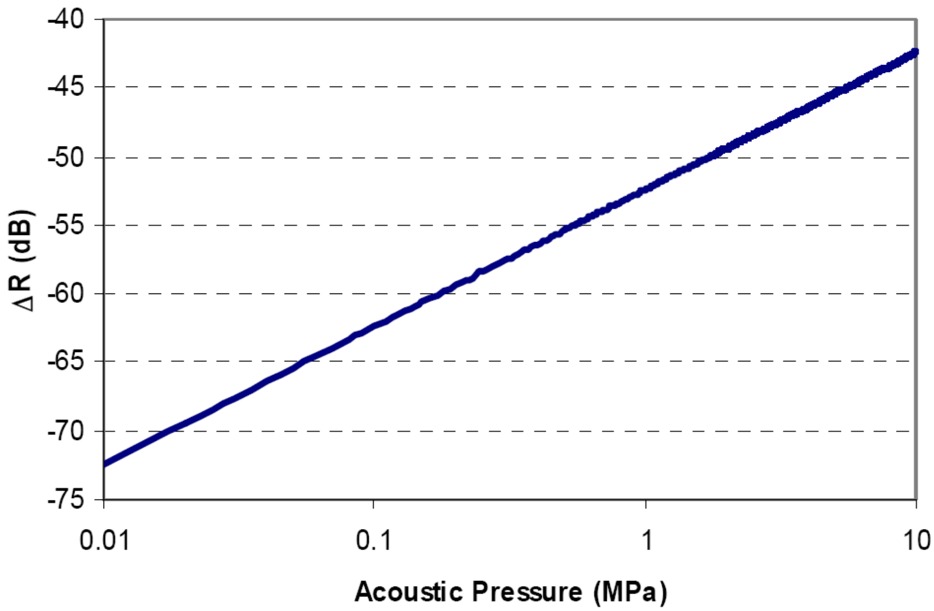

Figure 1.1: Variation in reflectivity according to applied acoustic pressure, adapted from Lewin et al [4].

This linear change in the reflectivity allows defining the piezo-optic constants for water and fiber optic as shown in following equations 1.3 &1.4 [1, 4].

$$\frac{\Delta n_w}{\Delta p} \approx 1.4 \times 10^{-4}\ \mathrm{MPa}^{-1} \qquad (1.3)$$

$$\frac{\Delta n_c}{\Delta p} \approx 5 \times 10^{-6}\ \mathrm{MPa}^{-1} \qquad (1.4)$$

where p is acoustic pressure.

Since $\frac{\Delta n_c}{\Delta p} \ll \frac{\Delta n_w}{\Delta p}$, it justifies the common practice of neglecting the change in refractive index of fiber optic during acoustic measurements.

Silicon photodetector receives the modulated laser light and gives temporally varying voltage output. The silicon photodetector detects the light in the range of 300nm to 1100nm and its output is dependent on its responsivity given in A/W. The transfer factor 'S' for the FOPH system can be defined as a ratio of voltage change ΔV to the pressure change Δp shown in equation 1.5 [1, 5].

$$S = \frac{\Delta V}{\Delta p} \qquad (1.5)$$

Thus, the temporally varying voltage output of FOPH can be converted to pressure values once the sensitivity of the system is determined by using this transfer factor. The reflectivity varies linearly with the applied pressure; hence the increase in applied pressure will result in increased voltage output of the photodetector. Since the speed of light is much higher than the speed of sound, these pressure variations due to compression and rarefactions appear stationary to the incident laser beam, therefore, the output signal of the photodetector represents the behavior of the acoustic source [8]. If the acoustic wave is perpendicular to the plane of the fiber tip, the theoretical bandwidth corresponds in the range of 3 GHz [3] and eliminates the problems associated with the bandwidth limitation. The bandwidth limitation comes with the associated electronics in the FOPH system. The calibration of the FOPH system only needs the determination of voltage output of photodetector in relation to the applied acoustic pressure amplitude [1]. Once the sensitivity is determined, the linearity of the FOPH system allows measuring the high acoustic pressures. Output of the FOPH system has been reported to be linear in the range of high acoustic pressures [3].

Water is usually recommended for acoustic measurements because its impedance properties are nearly identical to that of soft tissue; also, water is convenient to use and abundantly available [6, 18]. However, there are certain problems associated with using water instead of soft tissues as water offers negligible attenuation as compared to the soft tissues and it becomes difficult to extrapolate the water measurements to *in vivo* measurements [8]. Therefore, researchers have demonstrated the use the FOPH to perform *in vivo* measurements [5]. Gas content of water should be kept as low as possible to avoid formation of the microbubbles on the hydrophone and acoustic source surfaces, as it can lead to inaccuracy of the results [1]. Presence of the microbubbles at the fiber-tip can result in unstable waveform [18] or unusually high voltage signals assumed to be resulted due to cavitation [5, 7]. Also, the cavitation can damage the fiber tip which has to be recleaved and repositioned in that case [1].

Chapter 2

Methods

2.1 FOPH System

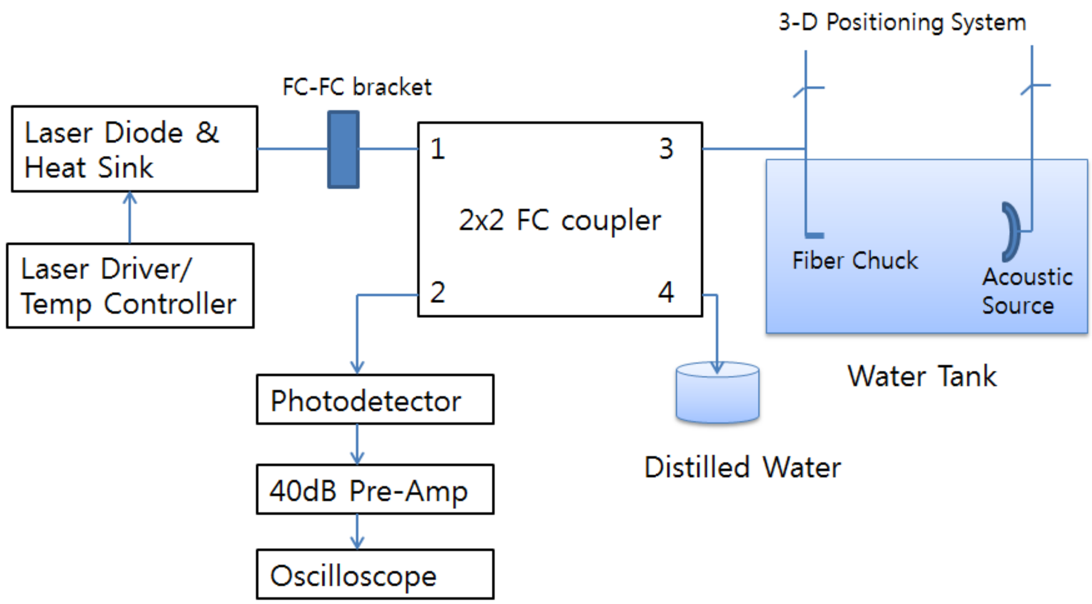

Figure 2.1: Schematic diagram of the FOPH system

Figure 2.1 shows the schematic diagram of the FOPH system used in this research.

The FOPH system contains a pig-tailed butterfly packaged laser diode (Axcel photonics, USA) having a typical power of 1500mW. The pig-tailed butterfly package allows minimizing the coupling losses as the optical fiber is directly coupled to the laser diode and the fiber is then terminated with a standard FC connector. The wavelength of laser diode is 850nm. The wavelength falls under IR range which gives an opportunity to readily check the functionality of the laser system and helps in the alignment procedures as well. The laser diode in mounted on a butterfly mount (model 744, Newport, USA) which allows connection to the laser diode driver and temperature controller. The FOPH system employs a laser diode driver (model 560-B, Newport, USA) which can be driven at a constant current. The temperature controller (model 350B, Newport, USA) is used to control and monitor the temperature of laser diode.

The system uses an optical coupler (Gould Fiber Optics, USA) which is a 2x2 arm 3-dB (50:50) bi-directional fused device consisting of two input leads and two output leads. 100/140μm multimode glass fiber is used in its manufacturing. Both input leads (port 1 and port 2) of the optical coupler are terminated with FC connectors, while one output lead of the optical coupler (port 3) is the bare fiber, used as the sensing arm in the measurements. The sensing fiber is 5m in length.

The output of the laser diode is connected to the port 1 of the optical coupler through an FC-FC bracket (model FCB1, Thorlabs, USA) which has specified insertion losses of about 0.2 dB. The sensing fiber (port 3) is cleaved by using a cleaver (model MAX CI-03, ILSINTECH, Korea). A fiber inspection scope (model CL-200, Thorlabs, USA) is used to observe the cleaved fiber endface. The fiber is recleaved whenever cracks are found on the fiber tip. The fiber is then inserted into the fiber chuck (model HFC-007, Thorlabs, USA) and then positioned in the water tank using a holder system connected to a 3D positioning system. One output lead (port 4), which is not used in the FOPH system, is terminated in distilled water. This termination ensures that the reflection from the unused output is of miniscule [1].

The second input lead of optical coupler (port 2) is connected to a silicon photodetector (model DET36A, Thorlabs, USA) which has peak responsivity of 0.65A/W at 970nm and about 0.55A/W at 850nm used in the current system. The photodetector has a rise time of 14nsec and damage threshold of 100mW/cm^2 which makes it a suitable photodetector in the current system because a high power laser diode is being used. A 50 Ω terminator resistor is connected at the output of the photodetector to convert the current signal into the voltage signal. The output of photodetector is amplified using a 40dB, 0.1-400MHz pre-amplifier (Model 8447A OPT 001 Dual Amplifier, HP, USA). The behavior of the pre-amplifier is crucial in terms of its linearity, bandwidth, maximum input and output voltages, input impedance and voltage gain as it can impose limitations on the overall response of FOPH system [18].

The laser light is launched into the optical coupler from the laser diode, which splits the light into two output leads, attenuating it by 3dB. When the sensor lead is positioned in the acoustic field, the reflected light is again attenuated by 3dB to each of the two input leads; therefore, the reflected light is directed back to the photodetector and the laser diode. The photodetector gives a time varying voltage signal as a result, which is amplified by the 40dB preamplifier and displayed on the oscilloscope. The laser diode also receives the reflected light, but it does not pose a threat to the laser diode as the reflectivity is about 0.3% of the incident light under static conditions and further attenuated by 3dB [1].

Function Generator (model 33250A, Agilent, USA) and RF Amplifier (model 500A-100A, Amplifier Research Corp, USA) are used to drive ultrasonic transducers at their resonance frequencies. Digital Oscilloscope (Model DPO 4054, Tektronix, USA) is used to display and record the signals received from the FOPH system.

2.2 Positioning System

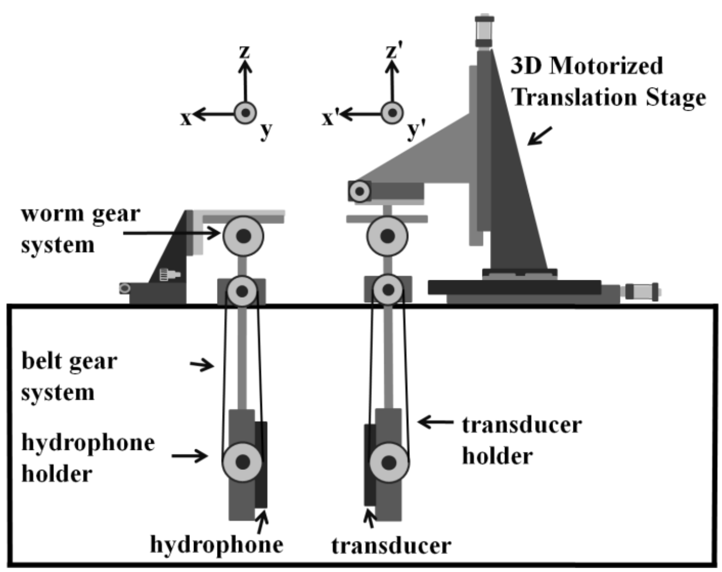

Figure 2.2: Schematic diagram of the positioning system

Figure 2.2 shows the schematic diagram of positioning system in detail. The positioning system contains a manual translation stage (S-120LRC, DPI, Korea) for the hydrophone holder and a 3D motorized stage (Parker, USA) for the acoustic transducer holder which is controlled by using customized Matlab programs. The 3D motorized stage is used to move the transducer in x'y'z' directions and the manual stage is used to move the hydrophone holder in xyz directions as shown in figure 2.2. The transducer and the hydrophone are mounted on the sub-holders, which are then attached to the holder systems as shown in figure 2.3. Holder systems contain angular positioning controls. Angular motion along vertical axis (z axis) is controlled using worm gear, while angular motion along horizontal axis (y axis) is controlled using belt gear in the holder systems. Belt gear system gives the option of performing angular adjustments outside water. Therefore, the positioning system provides the option of two independent angular alignments. These angular adjustments allow the gimbaling procedure to be performed in order to align the transducers according to respective planes.

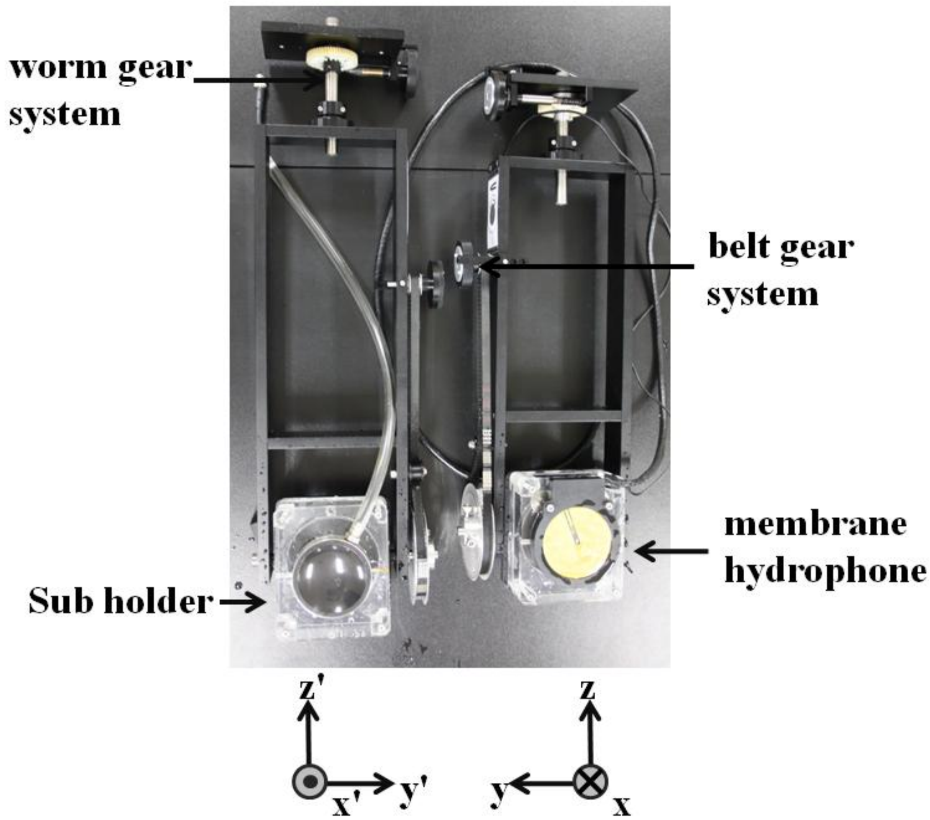

Figure 2.3 Holder systems for the transducer and membrane hydrophone

A photograph of the 3D positioning system is shown in figure 2.4, indicating the 3D motorized stage and the manual stage used for the FOPH system. The hydrophone is connected to the manual stage while the transducer is connected to the 3D motorized stage shown in the figure.

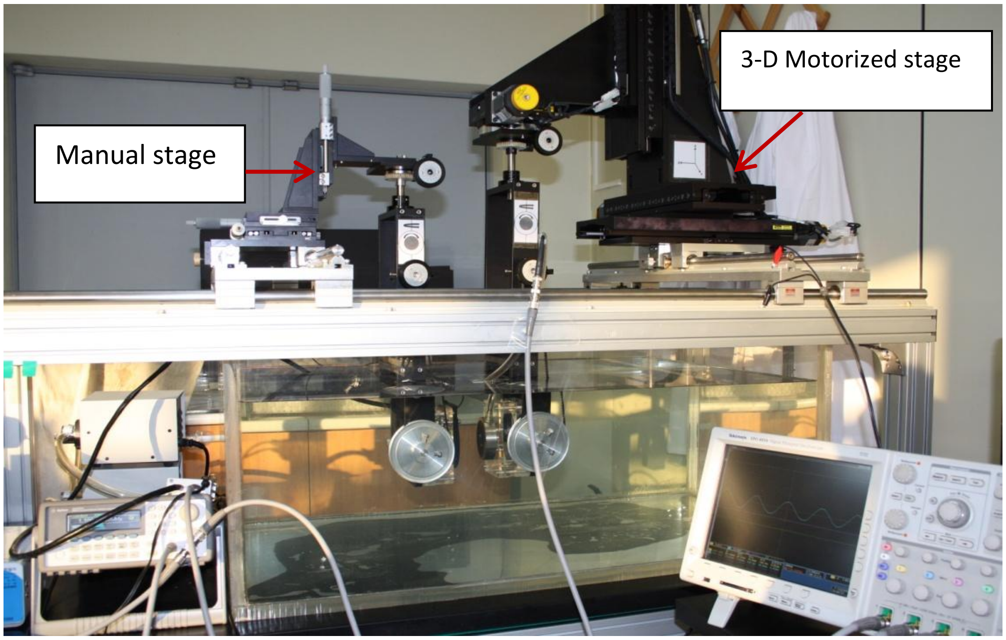

Figure 2.4: A photograph of the 3D Positioning system.

Alignment was carried out by connecting the transducer to the pulser/receiver (model 5073PR, Panameteric, USA). The reflected signal was then monitored to obtain the travelling time of reflected ultrasound when the transducer was moved in a vertical or horizontal direction using the 3D motorized stage. Firstly, the direction of hydrophone was aligned with respect to z' axis of the positioning system. For this purpose, angular positions were changed at the hydrophone holder system using the belt gear, and the transducer was moved 1 cm up/down in the z' direction to observe a change in the reflection time. The angular positions were changed until the difference of reflection time between two points (1 cm apart) became 5nsec (approximately), which indicated the angular error of 0.0004 radians. Secondly, the direction of the hydrophone was aligned with respect to the y' axis. The transducer was moved left/right in the y' direction. Angular positions at the hydrophone side were changed using worm gear until the reflection time between two points approximately became 5nsec.

After that, the direction of the transducer was aligned with respect to the hydrophone (z and y axes) by applying the condition of peak amplitude of the reflected waveform. Angular positions at the transducer side were changed by using worm and belt gear systems to get peak amplitude of reflected echoes. The process of hydrophone alignment with respect to transducer was once again repeated to ensure proper alignment.

After that the transducer was driven at the fundamental frequency using RF amplifier and the focus of the transducer was then determined by moving the transducer in 3-D using the motorized stage and locating the maximum signal from the hydrophone. This alignment procedure was followed in all measurements to ensure similar conditions and reliability of the results.

2.3 Calibration Method

Calibration was performed by comparing the FOPH output with that of a calibrated 0.2mm membrane hydrophone (Precision Acoustics, UK). The membrane hydrophone was calibrated from 300 kHz to 40 MHz by National Physics Lab, UK. The laser diode was operated at constant current of 1600mA and temperature controller was set at 25°C throughout the calibration process. Three transducers, a HIFU transducer of 64 mm diameter having resonance frequency of 1.1 MHz and two 19.5mm focused transducers having resonance frequencies of 2.2 and 5.6 MHz respectively were used in this procedure. The calibration process explained below was repeated for all of the transducers to establish the sensitivity and frequency response of the FOPH system.

During the calibration process, firstly the membrane hydrophone was positioned in the water tank using positioning system. Output of the function generator was connected to the RF amplifier, whose output in turn was connected to the transducer using an appropriate matching box. The input voltages to the transducer and corresponding pressure outputs of the membrane hydrophone were recorded. Matlab (Mathworks, USA) programs were encoded to control the function generator and the digital oscilloscope. Stepwise increase in the output of the function generator, led to a linear increase in the input voltage to the transducer which resulted in a linear graph of voltage input to the transducer versus output pressure of the membrane hydrophone. Since the sensitivity of the membrane hydrophone was already known because of its calibration data, the output was directly converted to pressure values. Then the membrane hydrophone was replaced by the FOPH. The voltage output of the FOPH was compared with the pressure output of the membrane hydrophone to determine the sensitivity of the FOPH. After that, the output of FOPH was converted to pressure values according to the sensitivity and both pressure outputs were compared. In case of the FOPH, the waveforms had to be inverted prior to converting them into the pressure values, because compressional (positive) pressures temporally increased the fluid density at the fiber tip, thus increasing the refractive index of water and reducing the mismatch of refractive indices, which resulted in a negative signal[1,5]. This phenomenon was also observed when the FOPH waveforms were compared with those of the membrane hydrophone. Water was degassed below 35% of the saturation level to avoid cavitation. If the fiber tip was found damaged due to the cavitation, it was taken out and recleaved to produce a new endface and repositioned to continue the measurements.

2.4 Shockwaves Measurement Method

The FOPH was used to measure the acoustic pressure generated by two ballistic shock wave therapy (SWT) devices. SWT device refers to use of focused shockwaves for physical therapy purposes for treatment of musculoskeletal problems. The SWT device releases high pressure acoustic waves into the target tissue for the treatment of the chronic pains and inflamed tissues. There is a significant mechanistic difference between a ballistic source and other shockwave devices which use electrohydraulic, electromagnetic or piezoelectric sources to generate the shockwaves [7]. The ballistic source consists of a pneumatic system; a transducer probe (hand-piece) within which compressed air is used to fire a projectile that strikes the metallic tip to generate shockwaves. These pneumatically generated shockwaves spread through the tissue and increase blood circulation in the affected tissue. The shockwaves can induce tissue repair processes and analgesic effects in the affected area. This improves the cell proliferation and tissue regeneration to repair the muscles and tendons [7]. The pressure settings of the compressed air can be selected from 1.4 bar up to 5 bar which changes the output pressure of the transducer probe accordingly.

The FOPH was used to measure these shockwaves for two SWT devices. Firstly, measurements for Storz probe (D-Actor 200, Storz Medical, USA), connected to Storz Medical SWT, were performed and its pressure output levels were compared to the reference data provided by its manufacturer. Secondly, Daeyang probe (Daeyang

Medical Co., Korea) was connected to Daeyang SWT and measurements were performed under similar conditions.

Measurements were performed in degassed water. Since the transducer probes were not entirely water proof, the holder system was modified to hold the transducer and fiber optic in a top-bottom position, so that only the tip of the transducer probe was inside water. The FOPH was connected to the motorized stage of the 3-D positioning system, while transducer probe was connected to the manual stage. Figure 2.5 shows the modified holder system. The 3D motorized system allowed determining and fixing the location of the optical fiber as the movements were done in discrete steps.

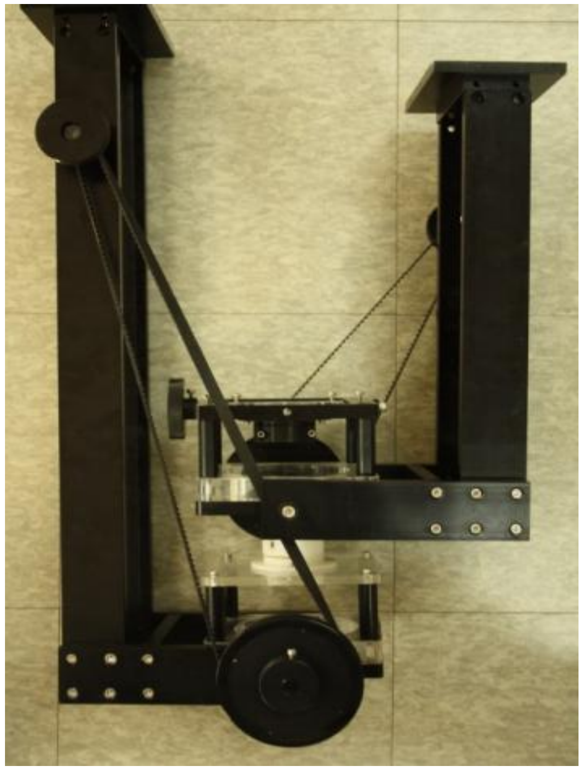

Figure 2.5: Modified holder system.

Before the measurement process, the location of the fiber tip was confirmed. The pulser/receiver system was connected to the 5.6 MHz transducer for this procedure, and the reflected echoes were used to determine the location of the optical fiber. Speed of sound was considered to be 1500m/sec to calculate the distance from travelling time of reflected echoes. After the location of the optical fiber was determined, the transducer probe was placed in the same position of

5.6 MHz transducer. Measurements were made by placing the optical fiber 1 cm away from the tip of the transducer probe. Alignment of the probe and fiber optic was done manually but carefully in order to achieve repeatable results.

Optical fiber was placed 1mm out of the fiber chuck to avoid the vibration of the fiber due to acoustic pressure. Water was degassed below 35% to avoid cavitation signals generated due to trapped microbubbles between the transducer probe and the fiber optic.

Chapter 3

Results

3.1 Calibration Results

Sensitivity for FOPH was obtained by comparing its output with a calibrated 0.2 mm membrane hydrophone used as the reference standard. Three transducers were used in the procedures and their results are shown in following figures. By comparison method, the sensitivity of FOPH was found to be 3.6±0.1mV/MPa. FOPH waveforms were inverted as mentioned earlier and averaged 512 times to reduce the signal to noise ratio as mentioned earlier. Figure 3.1 shows the output waveform signal of membrane hydrophone, while figure 3.2 shows the signal of FOPH. 1.1 MHz transducer was operated in burst mode of 20 cycles and burst period of 10msec. Both waveforms indicated close agreement. Output values were converted to pressure amplitudes according to their sensitivity. Voltage output of the function generator was controlled in order to control the input voltage to transducer by using customized Matlab program.

Incremental voltage input to the transducer resulted in incremental pressure outputs accordingly. The output of the FOPH could be considered linear and close in agreement with that of membrane hydrophone, as shown in figure 3.3.

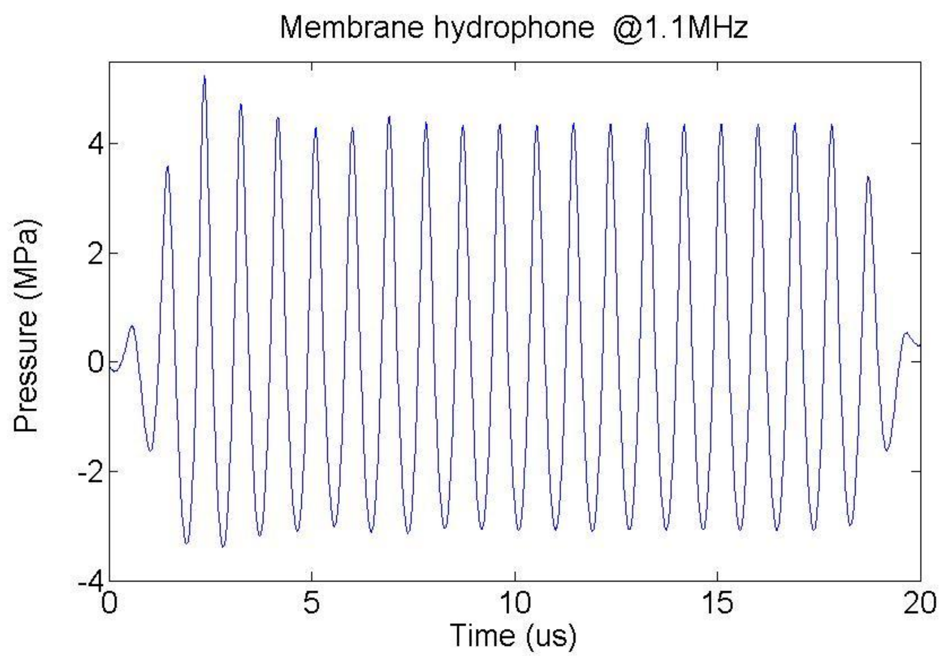

Figure 3.1 Membrane hydrophone signal for 1.1 MHz transducer

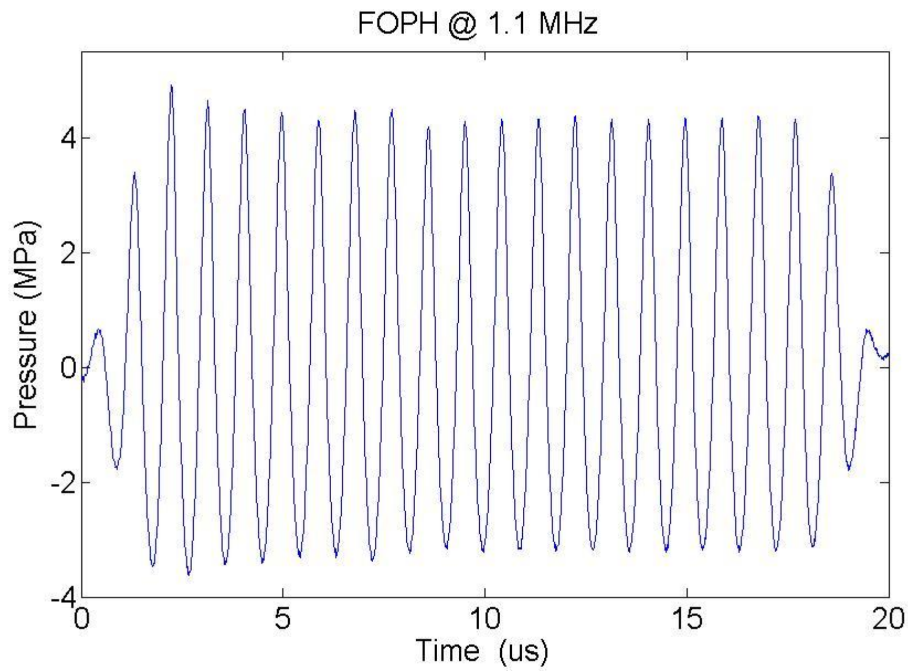

Figure 3.2 FOPH signal for 1.1 MHz transducer.

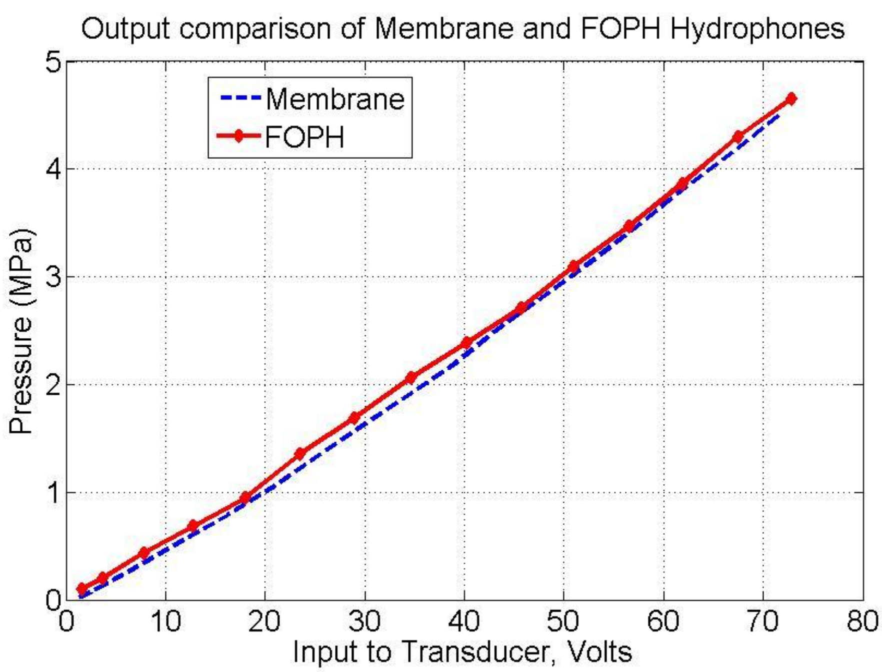

Figure 3.3: Comparison of the membrane and FOPH at 1.1 MHz

The sensitivity of FOPH was further confirmed by performing measurements for 2.2 MHz and 5.6 MHz transducers. Figure 3.4 shows the results membrane hydrophone for 2.2 MHz transducer while figure 3.5 shows output of FOPH for 2.2 MHz transducer. The transducer was operated in burst mode of 25 cycles for a burst period of 10 msec. The output of the FOPH tends to be noisier than that of membrane hydrophone, but both figures show a close resemblance. Figure 3.6 indicates that the pressure output is linear and comparable to that of membrane hydrophone.

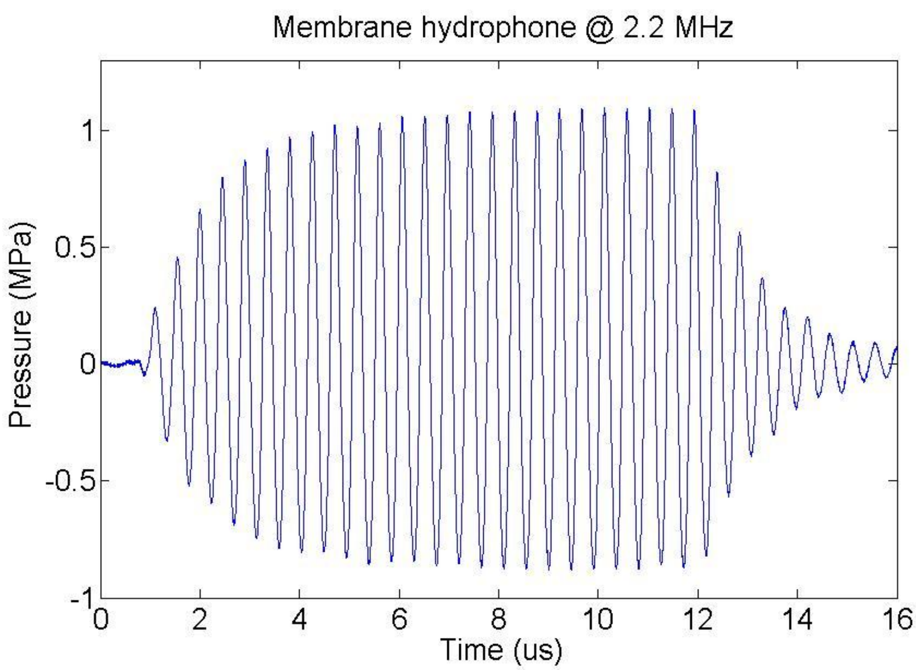

Figure 3.4: Membrane hydrophone signal for 2.2 MHz transducer

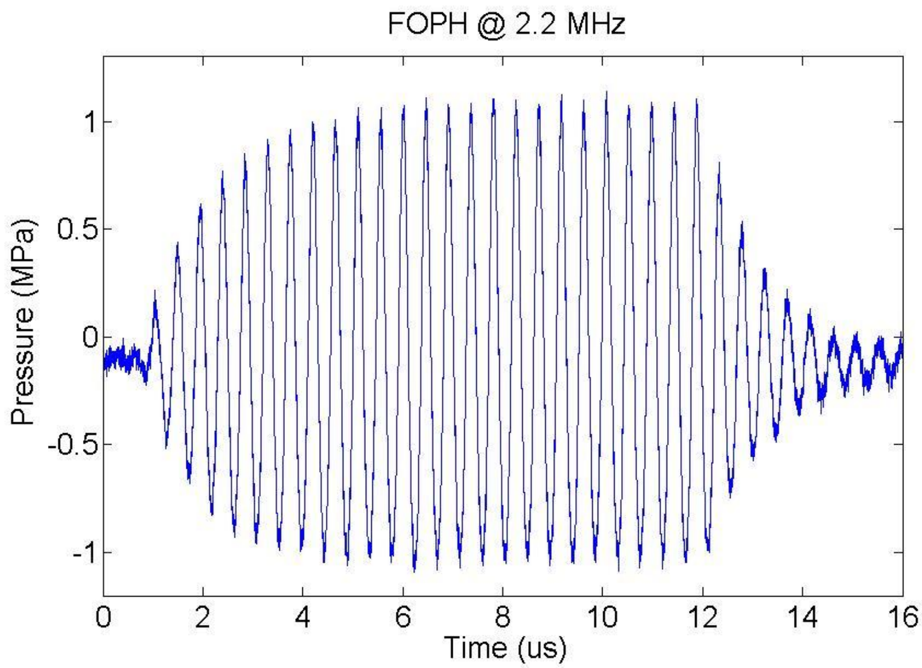

Figure 3.5: FOPH signal for 2.2 MHz transducer.

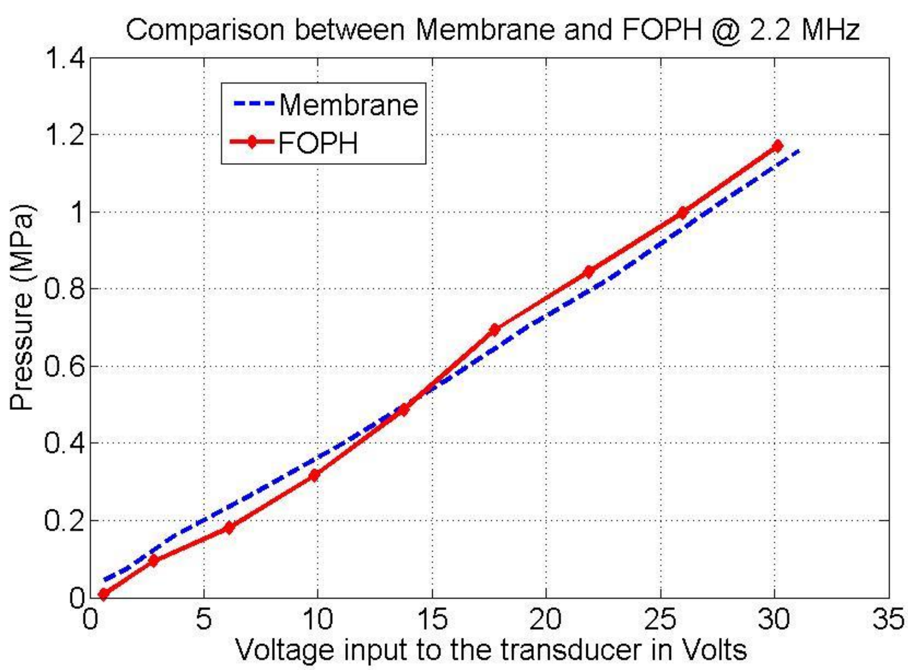

Figure 3.6: Comparison of Membrane and FOPH at 2.2 MHz

Figure 3.7 shows the waveform signal of membrane hydrophone, while figure 3.8 shows the output of the FOPH. Transducer is operated in burst mode of 25 cycles and burst period of 10 msec. Figure 3.9 shows the comparison of the membrane and FOPH outputs.

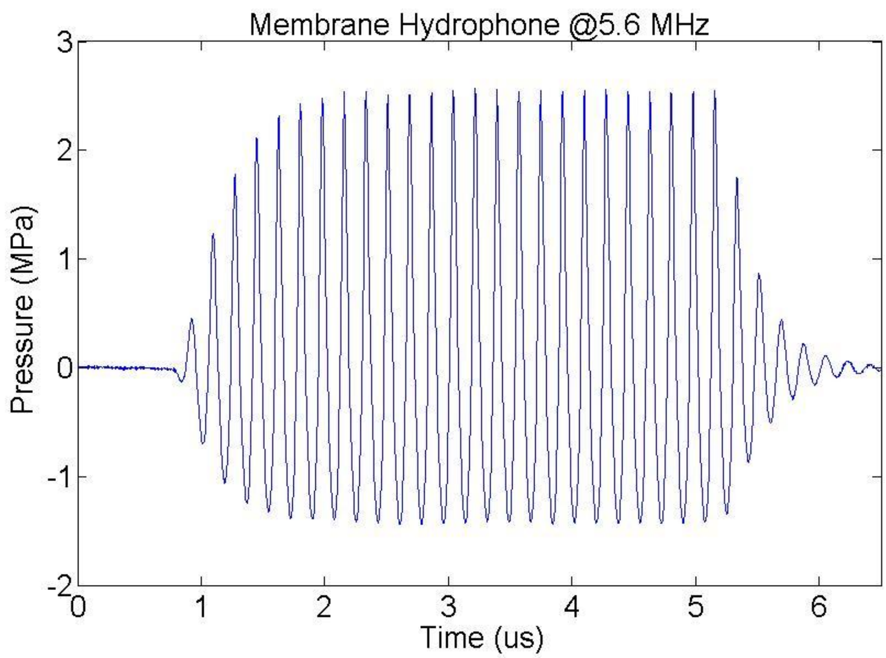

Figure 3.7 Membrane hydrophone signal for 5.6MHz transducer

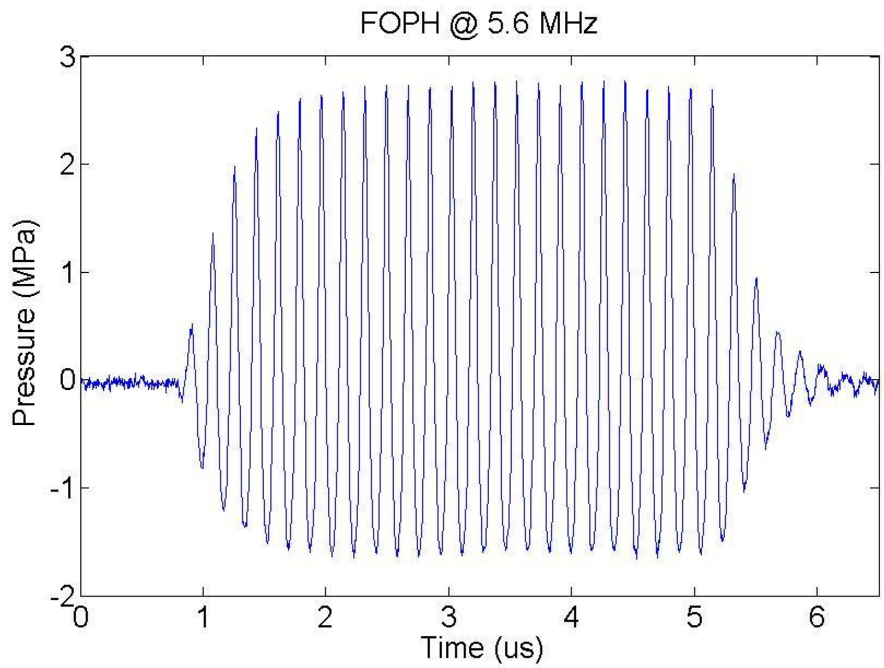

Figure 3.8: FOPH signal for 5.6 MHz transducer

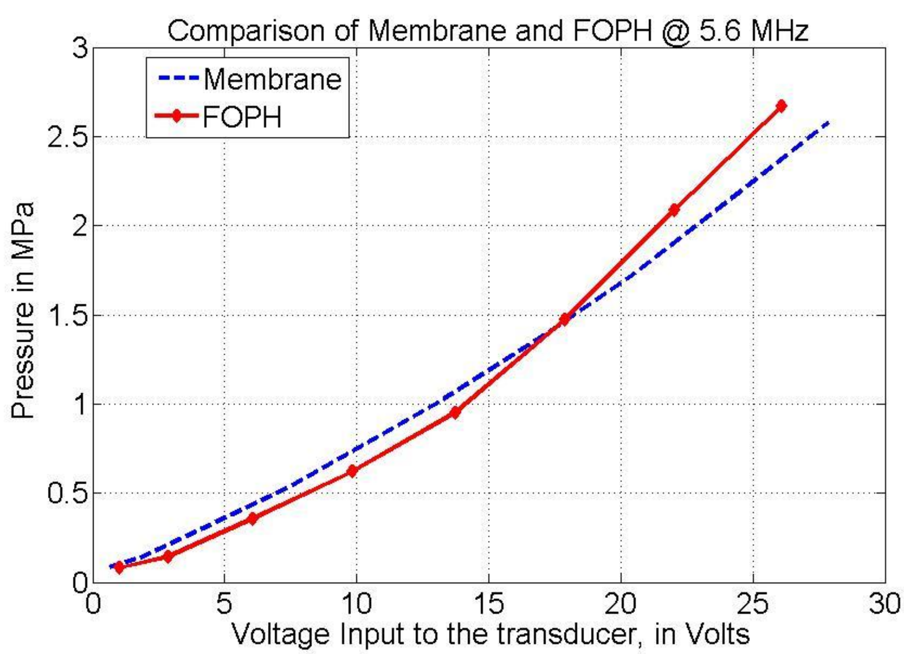

Figure 3.9 Comparison of membrane and FOPH at 5.6 MHz

3.2 Shockwaves Measurement Results

The FOPH was used to measure the shockwaves for two shockwave therapy devices and their respective transducer probes, i.e. the Storz Probe and the Daeyang Probe. The shockwaves results from both measurements are displayed in the following figures. Measurements were performed in water degassed up to 35% of saturation to avoid cavitation. The results show shockwaves resulting from different input pressure settings varying from 2 to 5 bar of the SWT devices. The acoustic pressures measured during the process were in the range of 4 MPa to 20 MPa. Specifications provided by the manufacturer for the Storz probe were 18MPa ±10% at 5 bar input pressure level. Measured values were found to be in agreement with these specifications. The acoustic pressure values were reduced at lower energy levels as indicated in the results. Shockwave pattern is clearly visible, but averaging for shockwaves was not possible, therefore, the pressure levels around 5 MPa have a high level of noise. Optical fiber was placed 1 cm away from the probe tip during the measurements.

Cavitation signals were observed during the measurement process when the magnitude of received signals suddenly went up to the level of volts. This high amplitude was observed as the result of microbubbles present between the probe tip and fiber optic, as the greater mismatch of refractive index could result in such high amplitudes [5, 7, 8].

3.2.1 Storz Probe Results

The results for the Storz probe are shown in following figures. The Storz medical shockwave therapy (SWT) device was operated from 2-5bar input pressure levels during the measurement process. Output pressure amplitudes are higher for high input pressure settings and reduce naturally at lower input pressure settings although the shockwave pattern is still visible. The noise level at 2 bar pressure level is high, which makes it difficult to identify the signal. Waveform averaging is not possible for measuring shockwaves; therefore, the output of the FOPH shows a high noise level. The pattern of waveforms is similar for all energy levels, though the rarefaction pressure is significantly higher at 5 bar input pressure setting, as compared to other input pressure levels.

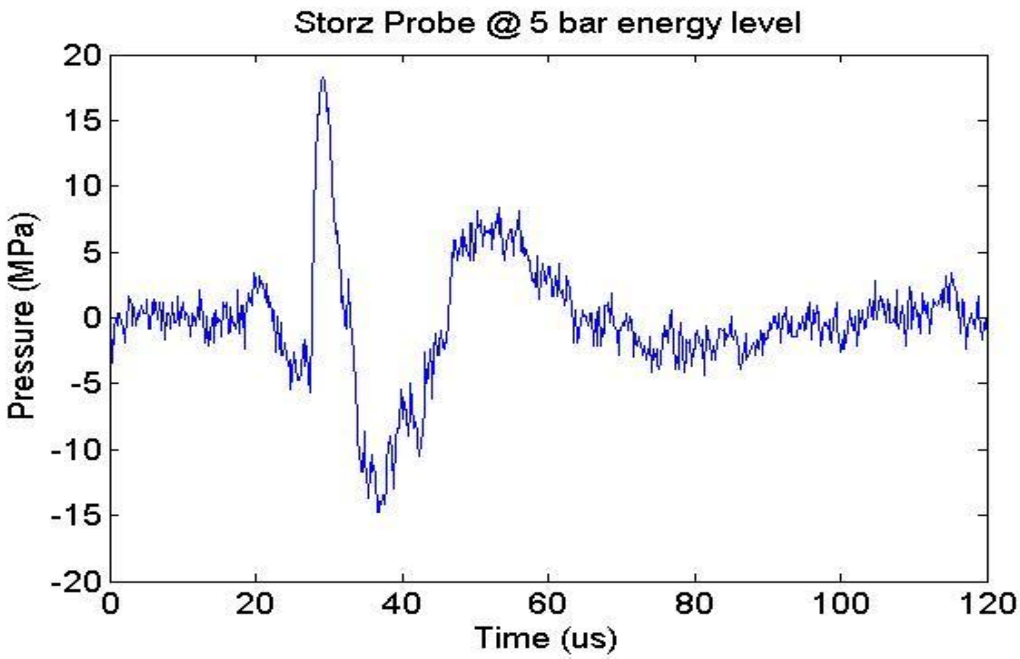

Figure 3:10 Storz probe output at 5 bar pressure setting

Figure 3.10 shows shockwave generated by Storz probe when the machine was operated at 5 bar (highest) energy level.

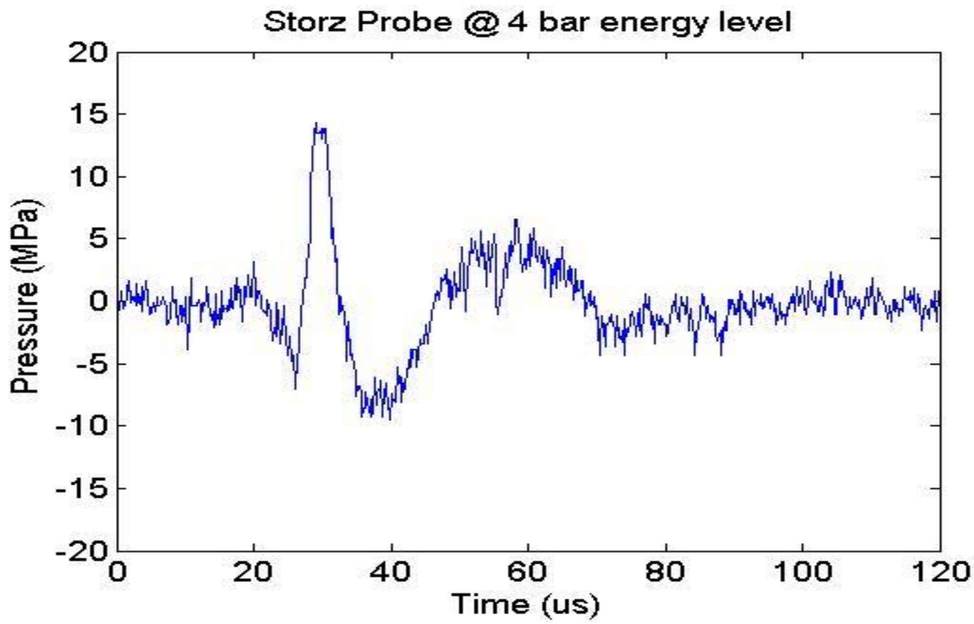

Figure 3.11 Storz probe output at 4 bar pressure setting

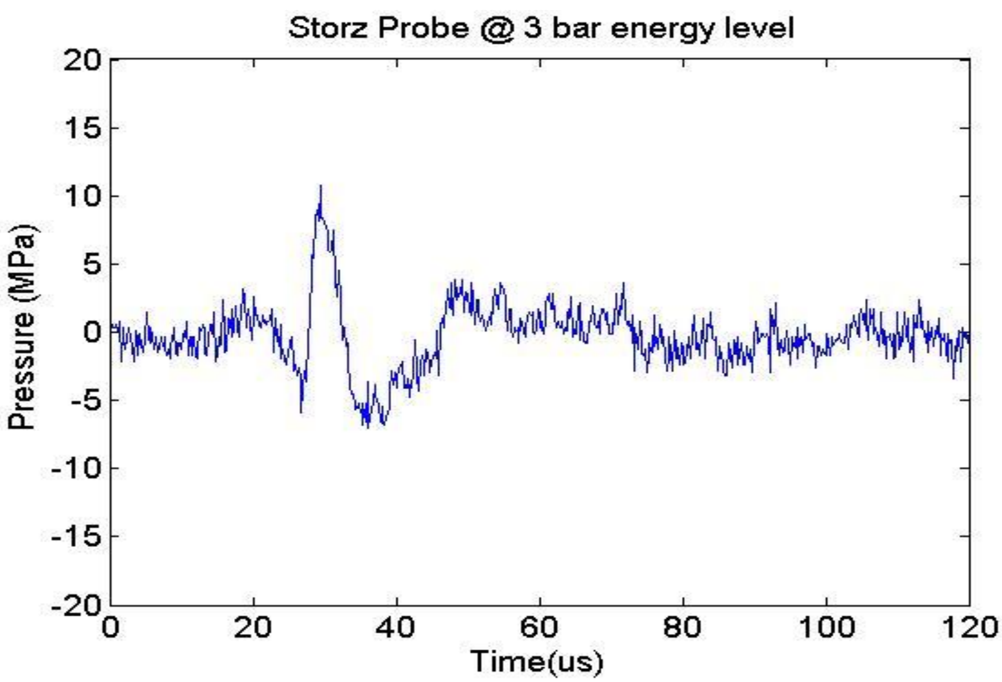

Figure 3.12 Storz probe output at 3 bar pressure setting

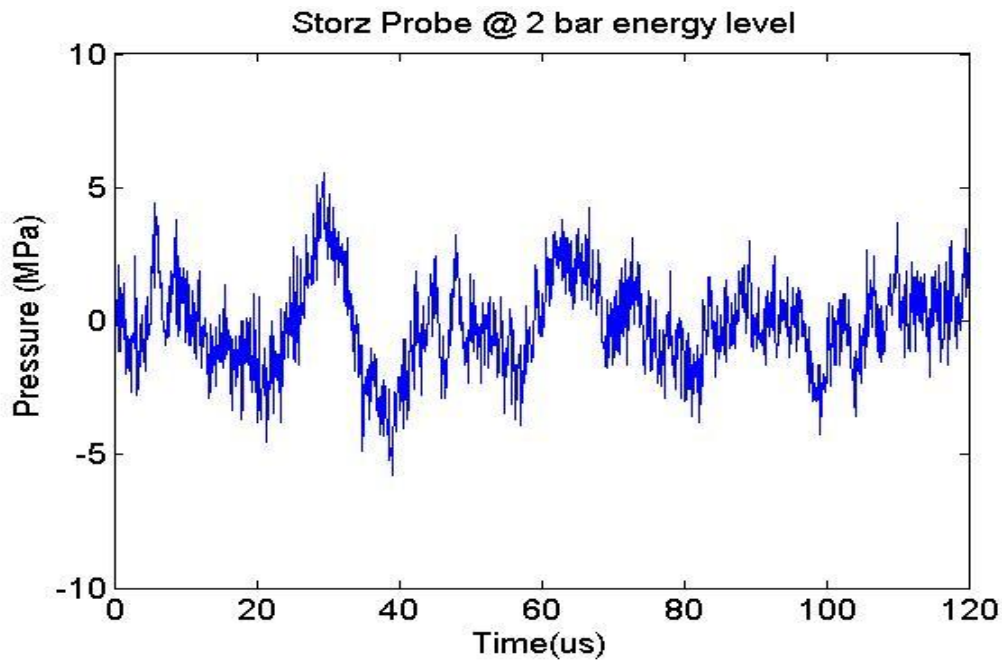

Figure 3.13 Storz probe output at 2 bar pressure setting

Figures 3.11 ,3.12 and 3.13 show the shockwaves generated by Storz probe according to the respect input pressure settings of 4,3 and 2 bars. The pressure amplitudes recorded during the experiments for Storz probe are shown in table 3.1.

Table 3.1 Pressure amplitudes for Storz probe.

Energy (bar)	Compressional Pressure (MPa)	Rarefactional Pressure (MPa)
5	19.33 ± 10%	−14.78± 10%
4	13.89 ± 10%	−9.22 ± 10%
3	9.22 ± 10%	−6.78± 10%
2	6.52 ± 10%	−5.78± 10%

3.2.2 Daeyang Probe Results

Results from the Daeyang transducer probe (connected to Daeyang SWT) are shown in following figures. The pressure amplitudes are smaller as compared to the Storz SWT. Shockwaves can be seen at the input pressure levels of 5 bar and 4 bar; but at the lower pressure settings, the shockwave pattern is difficult to observe. The output pressure amplitudes are also reduced significantly.

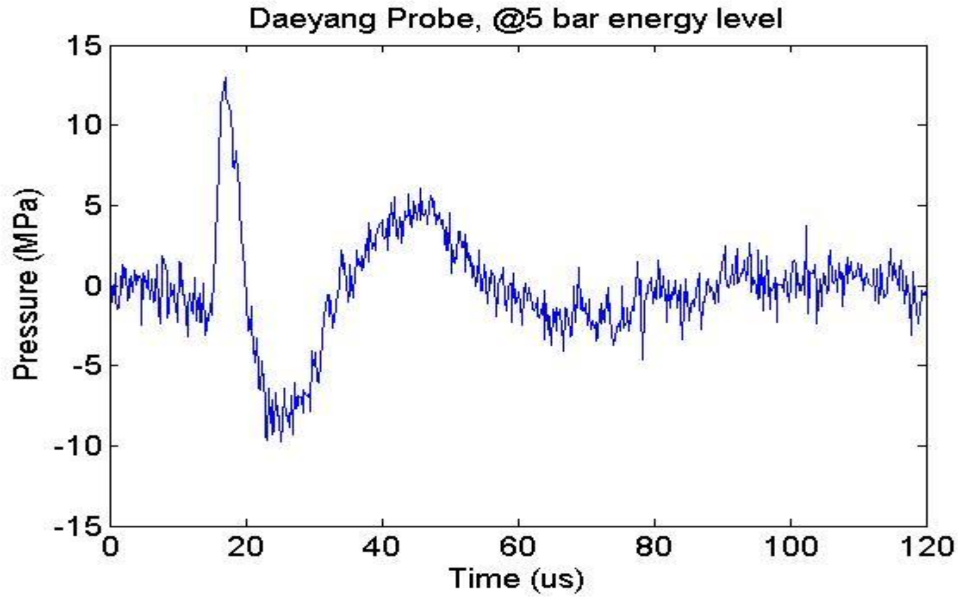

Figure 3.14 Daeyang probe output at 5 bar pressure setting

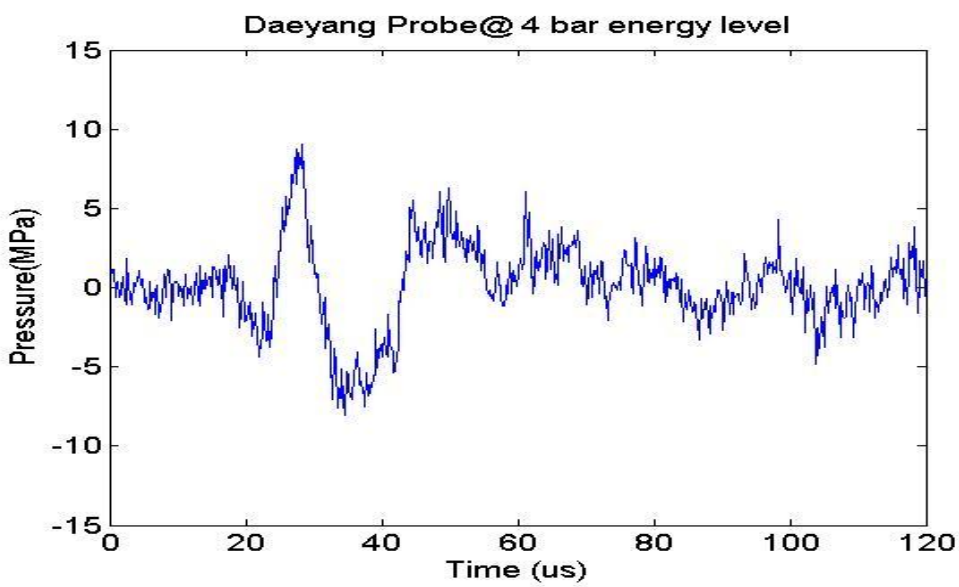

Figure 3.15 Daeyang probe output at 4 bar pressure setting

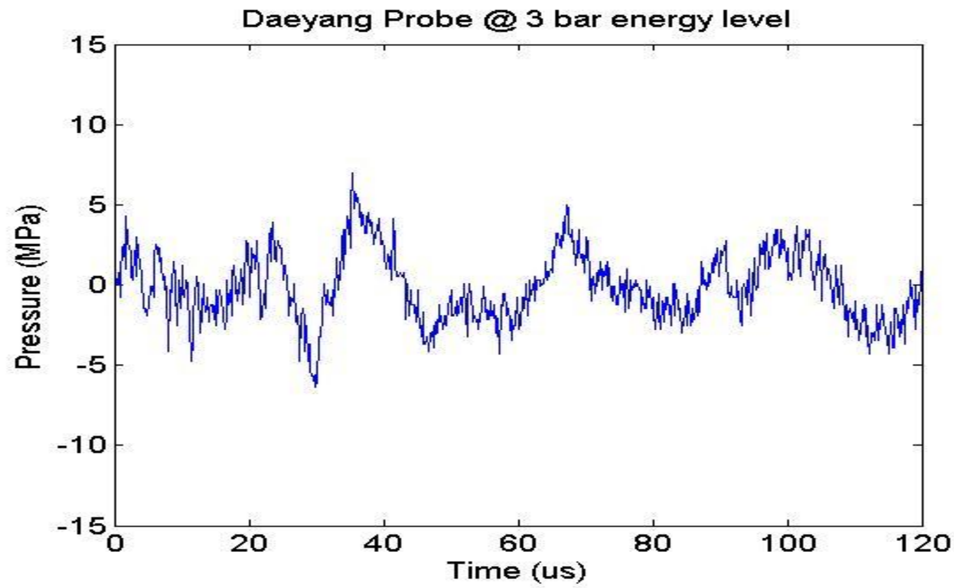

Figure 3.16 Daeyang probe output at 3 bar pressure setting

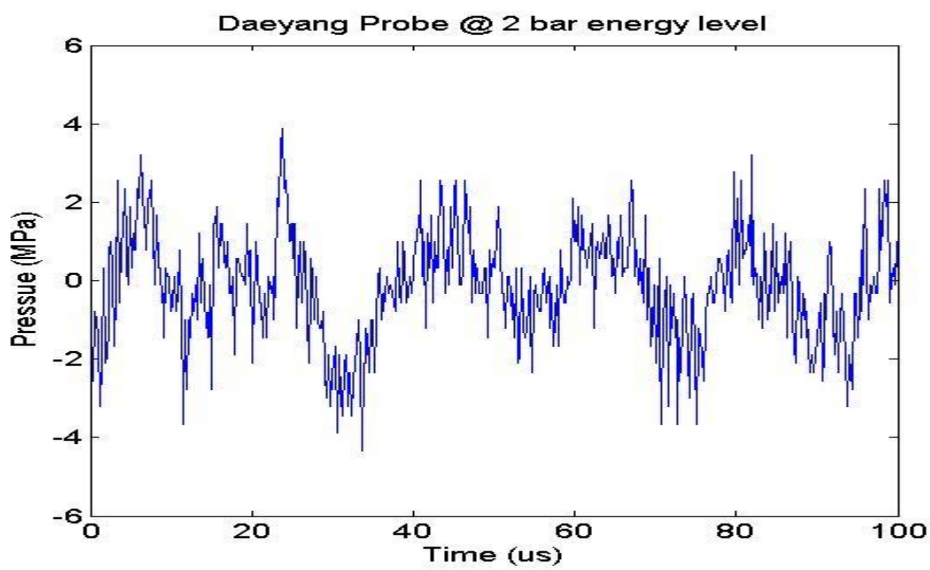

Figure 3.17 Daeyang probe output at 2 bar pressure setting

The pressure amplitudes recorded during the experiments for the Daeyang probe are shown in table 3.2.

Table 3.2 Pressure amplitudes for Daeyang probe

Energy (bar)	Compressional Pressure (MPa)	Rarefactional Pressure (MPa)
5	12.94± 10%	-9.29± 10%
4	9.05± 10%	-8.06± 10%
3	7.00± 10%	-6.33± 10%
2	3.9± 10%	-3.44± 10%

Chapter 4

Discussion

HIFU measurements require a robust hydrophone system which can withstand high pressures and provides adequate sensitivity, a small detector size and a wide bandwidth [15]. Increased bandwidth is necessary to measure acoustic pressures which are associated with nonlinear propagation [26] and it is typically advantageous in the measurement of shockwaves. Small detector size enables the hydrophone system to avoid spatial averaging [3]. The FOPH system which has been developed for HIFU and shockwave measurements provides an optimal solution for the above mentioned requirements.

The FOPH system was calibrated by comparing its voltage output with a calibrated 0.2mm membrane hydrophone using three different resonance frequency transducers of 1.1MHz, 2.2 MHz and 5.6 MHz. The sensitivity was determined as 3.6±0.1mV/MPa. The output results indicate a close agreement between the outputs of the FOPH and the membrane hydrophone at different levels of pressure. The output response of the FOPH can be considered linear and frequency independent. The waveforms of the membrane hydrophone and the FOPH also exhibit similarity. Waveform integrity is necessary because the pressure and intensity calculations are based on temporal waveform [18]. During the calibration procedures, waveforms of the FOPH were averaged 512 times to improve signal to noise ratio, however, at higher pressure levels the need for averaging is reduced and 64-256 times averaging shows acceptable results.

The core diameter of optical fiber used in FOPH system is 100μm which forms the active detector area and eliminates the need for spatial averaging in the range of HIFU frequencies. High-power pig tailed laser diode ensures that adequate laser light intensity is launched in the optical fiber. Additionally, the silicon photodetector in combination with a 40 dB preamplifier results in the sensitivity of the current system to be significantly higher than the one used by Jessica et al [1], who developed a system with 0.8mV/MPa. Minimum acoustic pressure measured during calibration procedures is about 0.1 MPa, which is significantly better than that of 0.9MPa which is demonstrated by Jessica et al [1].

Proper cleaving of the fiber tip is required for repeatable and reliable results. Cleaving problems may result in change of reflectivity at the fiber endface. Therefore, the tip of the optical fiber was inspected before every measurement process using the fiber inspection probe and recleaved in case it was cracked. During the measurement process, loss of waveform integrity and unusually high signals would relate to cavitation at the fiber tip [3]; however, it was noted that occasional cavitation events don't necessarily damage the fiber tip which shows robustness of the optical fiber in high pressure acoustic fields.

Alignment procedures are extremely important to get repeated results. The pressure outputs vary significantly if the transducer and hydrophone are not properly aligned. Therefore, alignment procedures were standardized using pulser/ receiver system and followed every time to align the transducer and fiber optic. Angular movements were achieved by worm gear and belt gear systems while translation motions in 3-D were achieved by using a 3D motorized stage and a manual stage. This positioning system allowed standard alignment procedures to achieve repeatable results. In order to place the hydrophone at the focal spot of the transducer, the 3-D motorized stage was used to move the transducer and obtaining the peak output of hydrophone.

The repairable probe tip and broader bandwidth makes the FOPH a desirable hydrophone for measuring shockwaves. Frequency independent sensitivity gives the FOPH a significant advantage in reliably measuring the pressure amplitude of shockwaves. Since it is not possible to perform averaging for shockwaves [5], the noise level is high in the signal. Shockwaves were measured for two ballistic shockwave therapy devices. The results exhibited consistency and reliability of the FOPH in measuring shockwaves at higher pressures, which could otherwise damage other PVDF hydrophones. The results of the Storz probe were found to be in agreement with the specifications provided by the company. After that shockwaves from Daeyang probe of Daeyang medical SWT were measured and the results were compared with that of Storz probe. Daeyang probe results were significantly lower than that of Storz probe. This may have been due to the mechanical differences in both devices. The results were having output pressure amplitudes in the range of 6 MPa to 20 MPa for Storz probe and in the range of 4 to 13 MPa for Daeyang probe. Higher noise level created a limitation to observe the shockwaves below 2 bar input pressure settings.

During the measurements, higher amplitude signals were observed in the case of cavitation. Especially during the measurements of shockwaves, when signals were having the amplitude in the range of volts, indicating an exceptionally high level of reflectance, which might be the result of refractive index mismatch of the microbubbles trapped between the fiber optic and the transducer probe; this effect was also mentioned by Huber et al [5]. The higher mismatch results in higher reflectivity and saturated signals in the range of volts. Considering the sensitivity was in the range of mV/MPa, these signals were highly unusual. A reasonable interpretation is that they were reflected from the microbubbles, whose refractive index was close to that of air, and hence they resulted in signals of extremely high amplitude. In order to reduce the cavitation events, water was filtered and degassed which helped in reduction of cavitation activity; as gas content plays a pivotal role during the measurement processes. Water should be significantly degassed to measure high pressures and avoid cavitation events. Still it is difficult to completely avoid the high amplitude signals, but their occurrence is limited.

Optical fiber was placed at 1 cm away from the tip of transducer probe in shockwave measurements[7]. At this distance, the output pressures for Storz probe were in the specified range and indicated a shockwave pattern. Therefore, the same distance was used for Daeyang probe as well to maintain similar conditions. Also, when the optical fiber tip was moved closer to the transducer probe tip, at a distance of 0.2-0.5mm, saturated signals were received more often which may be the result of perturbation or microbubbles present.

Optical coupler needs to be replaced once the sensing fiber becomes too short as a result of cleaving because the fiber is permanently integrated to the coupler. Permanent connection provides the advantage of avoiding the losses associated with removable connection.

The sensitivity can also be improved by coating the fiber tip with some dielectric or polymer film deposition, but since the fiber has to be recleaved often in case of measurement of shockwaves, it can be a tedious process to recoat the fiber tip [15, 17].

Chapter 5

Conclusion

The FOPH system was developed and tested in this research. Sensitivity was found to be 3.6mV±0.1mV/MPa. Measurements of shockwaves generated from a ballistic shockwave therapy device were successfully demonstrated in this research. The FOPH has been tested and verified to be a reasonable alternative to measure high intensity acoustic pressure and shockwaves.

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Abstract (In Korean)

본 연구는 높은 에너지의 음향 출력을 발생하는 Shockwave 시스템이나 High Intensity Focused Ultrasound (HIFU)시스템의 음장 및 음압을 측정하기 위한 Fiber Optic Probe Hydrophone(FOPH) 시스템 개발에 대한 연구이다. FOPH 시스템은 기존의 PVDF 하이드로폰에 비해 다양한 이점을 가진다. 기존 하이드로폰은 높은 음향 출력이나 Shockwave 출력 측정 시 캐비테이션 발생으로 인해 센서에 심각한 손상을 입는 경우 재생이 불가능하다. 그러나 FOPH의 경우 센서 손상 시 클리빙과 폴리싱 과정을 통해 쉽게 재생이 가능하며, 주파수 대역과 상관없는 일정한 감도를 가진다. 또한 작은 직경의 Fiber 사용으로 공간 분해능을 높일 수 있으며 전자장 간섭에도 영향을 받지 않는 이점이 있다. 제안된 시스템의 구성은 고출력의 pigtailed 레이저 다이오드 모듈(1.5W@850nm)과 2X2의 3dB 광 커플러가 FC-FC 어댑터로 연결되어 있으며 광 커플러 내부는 100/140um의 multimode 광섬유로 이루어져 있다. 커플러를 통해 발생된 신호는 초음파 에너지와 상호작용하여 실리콘 포토 디텍터 (0.55A/W@850nm)로 전달되어 전기신호로 변환되며 이 신호는 40dB 신호 증폭기를 통해 증폭되어 오실로스코프를 통해 확인 가능하다. 제안된 시스템은 중심주파수 1.1MHz, 2.2 MHz, 5.6 MHz의 초음파 변환기를 사용하여 맴브레인 하이드로폰에

의해 측정된 데이터를 기준으로 비교 교정되었다. 측정된 FOPH의 감도는 3.6±0.1mV/MPa이며 이 값은 주파수에 독립적이다. 제안된 FOPH 시스템을 통해 측정된 초음파 신호는 측정에 사용된 모든 초음파 변환기에서 맴브레인 하이드로폰을 통해 측정된 신호의 특성과 일치함을 볼 수 있다. 또한 고출력 신호인 충격파 측정을 위해 두 종류의 ballistic 충격파 발생기를 사용하여 음향 출력을 측정하였다. FOPH 시스템에 의해 측정된 Shockwave의 compressional pressure의 범위는 4~20MPa이며 rarefactional pressure의 범위는 -3~-15MPa로 측정되었다.