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Optimization of un-tethered, low voltage, 20–100 kHz flexural transducers for biomedical ultrasonics applications

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Abstract

This paper describes optimization of un-tethered, low voltage, 20–100 kHz flexural transducers for biomedical ultrasonics applications. The goal of this work was to design a fully wearable, low weight (<100 g), battery operated, piezoelectric ultrasound applicator providing maximum output pressure amplitude at the minimum excitation voltage.

Such implementation of ultrasound applicators that can operate at the excitation voltages on the order of only 10–25 V is needed in view of the emerging evidence that spatial-peak temporal-peak ultrasound intensity (I_{SPTP}) on the order of 100 mW/cm² delivered at frequencies below 100 kHz can have beneficial therapeutic effects. The beneficial therapeutic applications include wound management of chronic ulcers and non-invasive transdermal delivery of insulin and liposome encapsulated drugs.

The early prototypes of the 20 and 100 kHz applicators were optimized using the maximum electrical power transfer theorem, which required a punctilious analysis of the complex impedance of the piezoelectric disks mounted in appropriately shaped metal housings.

In the implementation tested, the optimized ultrasound transducer applicators were driven by portable, customized electronics, which controlled the excitation voltage amplitude and facilitated operation in continuous wave (CW) or pulsed mode with adjustable (10–90%) duty cycle. The driver unit was powered by remotely located rechargeable lithium (Li) polymer batteries. This was done to further minimize the weight of the applicator unit making it wearable. With DC voltage of approximately 15 V the prototypes were capable of delivering pressure amplitudes of about 55 kPa or 100 mW/cm² (I_{SPTP}). This level of acoustic output was chosen as it is considered safe and side effects free, even at prolonged exposure.

Keywords

Ultrasound low frequency (<100 kHz) transducers; piezoelectric flexural transducers; therapeutic ultrasound; maximum power transfer transducer optimization

1. Introduction

This paper describes optimization of un-tethered, low voltage, 20–100 kHz flexural transducers for biomedical ultrasonics applications. The goal of this work was to design a fully wearable, low weight (<100 g), battery operated, piezoelectric ultrasound applicator providing maximum output pressure amplitude at the minimum excitation voltage.

Such implementation of ultrasound applicators that can operate at the excitation voltages on the order of only 10–25 V is needed in view of the emerging evidence that spatial-peak temporal-peak ultrasound intensity (I_{SPTP}) on the order of 100 mW/cm² delivered at frequencies below 100 kHz can have beneficial therapeutic effects. These effects include, for instance, wound management of chronic ulcers and non-invasive, transdermal delivery of insulin and liposome encapsulated drugs [1–4].

The early prototypes of the 20 and 100 kHz applicators were optimized using the maximum electrical power transfer theorem, which required a punctilious analysis of the complex impedance of the piezoelectric disks mounted in appropriately shaped metal housings. The term optimization in this paper refers to the overall performance of the flexural transducers and is defined as the process of maximizing the output pressure amplitude whilst minimizing the applicator's excitation or driving voltage. Performance was chosen as the primary parameter for optimization, because the portability of the device is crucial for its possible clinical applicability and depends primarily on the number and size of rechargeable Li-Polymer batteries needed, and this in turn, depends on the maximum voltage and current that are required to drive the applicator.

To gain insight into the optimization procedure described in the following it might be useful to examine the current state of the art in low frequency (LF) transducer design and to identify the deficiencies impeding its immediate use in clinical applications. Although there exists substantial literature on the topic of flexural ultrasound transducers operating at frequencies below 100 kHz, in the following only the papers that were of direct relevance in the optimization procedure applied here are reviewed. Also, a brief discussion related to the selection of the frequency range of operation is included.

It should be noted that the ultrasound applicators considered in this work were intentionally designed for operation in 20–100 kHz range and this is in contrast to the fact that the majority of therapeutic applicators operate in the frequency range 1–3 MHz [1]. Despite the fact that the distinction between the different ultrasound intensities [5], such as spatial-peak temporal-peak (I_{SPTP}), spatial-average temporal-peak (I_{SATP}), spatial-average temporal average (I_{SATA}), and spatial-peak pulse-average (I_{SPPA}), were not always given in [1], this review was considered to be of interest because it covered relatively wide interval of intensity levels ranging from 0.5 mW/cm² to 3000 mW/cm².

As mentioned earlier the choice of frequency of operation in this work was selected to be an order of magnitude lower than that reviewed in [1] and was prompted by emerging evidence that LF ultrasound exposure could be more effective than that in the 1–3 MHz range. Specifically, Mitragotri [6] and his co-workers from Dr. Langer's laboratory at MIT demonstrated *in vitro* that the therapeutic effects of ultrasound related to mammalian tissue were more prominent in the low (10–100 kHz) range than those observed at higher (1–10 MHz) frequencies [6]. However, implementation of the LF ultrasound treatment in clinical practice is still impeded by the fact that the acoustic sources capable of generating ultrasound energy in 20–100 kHz bandwidth are in general unwieldy, bulky, and require heavy (25+ kg) electronic setup; for example a typical applicator driver design described in [2] included a function generator weighing about 2–8 kg and the ENI Power amplifier weighing on the order of 25 kg.

In addition to the need of using bulky equipment, there is one more aspect that contributed to the lack of the availability of portable therapeutic ultrasound devices in this frequency range and prompted the departure from the conventional mode of design of piezoelectric, thickness-polarized materials. In this design utilizing fundamental mode of vibration, in which the PZT thickness is equal to half-wavelength ($\lambda/2$) at the resonance frequency, the

solid or composite PZT ceramic thickness has to be on the order of 20 mm at 100 kHz and fivefold that at 20 kHz. Such thickness not only increases the weight and bulk of the ultrasound transducer, more importantly, it requires excitation voltages exceeding 100 V_{RMS} to produce acoustic output that could be considered clinically relevant, say on the order of 50–100 mW/cm² (I_{SPTP}).

Kost and his coworkers [7] proposed to employ a customized (stacked-ceramic) design transducer. Although their 42 kHz transducer was capable of delivering ultrasound intensity exceeding 2.5 W/cm² [7] it still required 100 V_{RMS} excitation voltage. The transducer was used in the set of well-conceived experiments permitting the research team to conclude that the observed sonoporation was due to inertial cavitation.

In contrast, the prototype of the applicator described here utilizes flexural mode of vibration and produces much lower intensity (100 mW/cm² I_{SPTP}), preventing the onset of inertial cavitation. Whereas flexural transducers were described as early as 1930's [8] their most practical implementation was proposed by Newnham and his Penn State team [9] almost five decades later. Their contributions are elegantly detailed in several scholarly publications [10], [11–12] and PhD dissertations of Tressler [13] and Snyder [14]. These publications described an acoustic source in which its individual elements could be arranged in 2×2 or 3×3 arrays referred to as moonie and cymbal transducers. Subsequently, these arrays were used in a successful non-invasive transdermal delivery of insulin using live pigs [2]. This *vivo* result can be considered as a milestone in ultrasonically assisted drug delivery because demonstrated that relatively low level (100 mW/cm² I_{SPTP}, below the inertial cavitation threshold) of acoustic energy was sufficient to push large molecular weight (>5000 Da) particles through the stratum corneum, which is otherwise impermeable for particles of this size. However, the array used for insulin delivery in [2] also required excitation voltages exceeding 100 V_{RMS} and the use of ancillary electronic setup including 55 dB ENI power amplifier. As indicated above the weight of this instrument alone about 25 kg and hence prevents any portable clinical application.

The flexural transducer proposed by Newnham [9] whose careful modification is described in this work lends itself to generate frequencies in 20–100 kHz bandwidth considered here. It uses (Fig. 1) thin (1 mm) standard Navy Type I ceramic disc. This material was selected due to the ease of manufacturing and acquisition even though other PZT types (such as PZT-8 or Navy type III) exhibit a slightly better transmitting voltage response. The disk is capped at both sides by "dome-like" shaped metal caps, which are firmly glued to the PZT ceramic using an adhesive. This adhesive introduces specific (displacement) boundary conditions and, as discussed in the following sections, is one of the key parameters governing the performance of the transducer. The design described in this work offers not only portable but fully wearable implementation of the device and its systematic step-by-step development including analysis of electromechanical parameters is presented. As already noted the ultimate goal of this research was develop a transducer that would deliver maximum displacement amplitude with the lowest possible excitation voltage and be capable of tether-free, self-contained battery operation for 4-5 hours (considered clinically relevant).

In the following sections the principle of operation of flexural transducers suitable for producing clinically relevant acoustic output in the considered bandwidth of 20–100 kHz is discussed, along with key parameters governing the acoustic output and frequency. Next, practical implementation of the early prototype is presented and a succinct summary of the experimental data obtained with the optimized design is given.

2. Theory of operation

Prior to presenting the optimization results, the principle of operation of the flexural transducer design optimized in this work is briefly described [15–16]. It is well known that piezoelectric material generates a voltage when under stress and deforms when a voltage is applied across it [15–16]. The voltage to pressure conversion was optimized making use of flexural operation [10, 17]. Briefly, this operation is controlled by the shaped metal cap bonded on the vibrating surface of the piezoelectric element (Fig. 1 Fig. 2). The vibration of the PZT disk in response to the excitation voltage applied across its electrodes causes it to change both its thickness and radial dimensions (Fig. 1b). As the metal caps are firmly bonded to the disk, the change in dimensions of the disk causes the shaped metal disk to flex or bend. This type of motion exercised by the shaped metal cap is called flexural motion. A more rigorous description of the theory of operation of flextensional transducers can be found in ([13], [14], [17]). These publications examine the role of the metal cap in more detail, specifically they explain how the cap aids to convert and maximize the force of contraction (or extension) of the piezoelectric surface into a amplitude displacement that - for a given voltage - is approx 5–10 fold that of a free vibrating PZT disk, which in turn translates to maximizing the achievable pressure amplitude. Furthermore, the use of the metal caps introduces an additional resonance frequency besides the fundamental ($\lambda/2$) thickness resonance. The numerical value of this additional resonance depends upon the following parameters: piezoelectric material type and geometry (thickness and diameter), the properties of metal cap material (density, Poisson's ratio, and Young's modulus), and its dimensions (thickness, height, diameter, apex diameter, and base diameter); see Fig. 2.

It is instructional to detail the principle of operation depicted in Fig. 1. Referring to this figure it can be seen that during the positive half wave or cycle the top surface electrode of the PZT material moves in the vertical Y direction. Concurrently, the shaped cap moves along in the same direction. Along with this movement the curved radial surface of the PZT (Fig 1b, Fig 2) moves inwards in radial X direction. This motion causes the cap that is bonded to the surface to convert the inward movement of the radial surface into a motion in the thickness direction (i.e. perpendicular to the disc surface), which is considered flexural motion. Hence, both the upwards (vertical) and inwards (radial) motions associated with the positive cycle are converted to a flexural motion in the thickness direction by the shaped metal caps. Consequently, the flexural transducer converts the force of radial contraction of the PZT into amplified vertical displacement. As already pointed out, the displacement amplitude is amplified comparison with that achievable using a fundamental mode of thickness vibration by the PZT disk alone. Furthermore, the displacement is generated at a frequency considerably lower (1–2 orders of magnitude) than that corresponding to the fundamental ($\lambda/2$) resonance of the disk thickness.

3. Construction

Fig. 2 and Table 1 show and list, respectively, the geometric parameters influencing optimization process for the ultrasound applicator considered here, along with the symbols used and the typical values applied in the early prototype development.

Here, it is appropriate to reiterate that, in this paper attention is focused on presenting the results fulfilling the primary goal, i.e. fabricating a prototype capable of generating required pressure amplitude (intensity) with minimum excitation voltage. A more detailed discussion of the influence of the parameters listed in Table 1 and other parameters determining the performance of the applicator is going to be presented in a separate paper. Therein, the optimization of the transducer parameters for maximum pressure output at a given voltage using Finite Element (FE) analysis in COMSOL is described along with the specific details

on FE modeling in COMSOL. The construction of the prototype used in the experiments described in the next section was preceded by meticulous modeling. The modeling process helped to develop a supplementary production test measurements needed to optimize the prototypes performance in terms of operating frequency and displacement amplitude.

The simulation results compared well (within 10%) with those experimentally determined and with those obtained using similar analysis performed employing other (ANSYS, ATILA) finite element software programs [18]. The results presented in Section 3 were obtained using flextensional ultrasound applicators made with brass caps. This material was chosen because of its availability and relatively easy machining. A punch-die was used to create brass disks from various thickness (0.125 - 0.25 mm) brass sheets. These disks were then carefully sanded to remove burrs and subsequently shaped using a custom-machined stainless steel shaping die. The Navy Type I equivalent (PZT-840) disks were acquired from American PiezoCeramics (State College, PA) with existing silver electrodes on both sides. A bonding epoxy was initially mixed as recommended by the manufacturer and then again mixed with silver powder to make it electrically conductive. The epoxy was then used to bond the shaped metal caps to both sides of the PZT disk (Fig. 1, Fig. 2). After the bonding epoxy has cured, silver epoxy was used to attach voltage signal wires to both sides of the metal end caps. Such made individual elements were then potted in polyurethane epoxy, in different spatial arrangements. Several – up to 9 – individual elements were combined to fabricate 2×2 or 3×3 arrays having footprint 35×35 or 55×55 mm respectively. This footprint can be adjusted depending on the specific clinical application needs.

4. Method: Measurement and experimental results

The final optimization of the acoustic performance was achieved by designing the complex conjugate network ensuring maximum power transfer condition. As the electrical impedance of the applicator depends on the mechanical loading, the impedance measurements were performed at each stage of applicator manufacturing to optimize fabrication process and minimize the likelihood of malfunctioning of the completed device.

All of the measurements presented in this paper were done using AIM 4170C antenna network analyzer and the guidelines provided in [16]. The analyzer measures the complex impedance in terms of resistance and reactance. Both the real and imaginary components of the impedance are needed separately to properly design the complex conjugate network. Although the AIM 4170C device provides auxiliary data such as magnitude of the impedance and Q value, only capacitive reactance and resistance are plotted here from 10–50 kHz, as only these two were used in the optimization process. Fig. 3 shows the typical complex impedance plots obtained prior to (top plots) and after (bottom plots) applying matching circuitry. The thick black line represents the resistance, and the grey line represents the capacitive reactance. The top graph in Fig. 3 shows the capacitive reactance approaching zero and exhibiting resonance behavior (Fig. 3a at approx. 26 kHz) [15]. The resistance also behaves in a similar way, this behavior helps to identify the low frequency (here about 26 kHz) resonance introduced by the flexural mode. It can be seen that the capacitive reactance does not reach zero, preventing the transducer from reaching ideal resonance operating conditions, at this frequency.

To overcome this shortcoming, maximum power transfer condition from the driving network the transducer must be ensured. This condition is achieved when the output impedance of the driving stage and the input impedance of the applicator are matched. The effectiveness of the complex conjugated matching process is demonstrated in the bottom or lower graph in Fig. 3 where it is shown that the phase shift present in unmatched circuit was

eliminated. The matching reduced the value of the reactance to virtually equal to zero Ohms and ensured the input resistance of the applicator to be close to 50 Ohms.

Fig. 4 summarizes the experimental data obtained in water at 22°C. The pressure amplitude another prototype operating at approximately 20 kHz was measured using a calibrated hollow cylinder hydrophone similar to that described in [19]. This hydrophone was used to obtain the results presented Fig. 4 and Fig. 5. The hydrophone's diameter was on the order of 2 mm to ensure that it was substantially lower than the shortest wavelength measured here (15 mm at 100 kHz) to minimize acoustic field disturbance. The hydrophone exhibited relatively uniform (± 2 dB) sensitivity (about $3.83 \mu\text{V}/\text{Pa}$) in frequency range 20 kHz – 500 kHz and a useable (-10 dB) response between 10 kHz and 800 kHz.

For comparison with the available literature data [2] the hydrophone recorded pressure amplitude signal was converted to intensity using a plane wave approximation. As shown in Fig. 4 the optimized design exhibited spatial-peak temporal-peak intensity (I_{SPTP}) level of $100 \text{ mW}/\text{cm}^2$ at the excitation level of approximately 15–20 V_{p-p}. For comparison, the measurements were also performed using the same hydrophone sampling the field produced by a similar ultrasound applicator used in insulin delivery experiments described in [2] the only publication that is immediately comparable to the design presented here. In [2] an order of magnitude higher (130 V) excitation voltage was needed to generate approximately $80 \text{ mW}/\text{cm}^2$ (I_{SPTP}) intensity level.

Fig. 5 shows the two-dimensional field distribution (XY plane) of the applicator with characteristics shown in Fig. 4. The data was recorded at axial distances ranging from $z = 2.5$ to $z = 10$ mm. The shortest distance (2.5 mm) was chosen to be shown in Fig. 5 because it is of immediate clinical relevance for such applications as transdermal drug delivery and wound management and corresponds to the pressure amplitude experienced by (skin) tissue. Maximum pressure amplitude measured at the axial distance of 2.5 mm was 55 kPa ($100 \text{ mW}/\text{cm}^2$ I_{SPTP}). At the axial distance of 8 mm, this amplitude was reduced by 3 dB.

5. Conclusions

The results of the experiments presented above indicate that a careful analysis of the flexural transducer design allows clinically relevant implementation. The early prototype used here to obtain the data shown in Fig. 4 and Fig. 5 weighed only 100 g and was driven by a custom made electronic driver controlling the output pressure amplitude, duty cycle and associated pulse repetition frequency (PRF). The prototype tested permitted operation at any frequency in the 20–100 kHz range and was powered using rechargeable batteries. The maximum pressure amplitude generated was determined to be about 77 kPa or $200 \text{ mW}/\text{cm}^2$ (I_{SPTP}), which is twice the level that was targeted ($100 \text{ mW}/\text{cm}^2$). The availability of intensity that is twice of that targeted initially adds flexibility in the application of the optimized ultrasound applicator. This is because it enhances the examinable range of the ultrasound field parameters, such as pressure or intensity, which affect the therapeutic efficiency of the ultrasound applicator.

As already mentioned, the intensity of $100 \text{ mW}/\text{cm}^2$ is considered clinically safe even for prolonged exposure [20], and is intended for clinically relevant therapeutic applications, including wound management and transdermal drug delivery. Being able to produce this intensity level with merely 15–20 V provides a fully wearable un-tethered ultrasound device, attractive from both the patients' and physicians' perspective, as it potentially increases the patient compliance with physician's recommendations.

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Fig. 1a: Deformation of a PZT disk under excitation voltage

Fig. 1b: The principle of operation of the ultrasound applicator optimized in this work:

Stage 0: Equilibrium State - occurs at zero crossing of the (sinusoidal) excitation voltage

Stage 1: Maximum vertical displacement - occurs at the peak positive cycle of the excitation voltage

Stage -1: Minimum vertical displacement corresponding to the maximum radial extension – occurs at the negative peak of the excitation voltage.

See Fig. 2 for a more detailed drawing of a single element of the applicator.

Fig. 1.

Fig. 2.

Geometry of the single element of the flextensional, ultrasound applicator considered in this work. Several – up to 9 – individual elements were combined to fabricate 2×2 or 3×3 arrays having footprint 35×35 or 55×55 mm – This footprint can be adjusted depending on the specific clinical application needs. See Table 1 for parameter and symbol descriptions.

Fig. 3.

Network analyzer measurements of the early prototype of the ultrasound applicator described here. Intensity versus voltage characteristics are shown in Fig. 4. Horizontal axis: Frequency in kHz, Vertical axis: Resistance (Rs) and Capacitive Reactance (Xs) in Ohms.

Fig. 4.

An example of the data obtained in water (22°C) with the optimized, early prototype of an ultrasound applicator using quasi-continuous wave (CW) and 20% duty cycle.

a. Over 100 mW/cm² (I_{SPTP}) achieved with a mere 15 V of excitation voltage.

b. Corresponding data of a similar applicator design described in [2], the only publication that is immediately comparable to this work. An order of magnitude higher (130 V) excitation voltage was needed to generate lower, approximately 80 mW/cm² (I_{SPTP}) intensity level.

Fig. 5.

Normalized two-dimensional surface pressure amplitude distribution measured at the axial distance of $z = 2.5\text{mm}$ of the applicator having characteristics shown in Fig. 4a.

Y Axis: Left to Right direction, X Axis: Front to Back direction

Grey scale intensity: Light (higher pressure amplitude, higher voltage) to dark (lower pressure amplitude, lower voltage). Maximum pressure amplitude measured at the axial distance of 2.5 mm was 55 kPa ($100\text{ mW/cm}^2 I_{\text{SPTP}}$). The pressure amplitude decreases by a factor of 3 dB at the axial distance of 8 mm. (see text for details)

Table 1

Parameters controlling resonance frequency and maximum displacement amplitude and their typical values.

Parameter	Symbol	Typical value(mm)
PZT Thickness	T	1.0
PZT Diameter	Φ_d	12.7
Cap Metal Thickness	t	0.25
Apex Diameter	Φ_a	3.0
Base Diameter	Φ_b	9.0
Cavity Height	h	0.5