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Multi-modal transducer-waveguide construct coupled to a medical needle

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ABSTRACT:
Annually, more than $16 \times 10^9$ medical needles are consumed worldwide. However, the functions of the medical needle are still limited mainly to cutting and delivering material to or from a target site. Ultrasound combined with a hypodermic needle could add value to many medical applications, for example, by reducing the penetration force needed during the intervention, adding precision by limiting the needle deflection upon insertion into soft tissues, and even improving tissue collection in fine-needle biopsy applications. In this study, we develop a waveguide construct able to operate a longitudinal-flexural conversion of a wave when transmitted from a Langevin transducer to a conventional medical needle, while maintaining high electric-to-acoustic power efficiency. The optimization of the waveguide structure was realized in silico using the finite element method followed by prototyping the construct and characterizing it experimentally. The experiments conducted at low electrical power consumption (under 5 W) show a 30 kHz flexural needle tip displacement up to 200 $\mu$m and 73% electric-to-acoustic power efficiency. This, associated with a small sized transducer, could facilitate the design of ultrasonic medical needles, enabling portability, batterization, and improved electrical safety, for applications such as biopsy, drug and gene delivery, and minimally invasive interventions. © 2023 Author(s). All article content, except where otherwise noted, is licensed under a Creative Commons Attribution (CC BY) license (http://creativecommons.org/licenses/by/4.0/).

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I. INTRODUCTION

The hypodermic needle is one of the most commonly used instruments in healthcare with $16 \times 10^9$ needles consumed annually. They are frequently used as standard procedure for tissue sampling or injection of fluids. However, only limited development has taken place in their design for several decades. This has caused complications in some applications, for example, targeting a specific area for anesthesia or targeting tumor in biopsy. In breast biopsy using Fine Needle Aspiration (FNA) and up to 21% in thyroid. Moreover, approximately 10% of the population avoid medical treatment due to fear of needles, potentially influencing access to healthcare. Therefore, considerable improvements to medical needles are required.

In FNA, the insertion and guidance of the needle are facilitated by its back-and-forth translation, rotation around the center axis, and tilting. At a much faster rate than possibly produced by a human operator, an alternative way consists of generating ultrasonic mechanical waves, resulting in microscale motions of the needle tip.

Actuation of the needle tip by ultrasound has recently been demonstrated to be useful for many medical purposes: to decrease the penetration force for an easier and spatially more accurate procedure, provide improved visibility of the biopsy needle under ultrasound Doppler imaging, and hypothetically causing less pain by reducing the penetration resistance.

We have recently developed an ultrasound-enhanced fine needle aspiration biopsy (USFNAB) method, which allows the introduction of flexural ultrasound waves to a medical needle and needle tip to improve the tissue yield in biopsy. In USFNAB, a piezoelectric actuator is employed to produce kHz longitudinal waves, which are converted into flexural waves, essentially producing mechanical movement of the needle tip laterally with respect to the needle center axis. However, optimization in power-efficiency has not been yet presented. Clinically, a small and fully portable device would typically be considered preferable for better usability, the potential to batterize, and patient or operator safety, which could be enabled by improved electrical-to-acoustical power-conversion efficiency.

In this study, we aimed to develop a new acoustic waveguide construct with the intent to optimize the electrical-to-acoustical power-conversion efficiency and be capable of converting longitudinal to flexural mode, followed by validation of the technical performance.

II. MATERIAL AND METHODS

Using the finite element (FE) method, we chose to first model the Langevin transducer that was selected as the
sound source (Fig. 1). This was followed by stepwise optimization of the geometry of the full system comprising a piezoelectric transducer, a waveguide, and a biopsy needle. Based on the simulations, we then built custom-made prototypes, followed by characterizing their performance and comparing the performance with simulations.

Two waveguide constructs were studied: (i) a system with a transducer center axis placed perpendicular to that of the needle [Fig. 2(A)] and (ii) a system with a transducer center axis aligned at an offset and parallel to that of the needle [Fig. 2(B)]. The needle is a standard hypodermic needle with a lancet tip (21 gauge, i.e., 21 G: $\varnothing_{in} \times \varnothing_{out} = 0.514 \times 0.8$ mm, length 120 mm, model: 466564/3, 100 Sterican, B. Braun Medical, Inc., Melsungen, Germany). The purpose of the waveguide is to convert the longitudinal wave mode in the transducer to a flexural wave mode in the needle, as well as to match the mechanical impedance of the transducer to the mechanical impedance of the needle. The appropriate mass distribution, shape, and positioning of the structures in relation to electrical input are critical to achieving optimum performance of the system. Therefore, to understand the wave propagation and optimize the system, an acoustic simulation was conducted in silico employing a FE method (COMSOL Multiphysics 6.1., Inc., Burlington, MA). In the following, the steps realized during this optimization are elaborated.

A. Numerical simulations

For optimization of the arrangement, the FE method was used. First, consider the piezoelectric phenomenon,
which consists of creating electric potential over the material if the material is under stress (direct electric piezo effect) or the generation of mechanical strain when an electrical field is applied on the material (reverse piezo electric effect).\textsuperscript{30,31} In our model, the stress-charge constitutive relation has been used:\textsuperscript{32}

$$\begin{align*}
\mathbf{T} &= e\mathbf{E} - e'e\mathbf{E}, \\
\mathbf{D} &= e\mathbf{S} + \varepsilon_3\mathbf{E},
\end{align*}$$

(1)

where \(\mathbf{T}\) is the stress vector (Pa), \(\mathbf{S}\) is the strain vector (unitless), \(\mathbf{E}\) is the electrical field vector (V m\(^{-1}\)), \(\mathbf{D}\) is the vector of electric displacement field (C m\(^{-1}\)), \(e\) is the piezoelectric constant (C m\(^{-1}\)), \(e'\) is the transposed piezoelectric constant (C m\(^{-1}\)), and finally \(\varepsilon_3\) is the dielectric permittivity at constant strain (F m\(^{-1}\)).

To study the vibration throughout the acoustic part of the device, the elastic wave equation, obtained from Newton’s second law, is then solved in a frequency domain perturbation study:

$$-\rho \omega^2 \mathbf{u} = \nabla \cdot (FS)^T + F_V e^{i\phi},$$

(2)

where \(\rho\) is the density of the material (kg m\(^{-3}\)), \(\omega\) is the complex angular frequency (rad s\(^{-1}\)), \(\mathbf{u}\) is the displacement (m), and \(FS\) is the first Piola-Kirchhoff stress (N m\(^{-2}\)), which when transposed is the nominal stress. \(F_V\) is the volume force (N m\(^{-3}\)) and \(e^{i\phi}\) refers to the alternating electrical field with phase \(\phi\) applied to the piezo elements.

The simulations were conducted in two steps, a stationary study was performed to calculate the deformation of the structure induced by the bolt pretension over the piezo stack using the following equation [Eq. (3)]:

$$0 = \nabla \cdot \mathbf{S} + \mathbf{F}_V,$$

(3)

where \(\mathbf{S}\) is the second Piola-Kirchhoff stress (N m\(^{-2}\)), followed by solving the elastic wave equation, Eq. (2).

To avoid infinite vibration amplitude, when the system is in resonance, an artificial damping ratio, i.e., Rayleigh damping, was set to 0.001 (unitless). The different parts of the model were meshed using the swept technique for the piezo stack and the free tetrahedral technique for the rest of the geometry. The mesh size was implemented to include at least 20 elements per wavelength.

The parameters of the different materials used in the models are given in the Table I.

### B. Simulation of the Langevin transducer

The modelled Langevin sandwich transducer, based on a commercially available piezo stack [P-010.10H, Physik Instrumente (PI) GmbH & Co. KG, Karlsruhe, Germany], was composed of 22 ring-type piezo elements (PIC151; dimensions = \(\phi_{\text{out}} \times \phi_{\text{in}} \times \text{thickness} = 10 \times 5 \times 0.5 \text{ mm}\) stacked and fastened by a bolt to achieve a pretension force (3.1 kN).

Each element was arranged so that the direction of the electric fields through the piezo was opposite as compared to the adjacent elements. Electrically, the elements were all connected in parallel: In this way, the required driving amplitude voltage was decreased by a factor of 22 compared to using a single element having a thickness of the full stack. At each end of the stack, an extra inactivated piezo element (PIC151, dimensions = \(\phi_{\text{out}} \times \phi_{\text{in}} \times \text{thickness} = 10 \times 5 \times 1 \text{ mm}\) was added to electrically isolate the transducer from the rest of the system for safety. To simulate the glue used by the manufacturer to hold the stack together, two layers of epoxy were added (epoxy, dimensions = \(\phi \times H = 10 \times 0.2 \text{ mm}\) at each end of the stack. Two masses of brass, custom-built, were added on each side of the piezo stack connected with the bolt, as detailed in the following, to shift the resonance frequency of the piezo stack to the targeted resonance of the entire construct (approximately 30 kHz). The front mass of the transducer [Fig. 1(A.a)] was cylindrical (brass, dimensions = \(\phi \times H = 14 \times 10 \text{ mm}\)). The back mass [Figs. 1(A.b) and 1(A.c)] was shaped to facilitate the fixation of the stack inside the enclosure. An M4 screw [Fig. 1(A.d)] (brass, \(H = 35 \times 4 \text{ mm}\)) was connected to a cylindrical front mass (brass, \(H = 3 \times 10 \text{ mm}\)) and to a cube-shaped back mass [Fig. 1(A.e)] (brass, \(H = 5 \times 10 \times 12 \text{ mm}\)) connected to a rectangular cuboid [Fig. 1(A.f)] (brass, \(H = 5 \times 10 \times 12 \text{ mm}\)).

To fasten all the elements mentioned previously to create a stack, the M4 bolt linked the back mass and the top mass in order to apply the pretension force (3.1 kN) over the stack. The bolt connects both ends through the top mass via a thread and exposes 3 mm of the bolt, which was used to connect the Langevin transducer to the waveguide, using the waveguide’s threaded hole. Finally, after a pre-study calculating the bolt pretension effect over the system, a frequency domain with perturbation simulation from 20 to 40 kHz was conducted to find the resonance frequency of the system.

| Material                | Density [kg/m\(^3\)] | Young’s modulus [GPA] | Poisson’s ratio | Isotropy     | Reference     |
|-------------------------|-----------------------|-----------------------|-----------------|--------------|---------------|
| Brass                   | 8750                  | 115                   | 0.307           | Isotropic    | Ref. 33       |
| Epoxy                   | 1250                  | 3.5                   | 0.3             | Isotropic    | Ref. 34       |
| PIC151                  | 7760                  | Table II              | 0.34            | Anisotropic  | Physik Instrumente datasheet$^a$ |
| EOS Stainless steel 316L| 8000                  | 193                   | 0.265           | Isotropic    | Ref. 33       |
| AISI 304                | 8070                  | 205                   | 0.275           | Isotropic    | Ref. 35       |

$^a$Data provided upon request by Physik Instrumente.
C. Simulations of the waveguides

Connecting the needle directly to the transducer without impedance matching would lead to compromised transmission of acoustic power from the transducer to the needle. The transducer stack is resonating in a longitudinal mode with a relatively small amplitude and has mechanically a high impedance. On the other hand, the needle should vibrate with a large amplitude, approximately 200 μm perpendicularly to its center axis. The needle is flexurally relatively compliant, thereby having a low mechanical impedance. Therefore, a mechanical impedance matching from the transducer to the needle is needed, which was implemented with a resonator (Fig. 2.1) connected to a horn (Fig. 2.2). In addition, the connection point of the transducer to a resonator converts wave modes from a longitudinal to a flexural mode at the needle.

We investigated two waveguide geometries: a linear waveguide and a right-angled waveguide [Figs. 2(A) and 2(B)].

The linear waveguide was made of EOS Stainless steel 316 L (surgical quality) and included a rectangular resonator in the proximal end [dimensions = L × W × H = variable length (from 8 to 100 mm with 0.1 mm steps) × 12 × 3 mm] (Fig. 2.1). From here, the waveguide extrudes as a tapered section named horn, with the width (W) and the height (H) decreasing in width and height exponentially along the length towards the distal end (Fig. 2.2) [L = variable length (from 5 to 50 mm with 0.1 mm increments), dimensions starting point = W × H = 12 × 3 mm, dimensions ending point = W × H = 0.9 × 0.5 mm] (Fig. 2.2). The tapering amplifies the acoustic signal and facilitates mechanical impedance matching: The constant flow of power will propagate as a wave with increasing amplitude into more flexible regions at smaller cross-sections of the horn while minimizing reflections of energy. The waveguide also included a groove along the length of the waveguide to provide a coupling area for the needle cannula (dimensions = φin × φout = 0.514 × 0.8 mm).

The second design, right-angled waveguide, included a linear portion with a 90° angle implemented in the beginning of the horn [Fig. 2(B)]. The rationale for this geometry is to improve the ergonomics by aligning the center axes of the transducer and the needle to be parallel.

We conducted the waveguide optimization in two steps: First, the length of the resonator was varied and simultaneously the absolute displacement obtained at one end of the resonator was recorded (Fig. 2.1, red dots). The smallest length of the resonator with a local maximum displacement was selected as the horn length for the next step. Second, the length of the horn was varied, and simultaneously the absolute displacement obtained at the distal end of the horn was recorded. The length of the horn was selected by choosing a length that provided a local maximum in the displacement at the tip of the needle (Fig. 2.2, red dots), still being sufficiently short to enable usable operation (subjective evaluation). The frequency used in optimizing resonator and horn lengths was 27.4 kHz, which is the natural resonance frequency of the transducer obtained from the transducer simulation study (Fig. 1).

D. Simulation of the distal needle length

After optimization of the waveguide, the distal part of the needle was included in the simulation to understand its effect on the transducer and find an optimal position. The distance from the tip of the waveguide (horn part) to the needle tip was varied from 8 to 60 mm at 100 μm increments without load constraint at the tip. The simulated needle follows the specification of a standard 21 G hypodermic needle (dimensions = φin × φout = 0.514 × 0.8 mm) made of AISI 304. The needle tip, based on standard hypodermic needle tip, is composed of two bevels which have opening angles of 17° and 11°, respectively, observed as a side projection to yz plane (Fig. 2.3). Similar to the waveguide optimization, the needle length was varied and simultaneously the absolute displacement obtained at the needle tip was recorded (Fig. 2.3, red dots). The distal needle length was selected by choosing a length that provided a local maximum in the displacement at the tip of the needle, allowing at least 50 mm penetration depth inside tissue as measured from the site of insertion.

E. Impedance characteristics of the experimental system and simulation

After building the experimental systems, the agreement between the experimental system and the simulation was verified in terms of impedance characteristics as follows: The Langevin transducer was coupled to the waveguide, which was a three-dimensionally (3D) printed EOS stainless steel 316 L object (3D Formtech Oy, Jyvaskyla, Finland). The needle (21 G × 120 mm, model: 466564/3, 100 STERICAN, B Braun, Melsungen, Germany), placed in the groove, was brazed onto the last 20 mm of the waveguide’s distal end.

The simulated electrical impedance of the arrangement was obtained from COMSOL using a frequency domain with perturbation simulation (20–40 kHz with 50 Hz steps). The impedance measurement was conducted using a Digilent Analog Discovery 2 combined with the WaveForms software (Digilent, Inc., Henley Court Pullman, WA).

F. Quantifying the needle tip displacement

To observe the needle tip during sonication we used a high-speed camera (model: Phantom V1612, Vision Research, Inc., Wayne, NJ) in conjunction with a macro lens (model: Canon MP-E 65 mm f / 2.8 1–5x Macro Photo, Canon, Inc., Tokyo, Japan). The camera was placed in front of an acrylic chamber (external dimensions = L × W × H = 21 × 14 × 15 mm, wall thickness = 5 mm) filled with air or de-ionized water degassed to 6 mg/L at ambient temperature (22.3 °C). The needle was immersed to a depth of 30 mm from the water surface.

The system was driven with an incoming signal generated by a Digilent Analog Discovery 2 at the resonance frequency (f ≈ 30–34 kHz), which was defined by the maximum power transmission (difference of the incident
and reflected powers), for ten cycles. This was connected to a custom-made class B amplifier with an output impedance of 50 Ohms coupled with a custom-made step-up transformer (11:20 turn ratio) delivering an incoming instantaneous power of 5 W (schematics of the setup are shown in Fig. 3).

High-speed camera videos were taken in order to acquire the needle tip displacement during sonication [sample rate = 10 × the driving frequency (frames per second), exposure time = 0.8 μs, resolution 128 × 128 pixels, angular magnification = 5x, pixel size = 5.6 μm]. The videos were then analyzed in MATLAB (Release 2020b, The MathWorks, Inc., Natick, MA) using an algorithm capable of tracking the needle position using cross correlation frequency technique to allow subpixel image registration based on the work of Perra et al.19 Translating the needle of a range of 50 μm at 1 μm steps with a high-precision stage (Z825B stage, Thorlabs, Inc., Newton, NJ; stage precision: 29 nm), the true and optically measured displacement correlated with Pearson correlation coefficient of 0.9982 with a slope of 1.005. A total of ten repetitions per variation were realized.

G. Calorimetry

To define the total acoustic power emitted from the needle, a calorimetric assessment was done. The calorimeter (material = thermal wall insulator, wall thickness = 3 cm) was filled with 25 ml of de-ionized water (at room temperature: 22°C) mixed with a magnetic stirrer (C-MAG HS 4, IKA-Werke GmbH & Co. KG, Staufen, Germany). The needle tip was then immersed to a depth of 30 mm from the water surface.

The system was operating at resonance frequency (f ≈ 30.5–32.0 kHz), which was defined as the frequency providing the maximum power transmission (difference of the incident and reflected burst-averaged electrical powers) within the frequency range 30–34 kHz (calibration). The signal used during sonication was composed of bursts (succession of activated signals over a certain period of time followed by a non-activated signal) of 300 cycles with a 55% duty cycle during 30 s. The incident and reflected powers were measured and time-averaged over a burst period; The consumed electrical energy is equal to their difference.37 Different consumed electrical energy were tested, from 0 to 3 J at 0.3 J steps. The water temperature change was recorded using a PT 3000 thermometer probe for 60 s before and after the sonication. After averaging the temperature data, the difference of the two temperatures gave the heat deposited into a known volume of distilled water for a certain incident energy. The calculation of the heat deposition was obtained from the Eq. (4),

\[ P_{\text{heat}} = \frac{m \times c \times \Delta T}{t}, \]

where \( m \) is the water mass (g), \( c \) is the specific heat capacity of water [J/(g°C)], \( \Delta T \) the difference of the average temperature before and after sonication (°C), and \( t \) is the sonication time (s). The efficiency of the system was defined by the slope of the line obtained from a linear fit on the measurements of energy deposition as a function of consumed electrical energy.

III. RESULTS

A. Langevin transducer

The first analysis aimed to define the resonance frequency of the transducer. The computational results indicate a peak vibration amplitude around 27.4 kHz (vibration of 2.9 μm at 5 W input electrical power), as shown in Fig. 1.

B. Optimization of the resonators, horns, and needle length

The purpose of this simulation was to find a geometry of the waveguide, which would provide a great displacement in y direction while avoiding “spurious” displacement in other directions due to different modes. In Fig. 2(A.1), there is a clear recurrence of the greater displacement values at every multiple of a flexural wavelength (23 mm) in the resonator (stainless steel, thickness: 3 mm) at 27.4 kHz. For practical reasons, the length of the resonator was selected to be the shortest possible i.e., 23 mm. Maximum displacement at the tip of the horn [Fig. 2(A.2)], occurred at every multiple of half of the flexural wavelength (stainless steel). However, the wavelength was shorter because the wavelength depends on the structure thickness,38 which decreases exponentially along the horn (varying thickness: from 3 to 1 mm).

To minimize the reflection of the flexural wave in the tapering part of the horn, one should select a long horn; however, the longer the horn is, the more compromised the usability is (maneuvering by the operator), thus 39.5 mm was selected. The displacement is greatly
amplified by the tapered shape of the horn [Fig. 2(A.2), flexural displacement of 48.6 μm at 5 W electrical input power] compared to that of the resonator structure [Fig. 2(A.1), flexural displacement of 4.8 μm]. The needle displacement [Fig. 2(A.3)] is also greater than the horn construct (flexural displacement of 88 μm), this is due to the tapered distal end of the needle, which behaves as an amplifier. From the data set, it seems that every half flexural wavelength provides a local maximum displacement (λ ≈ 16.6 mm) in AISI 304 material (cross-sectional shape is a ring: \( \text{ø}_{\text{out}} = 0.8 \text{ mm} \), \( \text{ø}_{\text{in}} = 0.514 \text{ mm} \)). The selected distance distal part of the needle to distal part of the horn was selected to be 58 mm, which was selected for practical reasons considering the biopsy applications.

The results shown in Fig. 2(B), using the right-angled waveguide, suggest a similar functioning as the linear waveguide. Similar dimensions were selected as 23 mm for the resonator (generating flexural displacement of 4.5 μm), 39.5 mm for the horn (flexural displacement of 35 μm), and 58 mm for the distal length of the needle (flexural displacement of 117 μm).

C. Validation model—System

To compare the difference between the simulations and our prototypes, an impedance measurement with a bare stack transducer and the whole systems were compared with the respective simulated impedances in the frequency range 20–40 kHz (Fig. 4). The simulation curve and the measured peaks exhibit similar behavior. The main impedance peak of the transducer is at 30.6 kHz, while with the linear and right-angled waveguide system, the impedance peak is measured at 29.7 and 27.5 kHz, respectively.

D. Experimental arrangement

The different manufactured prototypes are demonstrated in Fig. 5. Comparison of the waveguide structures are shown in Figs. 5 and 6 demonstrating that both of the designs can create flexural waves at the needle tip. Simulations at 5 W input power showed a displacement peak of 88 μm at the needle tip for the linear waveguide [Fig. 5(A)] and 117 μm for the right-angled waveguide [Fig. 5(B)]. Simulations suggested similar displacements...
The measurements at 5 W in air [Figs. 6(A.1) and 6(B.1)] recorded 89 ± 2 μm (mean ± standard deviation, SD; n = 10) and 102 ± 1 μm (n = 10) for the linear and right-angled waveguide, respectively. In water, the displacements at the needle tip were up to 87 ± 6 μm (mean ± SD; n = 10) and 92 ± 1 μm (n = 10), for the linear and right-angled waveguide, respectively. The coefficient of variation percentage (CV%) for displacements were in the range 1.5%–6.5%, suggesting acceptable repeatability. Moreover, it seems that a slightly greater electrical-to-acoustical power-efficiency of 73% was obtained with the right-angled waveguide compared to 69% with the linear waveguide [Figs. 6(A.3) and 6(B.3)].

IV. DISCUSSION

The results of the in silico structural optimization demonstrated the displacements at the needle tip were significantly dependent on the dimensions of the waveguide design and needle-to-waveguide coupling point, the greatest displacements were obtained when the system was in resonance. In the simulations, an increase in displacement was observed at the end of each stage, for example, resonator, horn, or needle, respectively. Importantly, the experimentally demonstrated power-efficiency was good, 69% with the linear waveguide, 73% with the right-angled waveguide, which is in line with the literature.37,39 Several factors may contribute to the power-efficiency, as outlined in the following.

The extrusion of the linear rectangular segment of the resonator extends symmetrically from the transducer coupling point towards opposite directions. Symmetry minimizes the generation of bending modes within the transducer, which, if present, could reduce electro-acoustic efficiency of the transducer. Bending waves in the transducer would yield tension and compression in different parts of the piezo stack simultaneously, compromising the performance, which is now eliminated. Furthermore, the exponential tapered shape of the horn gives an amplification of the signal by having approximately the same energy in the smaller cross section, while minimizing the reflection from the boundaries towards the source. The geometry, therefore, provides geometric amplification at the same time with successful impedance matching from the higher mechanical
impedance of the transducer to the lower mechanical impedance of the needle. The needle bevel provides further an additional state of displacement amplification towards the tip of the needle, where the action of the ultrasound is intended.  

The studied waveguides, together with the transducer alignment, successfully converted the longitudinal waves of the transducer to a flexural displacement of the needle tip. The experiments with needle in air provided large peak-to-peak displacement amplitudes up to 200 and 180 \( \mu \text{m} \) in water. The practical relevance of the potential to generate large displacements is selecting the relevant magnitude needed for a specific application in question. For example, gentle actuation of tissue by the needle or adjacent micro-bubbles could be used to improve tissue yield in a biopsy application, whereas generation of large amplitudes and violent cavitation clouds could be useful in an intervention application such as the histotripsy of a tumor. Several reports have shown that the actuation of the needle tip by ultrasound might be also useful for medical purposes such as improving spatial accuracy of needle insertion and avoiding tissue deformation or hypothetically pain reduction.  

None of the studies mentioned before reported on the optimization of electro-acoustic efficiency. Better efficiency could contribute to lower operation voltages enabling better patient safety and could further enable, for example, batterization and miniaturization if developed to the full potential. The instantaneous electrical input power into the transducer was below 5 W and the displacements demonstrated were sufficient required for the biopsy application. Compared to the previous research device used by Perra et al., a hand-piece volume and a mass decrease of 70% and 34%, respectively, have been achieved with the new design. Feedback from medical doctors confirmed the clinically acceptable usability of the miniaturized ultrasonic devices. However, the right-angled waveguide construct was preferred over the linear waveguide for its “pen-like” ergonomics [Fig. 5(F)] resembling needle devices currently used clinically.  

The loading of the needle is expected to affect the resonance frequency; therefore, the system could in the future be improved with resonance frequency tracking, which was not implemented in this study and is a limitation. In our experiment, the effect of the water on the frequency change was not investigated, but stiffer and more viscous media could have a greater effect on the frequency, leading to a reduction in the efficiency.  

V. CONCLUSION  

Ultrasound in medicine is a well-known technology and is widely used for different kinds of therapeutic procedures. However, flexurally oscillating needles have received limited attention in the literature. Since the needle is a common tool in healthcare but with clear limitations, this paper showed a novel way of improving needles. By actuating the needle with flexural waves, we may be able to decrease the penetration force needed during the insertion of the needle or improve tissue extraction during biopsies. Because of the efficiency of the system, the design could lead to better patient safety of flexurally oscillating medical needles, and enable batterization and miniaturization of the system, which are key factors influencing user adaptation to new ultrasonic technologies in the medical setting. Therefore, this study opens insights towards developing flexurally oscillating ultrasound-actuated medical needles for a variety of medical purposes including biopsy, drug and gene delivery, and minimally invasive interventions.  

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APPENDIX  

See Table II for material data set.

| Coefficient | unit | Value    | Coefficient | unit | Value    |
|-------------|------|----------|-------------|------|----------|
| s11E        | m² N⁻¹ | 1.683 \times 10^{-11} | s11D       | m² N⁻¹ | 1.436 \times 10^{-11} |
| s33E        | m² N⁻¹ | 1.900 \times 10^{-11} | s33D       | m² N⁻¹ | 9.750 \times 10^{-11} |
| s55E        | m² N⁻¹ | 5.096 \times 10^{-11} | s55D       | m² N⁻¹ | 2.924 \times 10^{-11} |
| s12E        | m² N⁻¹ | -5.656 \times 10^{-12} | s12D       | m² N⁻¹ | -8.112 \times 10^{-12} |
| s13E        | m² N⁻¹ | -7.107 \times 10^{-12} | s13D       | m² N⁻¹ | -2.250 \times 10^{-12} |
| s44E        | m² N⁻¹ | 5.096 \times 10^{-11} | s44D       | m² N⁻¹ | 2.924 \times 10^{-11} |
| s11E        | m² N⁻¹ | 4.497 \times 10^{-11} | s11D       | m² N⁻¹ | 4.497 \times 10^{-11} |
| c11E        | N m⁻² | 1.076 \times 10^{-11} | c11D       | N m⁻² | 1.183 \times 10^{-11} |
| c33E        | N m⁻² | 1.004 \times 10^{-11} | c33D       | N m⁻² | 1.392 \times 10^{-11} |
| c55E        | N m⁻² | 1.962 \times 10^{-11} | c55D       | N m⁻² | 3.420 \times 10^{-10} |
| c12E        | N m⁻² | 6.312 \times 10^{-10} | c12D       | N m⁻² | 7.376 \times 10^{-10} |
| c13E        | N m⁻² | 6.385 \times 10^{-10} | c13D       | N m⁻² | 4.436 \times 10^{-10} |
| c44E        | N m⁻² | 1.962 \times 10^{-10} | c44D       | N m⁻² | 3.420 \times 10^{-10} |
| c66E        | N m⁻² | 2.224 \times 10^{-10} | c66D       | N m⁻² | 2.224 \times 10^{-10} |
