Observation of the Ultrasonic Vibration Potential with an Instrumented Coaxial Needle Probe

Conor McDermott Systems and Computer Engineering Carleton University Ottawa, Canada conormcdermott@cmail.carleton.ca Hossein Asilian Bidgoli Systems and Computer Engineering Carleton University Ottawa, Canada hosseinasilianbidgo@cmail.carleton.ca Carlos Rossa Systems and Computer Engineering Carleton University Ottawa, Canada rossa@sce.carleton.ca

Abstract—Stereotactic core biopsies typically use ultrasound imaging to locate tumours and guide a biopsy needle toward them. Due to the poor resolution of ultrasound images, oftentimes it is difficult to ensure that the needle reaches the target before the tissue is sampled resulting in false negatives. Sensor-instrumented needles provide an efficient way of characterizing the tissue at the needle tip to improve tumour targeting. A modality that has not been considered for real-time tissue characterization is the ultrasonic vibration potential (UVP) of the medium - a small electric signal generated within the tissue when subjected to ultrasonic pressure. The magnitude of the UVP depends on the electroacoustic properties of the tissue, providing valuable information about the relative tissue composition.

This paper investigates, for the first time, the feasibility of measuring the ultrasonic vibration potential (UVP) of assorted media using an instrumented biopsy needle. The results show that the UVP can be effectively measured using the proposed instrumented needle providing electroacoustic data that can be used in future work for real-time tissue classification during stereotactic procedures.

Index Terms-component, formatting, style, styling, insert

I. INTRODUCTION

Needle-based stereotactic surgeries aim to perform localised, minimally invasive surgery (MIS) procedures on a targetted area inside a patient's body. Operations such as prostate brachytherapy, and percutaneous nephrolithotomy limit the size of incisions, thereby decreasing the recovery period with less tissue trauma, postoperative pain, and minimised scarring [1], [2].

Oftentimes, thanks to its portability, low cost, and lack of ionizing radiation, ultrasound (US) is used as the imaging modality to guide a needle to the point of interest within the body. However, US has limited resolution and only a slice of the needle or its profile is visible at a given time. As a result, either the depth of puncture is difficult to gauge or the needle's profile may be lost entirely as the sonographer moves the probe [3]. To improve needle localisation, the

Cette recherche a été financée par le Conseil de recherches en sciences naturelles et en génie du Canada (CRSNG), par les Instituts de recherche en santé du Canada (IRSC), et par le Conseil de recherches en sciences humaines du Canada (CRSH), [numéro de référence NFRFE-2018-01986]

needle's tips may be instrumented to inform a clinician about the composition of the tissue it is inserted in via real-time feedback, even when the needle or the target is not fully visible. In literature, this has been done using electrodes that enable real-time electrical impedance spectroscopy (EIS) or optical spectroscopy (OS) [4]–[6].

EIS is a modality used to analyze the electric properties of different materials and discriminate biological tissues in a variety of applications [4], [7]–[9]. EIS measures the tissue's electrical impedance when signals of different frequencies are injected into it [10], [11]. It has been shown that electrodes installed in biopsy needles can lead to highly localised impedance sensitivity and aid in prostate cancer tumour localisation and biopsy [4], [12]. Hypodermic needles can be coated in low-cost microeletromechanical systems to be used as a disposable impedance sensing electrode for breast cancer characterisation [13].

OS analyzes the optical properties of tissue over a broad spectrum of injected light following the photoelectric effect [14]. The injected light is transmitted and received using fibre optic cables, which can be easily integrated into surgical needles [15]. The measured absorption, reflection, and emission of light are then used to classify the tissue based on known light spectra [16]. OS has been used to detect brain tumours by identifying blood vessels characteristic of malignancy using a single injected light wavelength [17]. It has also been shown that biopsy needles sensored for OS can detect breast cancer [18].

Some pathological processes alter multiple characteristics of the tissue, such as its electrical and mechanical impedances. Multimodal instrumentation for tissue classification can be used to exploit the cumulative properties that no singular modality can achieve on its own. An example is the stereotactic probe proposed in [19], which is equipped with a strain gauge to measure mechanical force, a neuroendoscope to image brain tissue, and an optical scattering spectroscope to discriminate cancer tissue. Combined with a neural network, these sensors capture multimodal information about the tissue properties to increase confidence in cancer tissue classification.

A modality that has not been explored for tissue characterisation is the ultrasonic vibrational potential (UVP). A US wave passing through a tissue creates a small oscillating

We acknowledge the support of the Natural Sciences and Engineering Research Council of Canada (NSERC), the Canadian Institutes of Health Research (CIHR), and the Social Sciences and Humanities Research Council of Canada (SSHRC), [funding reference number NFRFE-2018-01986].

electrical potential whose frequency matches that of the injected mechanical wave. The magnitude of the UVP signal depends on the concentration of the suspension, temperature [20], the velocity gradient within the US waves, and the electrolyte's properties, while being nearly independent of the concentration of electrolyte in a uni-univalent solution [21]. Thus, the observation of the UVP effect can provide additional information about the tissue's ion masses, dielectric values, and concentrations.

The UVP is subdivided into two smaller signals: the ionic vibrational potential (IVP), when the acoustic wave passes through an ionic medium, and the colloidal vibrational potential (CVP), when the acoustic wave passes through a colloidal suspension. The magnitude of the CVP is typically larger than that of an equivalent IVP. In the IVP, acoustophoretic motion of the particles within the tissue generates alternating dipoles which in turn generate a macroscopic voltage potential [22].

The UVP was originally used as a tool for determining the mass of electrolytic ions in [23]. The original UVP model was incomplete as it only considered the force upon an ion from an oscillating electric field and the force from the viscous friction imparted on the ion by the solvent. The model was later expanded to include the relaxation force, the electrophoretic force, diffusion force, and pressure gradient force [24]–[26]. Between the 1970's and 1990's, interest in the effect concentrated on applications for aiding in industrial quality control [27]. More recently, the effect has been considered as an imaging modality with electrouacoustics as a contrast mechanism [28]–[31].

An attractive feature of the UVP is that it provides large contrasts, on a scale of 500 [32], in the measured magnitude of the effect between blood and muscle tissue. Radiographic, photoacoustic, optical, and US imaging alone do not typically have such high contrast. For these reasons, there is potential for the UVP to be used during US-guided stereotactic procedures for tissue characterization, or in conjunction with other classification modalities.

The first step towards using the UVP for tissue classification in stereotactic surgery is to develop an instrumented needle capable of measuring the effect, something that has not been attempted before in literature. The UVP is typically measured using a pair of electrodes placed some distance apart inside the tissue. Measuring the IVP with an instrumented needle where the electrodes are only separated by less than a millimetre represents an open technical challenge. Ideally, the electrode is bio-compatible with a small profile to suit stereotactic applications. After a suitable sensored needle electrode is created, measuring the UVP under various conditions is the first step toward predicting the value of the UVP for future tissue characterization. In this paper, measurement of the UVP using a sensored surgical needle electrode is presented for the first time.

The paper is structured as follows: Section II explains the conceptual background of the effect, which serves as a guide for the design of the sensored needle shown later in Section III, where the methods for detecting the effect in different media



Fig. 1. An illustration of the microscopic effects that leads to the IVP effect. Arrows indicate displacements of individual ions along a US wave propagation path. The lengths of the arrows indicate the relative displacement of the ions, with the displacement of cations assumed to be smaller. Region B will be charged positively compared to region A. If a set of electrodes is inserted in regions A and B, a cyclic voltage potential is measured, which corresponds to the IVP effect.

is presented. Section IV shows the experimental results, and Section V discusses the results and concluding remarks.

II. IVP MANIFESTATION

Fig. 1 shows the mechanism of the generation of the IVP [20]. Consider an US wave propagating through a medium charged with ions or colloids. As the mechanical wave passes through the medium, its individual particles are disturbed from their equilibrium positions. Charged particles with opposite charges tend to have slightly different masses and frictional coefficients and therefore experience different magnitudes of acceleration. This leads to a periodic excess of one charge or the other in a given volume of the media. At any instant, the particles are momentarily brought closer or further apart from one another, as the medium experiences compression and rarefaction, leading to a net difference in the local charge of the media at the location of the particle's movement. These microscopic dipoles, in summation, generate a macroscopic potential.

There are 4 components that are affected by the periodic oscillation induced by the US wave: the solvents velocity, the ions velocity, the electric field in the solution, and the concentration of the solution. The periodic change of the solvents velocity, v_0 , is given by

$$v_0 = v'_0 e^{i(\omega t - \sigma x)},\tag{1}$$

where v'_0 is the equilibrium solvent velocity, t is time, σ relates angular and linear velocity of the oscillatory pressure field, xis the distance along the propagation path, and $e^{i(\omega t - \sigma t)}$ shows the oscillation of a given quantity from equilibrium due to the mechanical wave. The periodic change of the j^{th} ion's velocity is

$$v_j = v'_j e^{i(\omega t - \sigma x)},\tag{2}$$

where v'_j is the equilibrium ion velocity. The periodic change of the electric field E generated by the aforementioned ion motion is

$$E = E'e^{i(\omega t - \sigma x)} \tag{3}$$

where E' is the equilibrium electric field. Finally, the periodic change of the concentration of the j^{th} ion in a given volume is

$$\nu_j = \nu'_j e^{i(\omega t - \sigma x)},\tag{4}$$

where ν'_{j} is the equilibrium ion concentration. These assumptions are subject to the force balance of all ions, that is

$$\sum_{j} F_j = m_j \frac{dv_j}{dt},\tag{5}$$

where F_j is the sum of forces on the j^{th} ion, and m_j is the mass of the j^{th} ion. Further, the concentration of ions within the suspension must remain constant and may only change in a given volume by the motion of ions already in the suspension. Thus

$$\frac{\partial n_j}{\partial t} + \frac{\partial (n_j v_j)}{\partial x} = 0, \tag{6}$$

where n_j is the concentration of ions per m³ of the j^{th} ion. From (5) and (6), there are j equations to account for j ions, but not enough to account for E, the electric field. Poisson's equation is applied to the model to account for this additional term and gives

$$\frac{\partial E}{\partial x} = \frac{4\pi}{D} \sum_{j} n_j e_j,\tag{7}$$

where D is the overall dielectric constant of the medium and e_j is the charge of the j^{th} ion. Lastly, the generated electric field E and the observed voltage potential Φ are related by

$$E = -\frac{\partial \Phi}{\partial x} \tag{8}$$

This relation with the above three equations form the basis of the model in [26].

An equation for the voltage potential in a suspension is formed when (8) is rearranged to be in terms of Φ and (5),



Fig. 2. a: Sensored electrode in introducer needle. b: Sensored electrode with macro shots of needle end and needle tip showing coaxial configuration.

(6), and (7) are substituted in, then integrated with respect to the US propagation path, x, to give

$$\Phi = \frac{v_{US}v_0'4\pi}{\omega D} \cdot e^{i(\omega t - \sigma x - \Delta)} \cdot \frac{\sum_j \frac{\bar{n}_j e_j m_j}{\rho_j} - \sum_j \frac{\bar{n}_j e_j V_j}{\rho_j} (\frac{kT\sigma^2}{\omega^2} + s_0)}{\frac{4\pi\lambda_\omega}{\omega D} + \frac{iD\omega}{D}} \quad (9)$$

where v_{US} is the velocity of the US wave in the medium, \bar{n}_j is the concentration of the j^{th} ion in a given volume, ρ_j the frictional coefficient between the j^{th} ion and the solvent, k is Boltzmann's constant, T the absolute temperature, s_0 the solvent density, V_j the volume of the j^{th} ion, λ_{ω} the specific conductance at angular frequency ω , and D_{ω} the dielectric constant of the solution at angular frequency ω . There will be 2j+1 equations to describe an ionic system of j ionic species.

From the above, an important conclusion follows: the magnitude of the measured effect depends on the separation distance between the measurement electrodes, where the measured magnitude of the effect will oscillate between zero and twice the maximum calculated value from (9) as the electrodes are separated between an integral number of wavelengths and half-integral numbers of wavelengths, respectively [20], [33], [34].

The measurement of the UVP is typically done between two electrodes in a medium with generally known electroacoustic properties. Measuring the magnitude of the UVP with a sensored needle has not been shown experimentally. There is a challenge to be solved wherein electrodes of a shape and size compatible with conventional medical needles are capable of quantifying the magnitude of the UVP in various media.

III. EXPERIMENTAL SETUP

Figure 2a shows a photo of the sensored needle. The needle is composed of a coaxial cable electrode (Mouser, model 095-902-462-009) inserted in a rigid surgical introducer needle (Argon Medical Devices, model MCXS1815LX). The coaxial cable has an inner diameter of 0.30 mm and an outer diameter of 1.16 mm, see the right side of Figure 2b. Seen in Figure 2b are the woven shielding of the coaxial cable acts as the outer electrode, and the inner cable as the inner electrode. The electrodes have a separation distance of approximately 0.35 mm.

The sensored needle electrode is connected to a bandpass filter (1 MHz - 3 MHz) and the output of the filter is amplified by 43 dB. The output of the amplifier is sent to an oscilloscope (Rigol, model DS1054Z) and the waveform is captured in realtime by coupling the trigger threshold of the oscilloscope to the trigger waveform of the function generator, and moving the time scale of the display window to when the effect is expected to appear.

The needle electrode and introducer needle assembly are inserted near the base of a water-tight measurement cell. The measurement cell is made of a non-conducting photo polymer to disallow spurious potentials from leaking into the



Fig. 3. a: Diagram of experimental setup described in Section III. b: Picture of experimental setup described in Section III.

test media. At the top of the measurement cell is placed an US transducer (KB-Aerotech, model 171383H), shown in Figure 3b, with a diameter of 19mm produces US pulses at 1.6MHz. The US transducer is excited using a high voltage pulser (JSR Ultrasonics, model DPR300 pulser/receiver). The pulser is triggered by a function generator (Rigol, model DG4062) at a repetition rate of 1000 Hz. To control the height of the US transducer relative to the electrodes, a 3-axis stage with micrometer dials for each axis is used.

Figure 4a and 4b show how inclusions of different media are placed in the measurement cell. Depending on the experiment being conducted (e.g., saline solution or solid colloid inclusions) a set of supports is placed in the base of the measurement cell to hold the inclusion at a constant height. This allows the needle electrode to be inserted in approximately the same location between experiments. To acoustically couple the US transducer to solid inclusions, distilled water is used, seen in Figure 4b.

A series of experiments to observe the magnitude and dependencies of the IVP and ensure the signal is not noise were conducted. Different media consisting of pure single salt ionic solutions, and gelatin or agar inclusions (25 by 25 by 25 mm) were placed in the test cell. In the experiments, the measurement cell was either filled with saline or had an inclusion placed in it, then the needle electrode was placed inside. Solid inclusions were prepared by taking agar or gelatin powder and heating it in a container of water, then allowing it to set.

IV. RESULTS

To ensure that the observed signals are not electric noise or electromagnetic interference, the following experiments are conducted:

- UVP presence and UVP temporal relationship: Saline is poured into the measurement cell with the pulser on, the 3-axis stage is lowered in the Z axis until the US transducer is no longer in contact with the saline. As the distance between the transducer and electrodes increases, a temporal change in the signal related by t = d/v, where t is the measured temporal shift, d is the distance between the US probe face and the needle electrode, and v is the speed of sound within the medium, is noted and the magnitude decreases slightly. Upon losing physical contact, the UVP disappears.
- Spatial dependence of UVP: with the transducer contacting the saline surface again, the probes position is adjusted in the XY plane. As the far-field focal point is moved further from the measurement electrode's location, the magnitude of the measured effect decreases while staying in phase.
- Determination of whether the measured signal is the UVP or spurious noise: with the transducer in the position that maximises the magnitude of the effect, the transducer is



Fig. 4. a: Measurement cell set for measuring saline. b: Modified measurement cell with stool for measuring solid inclusions.



Fig. 5. Top plot illustrates the trigger frequency and approximate areas of interest. a): Diagram of trigger waveform and approximate temporal location of key features (i.e.: UVP and pulse's echo) b): UVP waveform in 1% saline solution. c): UVP waveform in 1% saline agar. d): UVP waveform in 1% saline gelatin. e): Pulse echo response measured by US transducer. The highlighted regions in b), c), and d) are the regions in which the UVP is present.

disconnected from the pulser (in effect turning the probe off) and thus causing the effect to disappear.

With the UVP's presence determined, a series of experiments are conducted to measure its magnitude in different media. The used media are: 1) 0 - 3% saline solution, 2) agar with 0 - 3% saline concentration, and gelatin with 0 - 3% saline concentration. Saline is chosen as the independent variable as the UVP is dependent on the electroacoustic properties of the medium.

- Figure 5b: IVP waveform in 1% saline solution. The waveform has a peak value of 85 mV and a duration of approximately 5 μ s. The profile of the waveform is similar to that of the US pulse seen in Figure 5e
- Figure 5c: UVP waveform in 1% agar inclusion, see Figure 4b. The waveform has a peak value of 70 mV and a duration of approximately 5 μ s.
- Figure 5d: UVP waveform in 1% gelatin inclusion, as per the setup shown in Figure 4b. The waveform has a peak value of 45 mV and a duration of approximately 5 μs.
- Figure 5e: 1.6 MHz US pulse echo. The first echo has a



Fig. 6. Maximum magnitude of the UVP in assorted media as a function of saline concentration.

duration of approximately 5 μ s, and the following echoes which are likely ringing effects from the excitation of the US probe.

The needle is placed in approximately the same location for each experiment, however, the inclusion experiments (Figures 5c-d) have the UVP generated within them slightly after the saline (Figure 5b), explained by a difference in the speed of sound within the media. Figure 5 shows the measured IVP waveform in 1% saline, in a 1% saline gelatin phantom, and in 1% saline agar. It can be seen that the magnitude of the effect changes as it translates from one media to another. Figure 6 shows the maximum measured magnitude of the effect given different media and saline concentrations. The magnitude of the the UVP in agar tends to be greater, and the concentration dependant relation is reflected in literature.

V. DISCUSSION AND CONCLUSION

The UVP depends on both the acoustic and electroacoustic properties of the media. With conventional US-guided procedures, tissue identification is difficult as US images have limited resolution and many conditions appear as isoechoic in US images. US imaging depends on acoustic contrast which has low resolution and does not provide detailed information on the structure of the target tissue. By sensorizing a needle, UVP magnitude measurements can be exploited to provide additional information on the tissue, which is particularly useful for malignant tumour detection, as tumours are often highly vascularised [22], [31]. The contrast in the signal's magnitude, shown in Figure 5a-c as well as Figure 6, may be useful as an aide in US guided MIS to provide better information on inclusion boundaries.

The results show that the UVP in assorted media and saline concentrations are quantifiable with the novel sensorized needle. The observed order of magnitude is similar to those previously reported in previous literature [20], [35]. As the saline content increase, the magnitude of the effect increases to a maximum, and then tapers off, which is consistent with literature. The results indicate that the sensorized needle is capable of quantifying the effect in a given medium. Since the experiments are conducted at a single US frequency, individual results do not contain sufficient information to determine what the composition of the medium. The response of the tissue to a multi-frequency waveform can be combined with methods, such as interface relationships, zeta potentials, or TOF relationships, to develop a robust classification method based on multi-modal spectral data.

REFERENCES

- P. Puangmali, K. Althoefer, L. D. Seneviratne, D. Murphy, and P. Dasgupta, "State-of-the-art in force and tactile sensing for minimally invasive surgery," *IEEE Sensors Journal*, vol. 8, no. 4, pp. 371–381, 2008.
- [2] M. Li, H. Liu, A. Jiang, L. D. Seneviratne, P. Dasgupta, K. Althoefer, and H. Wurdemann, "Intra-operative tumour localisation in robot-assisted minimally invasive surgery: A review," *Proceedings of the Institution* of Mechanical Engineers, Part H: Journal of Engineering in Medicine, vol. 228, no. 5, pp. 509–522, 2014.
- [3] R. J. Hallifax, J. P. Corcoran, A. Ahmed, M. Nagendran, H. Rostom, N. Hassan, M. Maruthappu, I. Psallidas, A. Manuel, F. V. Gleeson *et al.*, "Physician-based ultrasound-guided biopsy for diagnosing pleural disease," *Chest*, vol. 146, no. 4, pp. 1001–1006, 2014.
- [4] V. Mishra, A. Schned, A. Hartov, J. Heaney, J. Seigne, and R. Halter, "Electrical property sensing biopsy needle for prostate cancer detection," *The Prostate*, vol. 73, no. 15, pp. 1603–1613, 2013.
- [5] U. Dinish, F. Beffara, G. Humbert, J.-L. Auguste, and M. Olivo, "Surface-enhanced raman scattering-active photonic crystal fiber probe: Towards next generation liquid biopsy sensor with ultra high sensitivity," *Journal of Biophotonics*, vol. 12, no. 11, p. e201900027, 2019.
- [6] J. Sierra, J. Marrugo-Ramírez, R. Rodriguez-Trujillo, M. Mir, and J. Samitier, "Sensor-integrated microfluidic approaches for liquid biopsies applications in early detection of cancer," *Sensors*, vol. 20, no. 5, p. 1317, 2020.
- [7] R. J. Halter, A. Schned, J. Heaney, A. Hartov, and K. D. Paulsen, "Electrical properties of prostatic tissues: I. single frequency admittivity properties," *the Journal of Urology*, vol. 182, no. 4, pp. 1600–1607, 2009.
- [8] J. Jossinet, "The impedivity of freshly excised human breast tissue," *Physiological measurement*, vol. 19, no. 1, p. 61, 1998.
- [9] B. Kent and C. Rossa, "Electric impedance spectroscopy feature extraction for tissue classification with electrode embedded surgical needles through a modified forward stepwise method," *Computers in Biology* and Medicine, vol. 135, p. 104522, 2021.
- [10] F. Kremer and A. Schönhals, *Broadband dielectric spectroscopy*. Springer Science & Business Media, 2002.
- [11] L. A. Buscaglia, O. N. Oliveira, and J. P. Carmo, "Roadmap for electrical impedance spectroscopy for sensing: a tutorial," *IEEE Sensors Journal*, vol. 21, no. 20, pp. 22246–22257, 2021.
- [12] B. Kent and C. Rossa, "Development of a tissue discrimination electrode embedded surgical needle using vibro-tactile feedback derived from electric impedance spectroscopy," *Medical & Biological Engineering & Computing*, vol. 60, no. 1, pp. 19–31, 2022.
- [13] J. Yun, G. Kang, Y. Park, H. W. Kim, J.-J. Cha, and J.-H. Lee, "Electrochemical impedance spectroscopy with interdigitated electrodes at the end of hypodermic needle for depth profiling of biotissues," *Sensors and Actuators B: Chemical*, vol. 237, pp. 984–991, 2016.
- [14] G. R. Rossman, "Optical spectroscopy," *Reviews in Mineralogy and Geochemistry*, vol. 78, no. 1, pp. 371–398, 2014.

- [15] R. H. Wilson, K. P. Nadeau, F. B. Jaworski, B. J. Tromberg, and A. J. Durkin, "Review of short-wave infrared spectroscopy and imaging methods for biological tissue characterization," *Journal of biomedical optics*, vol. 20, no. 3, p. 030901, 2015.
- [16] W. W. Parson, Modern optical spectroscopy. Springer, 2007, vol. 2.
- [17] N. Markwardt, A. von Berg, S. Fiedler, M. Goetz, N. Haj-Hosseini, C. Polzer, H. Stepp, P. Zelenkov, and A. Rühm, "Optical spectroscopy for stereotactic biopsy of brain tumors," in *European Conference on Biomedical Optics*. Optica Publishing Group, 2015, p. 954208.
- [18] B. Yu, E. S. Burnside, G. A. Sisney, J. M. Harter, C. Zhu, A.-H. Dhalla, and N. Ramanujam, "Feasibility of near-infrared diffuse optical spectroscopy on patients undergoing image-guided core-needle biopsy," *Optics express*, vol. 15, no. 12, pp. 7335–7350, 2007.
- [19] R. Andrews, R. Mah, A. Aghevli, K. Freitas, A. Galvagni, M. Guerrero, R. Papsin, C. Reed, and D. Stassinopoulos, "Multimodality stereotactic brain tissue identification: the nasa smart probe project," *Stereotactic and functional neurosurgery*, vol. 73, no. 1-4, pp. 1–8, 1999.
- [20] R. Zana and E. B. Yeager, "Ultrasonic vibration potentials," in *Modern aspects of electrochemistry*. Springer, 1982, pp. 1–60.
- [21] E. Yeager, J. Bugosh, F. Hovorka, and J. McCarthy, "The application of ultrasonic waves to the study of electrolytic solutions ii. the detection of the debye effect," *The Journal of Chemical Physics*, vol. 17, no. 4, pp. 411–415, 1949.
- [22] V. A. Hackley, *The use of nomenclature in dispersion science and technology*. US Department of Commerce, Technology Administration, National Institute of ..., 2001, vol. 960, no. 3.
- [23] P. Debye, "A method for the determination of the mass of electrolytic ions," *The Journal of chemical physics*, vol. 1, no. 1, pp. 13–16, 1933.
- [24] S. Oka, Proceedings of the Physio-Mathematical Society of Japan.
- [25] J. Hermans, "Xxxv. charged colloid particles in an ultrasonic field," *The London, Edinburgh, and Dublin Philosophical Magazine and Journal of Science*, vol. 25, no. 168, pp. 426–438, 1938.
- [26] J. Bugosh, E. Yeager, and F. Hovorka, "The application of ultrasonic waves to the study of electrolytic solutions i. a modification of debye's equation for the determination of the masses of electrolytic ions by means of ultrasonic waves," *The Journal of Chemical Physics*, vol. 15, no. 8, pp. 592–597, 1947.
- [27] R. Pal, "Techniques for measuring the composition (oil and water content) of emulsions—a state of the art review," *Colloids and Surfaces A: Physicochemical and Engineering Aspects*, vol. 84, no. 2-3, pp. 141– 193, 1994.
- [28] V. E. Gusev and G. J. Diebold, "Imaging with the ultrasonic vibration potential: A theory for current generation," *Ultrasound in medicine & biology*, vol. 31, no. 2, pp. 273–278, 2005.
- [29] S. Wang, A. C. Beveridge, S. Li, G. J. Diebold, and C. K. Nguyen, "Frequency domain vibration potential imaging: Objects with symmetry in one dimension," *Applied physics letters*, vol. 89, no. 24, p. 243902, 2006.
- [30] P. Guan, M. Wang, H. I. Schlaberg, J. I. Khan, and V. Speirs, "Towards ascan imaging via ultrasonic vibration potential measurements," *Nuclear engineering and design*, vol. 241, no. 6, pp. 1994–1997, 2011.
- [31] A. C. Beveridge, S. Wang, and G. J. Diebold, "Imaging based on the ultrasonic vibration potential," *Applied physics letters*, vol. 85, no. 22, pp. 5466–5468, 2004.
- [32] A. Beveridge, S. Wang, V. Gusev, and G. Diebold, "The theory of ultrasonic vibration potential imaging," in *Journal de Physique IV* (*Proceedings*), vol. 125. EDP sciences, 2005, pp. 69–72.
- [33] N. Tseng and B. J. Roth, "The potential induced in anisotropic tissue by the ultrasonically-induced lorentz force," *Medical & biological engineering & computing*, vol. 46, no. 2, pp. 195–197, 2008.
- [34] A. C. Beveridge, S. Wang, V. Gusev, and G. J. Diebold, "Vibration potential imaging: theory and preliminary results," in *Photons Plus Ultrasound: Imaging and Sensing*, vol. 5320. SPIE, 2004, pp. 95–100.
- [35] A. Hunter and T. Jones, "The debye effect in electrolytes and colloids," *Proceedings of the Physical Society (1958-1967)*, vol. 80, no. 3, p. 795, 1962.