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Evaluating the Feasibility of Acoustic Radiation Force Impulse Shear Wave Elasticity Imaging of the Uterine Cervix With an Intracavity Array: A Simulation Study

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Abstract

The uterine cervix softens, shortens, and dilates throughout pregnancy in response to progressive disorganization of its layered collagen microstructure. This process is an essential part of normal pregnancy, but premature changes are associated with preterm birth. Clinically, there are no reliable noninvasive methods to objectively measure cervical softening or assess cervical microstructure. The goal of these preliminary studies was to evaluate the feasibility of using an intracavity ultrasound array to generate acoustic radiation force impulse (ARFI) excitations in the uterine cervix through simulation, and to optimize the acoustic radiation force (ARF) excitation for shear wave elasticity imaging (SWEI) of the tissue stiffness. The cervix is a unique soft tissue target for SWEI because it has significantly greater acoustic attenuation ($\alpha = 1.3$ to 2.0 dB·cm⁻¹·MHz⁻¹) than other soft tissues, and the pathology being studied tends to lead to an increase in tissue compliance, with healthy cervix being relatively stiff compared with other soft tissues ($E \approx 25$ kPa). Additionally, the cervix can only be accessed *in vivo* using a transvaginal or catheter-based array, which places additional constraints on the excitation focal characteristics that can be used during SWEI. Finite element method (FEM) models of SWEI show that larger-aperture, catheter-based arrays can utilize excitation frequencies up to 7 MHz to generate adequate focal gain up to focal depths 10 to 15 mm deep, with higher frequencies suffering from excessive amounts of near-field acoustic attenuation. Using full-aperture excitations can yield ~40%

increases in ARFI-induced displacements, but also restricts the depth of field of the excitation to ~0.5 mm, compared with 2 to 6 mm, which limits the range that can be used for shear wave characterization of the tissue. The center-frequency content of the shear wave particle velocity profiles ranges from 1.5 to 2.5 kHz, depending on the focal configuration and the stiffness of the material being imaged. Overall, SWEI is possible using catheter-based imaging arrays to generate adequate displacements in cervical tissue for shear wave imaging, although specific considerations must be made when optimizing these arrays for this shear wave imaging application.

I. Introduction and Background

A. Remodeling and Softening of the Uterine Cervix During Pregnancy

Preterm birth affects 13 million babies every year, and more than 35% of neonatal deaths are attributable to this cause [1], [2]. Premature babies who survive the perinatal period remain at lifelong risk for serious complications such as cerebral palsy, respiratory morbidity, mental retardation, blindness, deafness, cardiovascular disease, and cancer [3].

Spontaneous delivery, term or preterm, is accompanied by progressive remodeling of the cervical collagen microstructure in four phases: 1) softening, 2) shortening and further softening, 3) active dilation, and 4) post-delivery recovery [4]–[6]. The remarkably heterogeneous cervical microstructure consists of 3 layers of aligned collagen bands of differing orientations [7]–[14]. Data from animal studies suggest that these initially distinct layers become relatively indistinct by the time of delivery, and that a central circumferential layer undergoes more pronounced collagen disorganization than its flanking longitudinal layers [8], [15]–[18].

Biopsies from human cervixes demonstrate remodeling-associated microstructural and histological changes [19], [20], but invasive study is impractical *in vivo*. Unfortunately, noninvasive methods to evaluate cervical microstructure, such as light-induced fluorescence [21], electrical impedance [22], and ultrasonic acoustic attenuation [23], [24], have shown limited success. The inability to objectively describe and characterize microstructural rearrangement limits meaningful study of cervical remodeling, which in turn limits targeted exploration of associated molecular processes. Ultimately, this makes it difficult to conceive of novel approaches to predict and prevent preterm birth.

Biomarkers of cervical softening may be of particular value because softening begins soon after conception and continues throughout gestation. Further, it is experimentally clear that a soft cervix early in pregnancy increases the risk of preterm birth. However, what defines “too soft” is obscure because assessment of softening is subjective; the cervix is described as soft, medium, or firm based exclusively on digital examination. Because this is clearly inadequate, noninvasive ultrasonic techniques to describe softening, such as elastography, are an active area of research [25]–[28]. Recent studies, however, demonstrated that freehand elastography may be too user-dependent to be a reliable clinical tool. This has motivated the development of an elasticity imaging modality that is less dependent on user interaction. Acoustic radiation force-based methods are promising for this clinical imaging need.

B. Acoustic Radiation Force Impulse (ARFI) Shear Wave Elasticity Imaging (SWEI)

Acoustic radiation force-based elasticity imaging has recently been developed for investigation of mechanical properties of human soft tissues, including liver, arteries, heart, prostate, and breast [29]–[37], and it provides a unique tool to use to characterize the cervix during pregnancy. The acoustic radiation force $\vec{F}$ generated in soft tissues by focused ultrasound can be described by [38], [39]

$$\vec{F} = \frac{2\alpha \vec{I}}{c}, \quad (1)$$

where α is the acoustic attenuation coefficient of the tissue, c is the tissue's sound speed (assumed to be 1540 m/s in these studies), and $\vec{I}$ is the acoustic intensity at a given point in space. The acoustic radiation force is generated by a transfer of momentum from the propagating acoustic wave to the propagation medium through attenuation mechanisms such as absorption and scattering of the ultrasonic wave.

Focusing the acoustic radiation force allows for shear waves to be generated directly in tissues of interest, and the shape and size of these shear waves is dictated by both the acoustic excitation parameters, such as f-number and frequency, and the acoustic properties of the tissue, such as the acoustic attenuation [40], [41]. In linear, isotropic, elastic solids, the speed c_T of shear wave propagation is related to shear modulus (μ) and Young's modulus (E) by

$$c_T = \sqrt{\frac{\mu}{\rho}} = \sqrt{\frac{E}{2(1+\nu)\rho}}, \quad (2)$$

where ρ is the mass density and ν is Poisson's ratio. Thus, by measuring the speed of shear wave propagation in tissue, one can infer a quantitative estimate of the tissue stiffness. For incompressible solids, Poisson's ratio is 0.5 and $E = 3\mu$. All three metrics (E , μ , and c_T) related to stiffness are commonly quoted in the literature and provided by commercial systems. It should be noted that although it is very common for shear wave speeds to be equated with elasticity using (2) under an assumption of pure elasticity (i.e., ignoring viscosity), it is known that soft tissues are appreciably viscoelastic [42]–[45]. To that end, the spectral characteristics of the shear waves have been quantified to establish the frequency range for shear wave phase velocity estimates. Further, the physical characteristics of the tissue under investigation (e.g., size compared with the shear wave wavelength, structural anisotropy) violate the assumptions leading to (2).

Robust shear wave speed estimation is dependent on maximizing the displacement amplitude that can be generated for an acoustic radiation force excitation so that propagation can be monitored over larger propagation distances, allowing second-order parameters such as viscosity to be reconstructed. The algorithms that have typically been utilized in SWEI imaging include the lateral time-to-peak (LateralTTP) [46], Radon sum transformation [47], and random sample consensus (RANSAC) [48] methods. All of these methods are time-of-flight (TOF)-based algorithms that rely on assumptions of propagation direction and a propagating shear wave front that is not affected by diffraction effects. For this reason,

displacement amplitude, shear wave spectral content, and its distribution of displacement amplitude as a function of depth, are some of the primary output metrics characterized in this work.

C. Unique Challenges for SWEI of the Cervix

Although SWEI has had several years of development and clinical testing in the liver [34], [43], [49]–[52], the uterine cervix provides a unique ultrasonic and mechanical environment that requires changes to acquisition and processing of SWEI data. Unlike the liver, the pathologic state of the uterine cervix may result in a premature softening of the soft tissue from a baseline healthy (non-pregnant) stiff ($E \approx 25$ kPa) state [53]. SWEI has also been performed in organs, such as skeletal and cardiac muscle, where stiffness can range from $E = 16$ to 90 kPa, depending on the contractile state and orientation of the highly anisotropic muscle tissue, which can be higher Young's moduli than those of the relatively stiff, non-pregnant cervix [54], [55].

Previous studies of shear wave speed reconstruction accuracy have demonstrated that TOF-based algorithms for shear wave speed estimation methods suffer from greater variance in stiffer tissues caused by decreased displacement amplitudes and finite temporal and spatial resolution of propagating shear waves [46]. Uterine cervix tissue also has a greater acoustic attenuation than liver tissue (1.3 to 2.0 dB·cm⁻¹·MHz⁻¹ versus 0.5 to 0.7 dB·cm⁻¹·MHz⁻¹), which decreases the SNR of the raw radio-frequency data used to estimate displacements, reduces the penetration depth of the acoustic radiation force magnitudes that can generate measurable displacements, and generally reduces the peak displacement that can be generated in tissue [41].

In vivo cervical imaging typically employs an end-fire endovaginal array. Although highly effective for basic B-mode imaging, for instance to measure cervical length, challenges of this transducer design for quantitative evaluation of the cervix relate to its depth-dependent signal-to-noise ratio in tissue relative to the focal depth of the probe; in other words, regions that are deep and proximal are difficult to adequately access. The cervix consists of layers of aligned collagen that progressively disorganize before delivery. The most critical regions appear to be in the central layer of the cervix, and at its proximal end; a clear stiffness gradient is evident along the length of the cervix, and the greatest differences between subjects are present proximally [56]. Therefore a catheter-based imaging array may be more practical because it could be used for scanning along the length of the cervix from either within the cervical canal (Fig. 1, where the natural contours would create compression) or its external surface (with the transducer attached to the operator's finger so that it is compressed between the cervix and finger). Specifically, in contrast to standard end-fire endovaginal arrays, a catheter-based array allows for stiffness gradients to be seen at constant depths from the transducer face as a function of lateral position across the array. Disadvantages of catheter-based arrays compared with their handheld counterparts, however, are that they tend to have smaller elements, lack an elevation lens for focusing, and have limited fields of view, all of which can reduce the amplitude of acoustic radiation force-induced displacements and the precision of the resultant shear wave speed estimates. That said, although hand-held transducers work well for SWEI of certain tissues such as the

liver, they are not practical for endocavitary scanning of the smaller, more mobile cervix, in which the most desired regions to assess lie proximally and deep within the tissue.

This manuscript explores the feasibility of using a catheter-based array for performing SWEI in the uterine cervix using validated finite element method (FEM) and Field II [57] simulations, described in Section II. Section III presents the acoustic radiation force fields that are feasible to generate in cervical tissue using these arrays, and an analysis of the displacement and shear wave fields that result are presented. Finally, Section IV discusses the current elastic characterization of cervical tissue that can occur with acoustic radiation force methods, and the limitations and future challenges of these methods are also discussed for excised cervical tissue and *in vivo* cervical imaging.

II. Methods

The intended approach to cervical assessment is to insert the prototype catheter transducer (described below) into the endocervical canal, which will provide alignment of the transducer array with the gross anatomical structure parallel to the endocervical canal. The average thickness of the cervix wall (distance from the canal to the cervical serosa) is about 15 mm. An upper limit of cervix wall thickness is about 20 mm, so the following simulations investigated the shear wave properties in uniform media to this maximum depth. Although it is known that the cervix is composed of layers of aligned collagen [14], assuming homogeneous, isotropic media in this initial study is justified to provide guidance in system configuration and data acquisition.

A. Finite Element Method Simulations

Previously-validated FEM simulations [41], [58] were used to evaluate the spatial and temporal displacement fields generated by a variety of focal configurations associated with the prototype intracavity array transducer (Table I) for use on a Siemens Acuson S2000 ultrasound scanner (Siemens Medical Solutions USA, Malvern, PA).

Field II [57] was used to simulate the acoustic output from all transducer configurations (Table I) [41], [58]. All acoustic outputs were normalized to the maximum intensity present in the $\alpha = 1.3 \text{ dB} \cdot \text{cm}^{-1} \cdot \text{MHz}^{-1}$ simulation using an F/1 focal configuration; this was an arbitrary choice for these simulations, but all results presented herein can be linearly scaled based on empirical I_{sppa} measurements for specific experimental configurations. 3-D acoustic intensity fields (I) were converted to acoustic radiation forces (1). These body forces were converted to nodal point loads and applied throughout the 3-D mesh for the 400-cycle excitation duration (Table I).

The mesh for finite element simulations utilized 0.1 mm node spacing, tri-linear, cubic elements, spanning a total volume of $0.5 \times 0.7 \times 4.0 \text{ cm}$ (elevation $\times$ lateral $\times$ axial) and was generated using LS-PrePost (Version 3.2 β , Livermore Software Technology Corp., Livermore, CA). Flanagan–Belytschko hourglass control (hourglass coefficient = 0.1) was utilized for the mechanical simulations. Nonreflecting boundary conditions were imposed on all surfaces of the model to prevent shear wave reflections from the finite volume of the mesh into the region of interest. The elements were modeled as a linear, isotropic, elastic

material with Young's moduli ranging from 1 to 25 kPa and a Poisson's ratio of 0.499 (nearly incompressible). This range of Young's moduli was chosen based on studies of healthy excised human cervixes where the normal, non-pregnant cervix is stiffer (~25 kPa) than the ripened cervix (assumed to be 1 kPa in these studies). LS-DYNA 3D (Version ls971d R6.1, Livermore Software Technology Corp.) was used to solve the mechanical models using a time-domain, explicit solver [41].

As shown in Table I, three focal properties were tested (full-aperture, F/1.0, and F/2.0) and focal depths were varied from 0.5 to 2.0 cm. A transducer length of 14 mm (128 elements $\times$ 0.11 mm), can achieve f-numbers of F/1.0 and F/2.0 up to maximum focal depths of 14 mm and 28 mm, respectively. The simulations that follow include regions beyond these limiting depths for full-aperture and F/1.0 configurations to allow investigation of the acoustic intensity, radiation force, and axial displacement distributions for these limiting conditions.

The acoustic attenuation in the uterine cervix has been estimated to be approximately $1.3 \text{ dB}\cdot\text{cm}^{-1}\cdot\text{MHz}^{-1}$ [24], [59]. However, to investigate the influence of high acoustic attenuation on shear wave generation, simulations were performed assuming acoustic attenuations of 1.3, 1.7, and $2.0 \text{ dB}\cdot\text{cm}^{-1}\cdot\text{MHz}^{-1}$.

B. Characterizing Shear Wave Spectral Content

Although many tissue elasticity imaging methods make assumptions about soft tissues being purely elastic, it is also commonly accepted that most soft tissues are viscoelastic; therefore, the frequency content of an ARF-generated shear wave is important to characterize. The spectral content of simulated shear waves were characterized by taking the Fourier transform of particle velocity profiles generated from displacement data. Simulated displacement data were upsampled to 100 kHz and low-pass filtered using the interp function in Matlab (R2012b, The MathWorks Inc., Natick, MA).

III. Results

The relatively high acoustic attenuation of cervical tissue (compared with most soft tissues, such as liver) is a primary determinant of how much acoustic energy is available to generate appreciable acoustic radiation force, and there is a balance between focal gain and depth-dependent attenuation to determine the optimal focal depth to achieve greatest acoustic radiation force. Fig. 2 (top row) shows the normalized maximum acoustic intensity distributions in the imaging plane for 4- to 12-MHz excitations in media with acoustic attenuations of 1.3, 1.7, and $2.0 \text{ dB}\cdot\text{cm}^{-1}\cdot\text{MHz}^{-1}$ for full-aperture (128 element), F/1, and F/2 focal configurations. Each data point in each subplot is the result of a separate simulation of the acoustic intensity distributions in the imaging plane for a particular combination of excitation frequency, focal depth, and acoustic attenuation. The high acoustic attenuation of cervical tissue can cause the location of the maximum intensity to be shallower than the focal depth. This effect grows increasingly worse with higher attenuating media, with higher excitation frequencies, and in higher f-number focal configurations. (Fig. 3).

The radiation force magnitude that can be generated from an acoustic field is not exclusively dependent on the acoustic intensity (Fig. 2) but it is also directly scaled by the acoustic attenuation (1). Fig. 4 shows the normalized maximum acoustic radiation force generated for the acoustic intensity fields shown in Fig. 2 for the intracavity array. As in Fig. 2, each data point in each subplot is the result of a separate simulation of the acoustic intensity distributions in the imaging plane for a particular combination of excitation frequency, focal depth, and acoustic attenuation.

The axial distribution of acoustic radiation force is of great importance when generating shear waves where greater depths of field promote more uniform shear wave fronts and larger spatial domains over which shear wave reconstructions can be made. Fig. 5 shows the axial distributions of normalized acoustic radiation force for the full-aperture, F/1.0, and F/2.0 focal configurations with attenuation values of 1.3, 1.7, and 2.0 $\text{dB}\cdot\text{cm}^{-1}\cdot\text{MHz}^{-1}$. The depth of field for each excitation configuration was calculated from the simulated force data (Fig. 5) as the full-width at half-maximum of a Gaussian fit of the envelope of each profile to determine the axial range over which uniform shear wave fronts could be estimated for shear wave speed reconstruction (Table II) [46].

Shear wave elasticity imaging of viscoelastic soft tissues requires knowledge of the frequency content of the shear waves, which is dependent on the spatial distribution of the excitation beam in the lateral and elevation dimensions. Fig. 6 shows the normalized lateral distributions of acoustic radiation force for the full-aperture, F/1.0, and F/2.0 focal configurations in the 1.3, 1.7, and 2.0 $\text{dB}\cdot\text{cm}^{-1}\cdot\text{MHz}^{-1}$ attenuating media.

The transient shear waves that are generated by these acoustic radiation force excitations depend on the acoustic radiation force excitation geometry, the elasticity of the material, and the acoustic attenuation. Fig. 7 shows a representative set of images in an $\alpha = 2.0$ $\text{dB}\cdot\text{cm}^{-1}\cdot\text{MHz}^{-1}$ attenuating medium with full-aperture and F/1 focal configurations for 1, 10, and 25 kPa materials at three points in time after the radiation force excitation. The maximum displacement associated with each of these displacement fields varies as a function of the stiffness of the material, as does the time when this maximum displacement occurs (Fig. 8).

In addition to the axial extent of the displacement field and the displacement amplitudes that are generated, the lateral width of the excitation (Fig. 6) will affect the spatial frequency content of the resulting shear waves, which can be important in media with a significant viscous component. Spatially compact shear waves have benefits in improving the spatial resolution of quantitative shear modulus estimates when using smaller motion tracking kernels, but they can be more susceptible to noise when estimating the shear wave arrival time at different spatial locations. Fig. 9 shows the spectral content (b) of the particle velocity profiles (a) 3.5 mm offset from the excitation for a representative F/1.0 focal configuration in an $\alpha = 2.0$ $\text{dB}\cdot\text{cm}^{-1}\cdot\text{MHz}^{-1}$ medium with Youngs' moduli ranging from 5 to 25 kPa. The center frequencies and half-power bandwidth points (error bars) for these configurations are shown in Fig. 9(c). Table III summarizes the center frequencies and half-power bandwidths for the full aperture, F/1, and F/2 focal configurations.

IV. Discussion

Although SWEI has become popular in organs such as the liver, SWEI of the cervix provides several unique challenges that were addressed in this simulation study. As mentioned in Section I-C, the healthy state of the non-pregnant cervix involves the organ in its stiffest elastic state (~ 25 kPa), meaning that higher acoustic radiation force magnitudes must be generated to induce shear wave displacements that can be adequately tracked [60]. The stiffer cervical tissue also results in higher shear wave speeds (2), meaning that with limited-pulse-repetition-frequency (PRF) ultrasonic imaging systems, shear waves will traverse larger spans of the active imaging aperture in adjacent time steps, reducing the accuracy and precision of the shear wave speed estimates [46]–[48]. This challenge in tracking the shear wave speed laterally can be improved by having a greater axial extent over which the shear wave speed is estimated [61].

In addition to the higher baseline stiffness of the cervix, the higher acoustic attenuation compared with other soft tissues also makes generating acoustic radiation force challenging, where penetrating deeper than 10 mm with appreciable energy is optimized by using frequencies lower than 7 MHz, with a focal depth of ~ 7 mm for this intracavity array (Figs. 2 and 4), even though these lower frequencies deviate from center frequency of this transducer (Table I). Setting focal depths deeper than this optimized location results in peak energy being deposited shallower than specified focal depth; this near-field shift in peak acoustic intensity relative to the focus is even more egregious for higher frequency excitations and higher acoustic attenuation (Fig. 3).

Higher f-number focal configurations and deeper focal positions provide longer focal zones (Fig. 5 and Table II) and more uniform shear wave wavefronts (Fig. 7) for improved shear wave speed estimation, but that benefit comes at the expense of a loss in acoustic radiation force amplitude and the resulting displacement amplitude (Fig. 8). Given the small elements of an intracavity array (Table I) and the high acoustic attenuation of cervical tissue, it is expected that concessions will need to be made (reducing the extended depth of field of higher f-number focal configurations to optimize the active aperture of the excitation beam for increased displacement amplitude). To compensate for the reduced axial extent of the shear wave front (Fig. 7), larger lateral tracking apertures will need to be used to track more lateral locations of shear wave propagation in the tissue for higher accuracy and precision in shear wave speed estimation [61].

It should be noted that these highly-focused (full-aperture or F/1.0), high-excitation-frequency acoustic radiation force pulses yield spatially tighter excitation beams than have been used for SWEI in deeper organs, such as the liver, where 2-MHz pulses with F/2 focal configurations focused at 5 to 8 cm are common [34]. These highly-focused shear waves also have very limited depths of field (< 2 mm, Fig. 5), which could limit characterization of the entire cervical tissue thickness, but could be desirable in future studies where the individual layers of the cervix are targeted for shear stiffness characterization.

The narrower shear waves from highly-focused beams have considerably higher frequency content (Fig. 9, Table III) than what is found in liver SWEI (~ 60 to 600 Hz) [62]. This

higher shear wave spectral content can have significant implications in shear wave speed estimates for the viscoelastic cervix, where the phase velocity at these higher frequencies may yield higher shear modulus estimates than lower frequency or quasi-static stiffness measurements on similar tissue. It should be noted that the frequency content presented in this manuscript (Fig. 9) represents spectral energy that is supported in an elastic medium, where the propagation medium itself does not filter the spectral content of the generated shear wave. Viscous tissues will inherently act as filters for the propagating shear waves, which will skew the center frequencies and bandwidths that would be measured with this simulated experimental setup in viscous materials. This skew would result in a biasing of group velocity estimates to phase velocity components that are preserved by the transfer function of the material, with stiffer media creating biases toward higher phase velocities, and more viscous media attenuating the higher frequency phase velocities as a function of propagation distance.

One significant assumption in these models has been that the expression for acoustic radiation force magnitude (1) and the Field II simulation of the acoustic intensity fields are representative of what would be present *in vivo*. The expression for acoustic radiation force was derived for plane wave propagation (i.e., not highly focused) [38], and the Field II simulations are linear and do not take into account nonlinear propagation effects. Nonlinear propagation would lead to an effectively higher attenuation of the tissue, leading to more shallow energy deposition (closer to the focus of the excitation), and the beamwidth in the focal depth of field would be narrower because of the higher frequency content. These effects are not reflected in these simulations. Additionally, the effect of phase aberration, which would distort the beam focusing, could also have a significant effect on the excitation and the resultant tracking of propagating shear waves, but that has also not been addressed in these simulations.

Although these simulations have addressed the feasibility of using a catheter-based array for cervical imaging, they have not addressed some important analyses that must be done before this imaging is performed *in vivo*—specifically, safety studies. The focal configurations evaluated in this manuscript must be evaluated in the context of mechanical and thermal indices, using absolute pressure and intensity values that must be measured with hydrophones. Additionally, the thermal burden on the catheter arrays must also be measured, both for patient safety concerns and to ensure the long-term performance of these arrays.

V. Conclusions

ARFI excitations with an intracavity ultrasound array in the uterine cervix can cause adequate displacements to generate shear waves in cervixes (25 kPa) when using excitation frequencies less than 7 MHz with full-aperture excitations. With an optimized configuration of 5 to 6 MHz excitation frequency focused at 7 mm for acoustic attenuation ranging from 1.3 to 2.0 dB·cm⁻¹·MHz⁻¹, relatively planar shear wave fronts are created. The use of higher frequency excitations results in significant shifts (toward the transducer) in the location of the peak acoustic radiation force generated, which would result in shear waves not being localized where the acoustic energy is being focused. The use of larger excitation apertures can significantly improve the displacement amplitudes generated by the acoustic radiation

force in these highly attenuating tissues, but these large apertures restrict the axial extent of the resulting shear wave front to as little as 2 mm. These tightly focused acoustic radiation force excitations can also result in shear waves that have significantly higher frequency content (1.5 to 2.5 kHz) than the shear waves used in imaging deeper organs, and that could result in higher elastic modulus estimates in soft tissues with significant viscosity and a significant spread in shear wave phase velocities. These simulation studies support the feasibility of cervical tissue SWEI using a catheter-based array and provide a foundation for progressing to *ex vivo* and *in vivo* studies.

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Fig. 1.

(a) Envisioned placement of a catheter probe in the cervical canal to image stiffness gradients along the length of the cervix. (b) A close-up diagram of a propagating shear wave front being generated by the catheter array in the cervical tissue when placed in the cervical canal. The catheter array could also be placed adjacent to the external surface of the cervix, fixed in place between the operator's finger and the cervix. In both catheter placements, the array would be secure to avoid motion-based artifacts during imaging.

Fig. 2.

Simulated maximum acoustic intensity distributions in the intracavity array imaging plane for 4- to 12-MHz excitations, as a function of focal depth for 1.3, 1.7, and 2.0 $\text{dB}\cdot\text{cm}^{-1}\cdot\text{MHz}^{-1}$ attenuating media for full-aperture, F/1, and F/2 focal configuration. The maximum intensity images for each f-number have been normalized to the $\alpha = 1.3$ $\text{dB}\cdot\text{cm}^{-1}\cdot\text{MHz}^{-1}$ acoustic attenuation case (the color bar shows the ratio of values in decibels). Note that the peak intensity drops with greater acoustic attenuation and that each configuration has a “sweet spot” between 5 and 7 mm where the balance between focal gain and depth-dependent attenuation favor increased acoustic output. Excitation frequencies above 8 MHz generate significantly smaller acoustic radiation force amplitudes compared with the lower frequency excitations. Note also that the F/1 focal configuration only applies for depths up to 14 mm. Beyond that depth, the results are identical to those for the full-aperture case.

Fig. 3.

Simulation results showing the axial shift in location of maximum in the acoustic intensity distribution for the intracavity array for 4- to 12-MHz excitations, as a function of focal depth. Three different attenuation values were tested [1.3 (left column), 1.7 (middle column), and 2.0 (right column) $\text{dB}\cdot\text{cm}^{-1}\cdot\text{MHz}^{-1}$] in combination with three different focal configurations [full-aperture (top row), F/1.0 (middle row), and F/2.0 (bottom row)]. Each data point on each subplot shows the result of a simulation for a particular combination of focal depth, excitation frequency, attenuation value, and focal configuration (the color bar shows the difference in location of the peak acoustic intensity and the location of the acoustic focus in centimeters). Note that the relatively high acoustic attenuation of the cervix can cause the location of this peak intensity to be shallower than the focal depth. Simulation parameter combination results shown in blue experience the greatest shift toward the transducer face (negative shifts).

Fig. 4.

Normalized maximum acoustic radiation force generated with the intracavity array as a function of excitation focal depth, frequency, and acoustic attenuation for full-aperture, F/1, and F/2 focal configurations. These forces correspond to the normalized intensity fields shown in Fig. 2, where the normalization for each focal configuration is based on the maximum intensity associated with the $\alpha = 1.3 \text{ dB}\cdot\text{cm}^{-1}\cdot\text{MHz}^{-1}$ acoustic attenuation field (the color bar shows the ratio of values in decibels). Note that the F/1 focal configuration only applies for depths up to 14 mm; beyond that depth, the results are identical to those for the full-aperture case.

Fig. 5.

Normalized axial distribution of acoustic radiation force along the center of the region of excitation for the maximum force configurations obtained from each focal configuration and attenuation combination presented in Fig. 4. Note that, as expected, the smaller excitation apertures (higher f-numbers) yield larger depths of field, with increased near-field intensity distributions that can complicate shear wave speed estimation.

Fig. 6.

Normalized lateral distribution of acoustic radiation force for the maximum force obtained from each focal configuration and attenuation combination presented in Fig. 4. These lateral excitation (intensity and force) distributions dictate the frequency content of the resultant shear waves.

Fig. 7.

Displacement fields illustrating shear wave propagation in the imaging plane for full-aperture (left three columns) and F/1 (right three columns) focal configurations with an ARF excitation frequency of 6 MHz focused at 7 mm in a medium with an attenuation coefficient of $\alpha = 2.0 \text{ dB} \cdot \text{cm}^{-1} \cdot \text{MHz}^{-1}$ and Young's moduli of 1 kPa (first column), 10 kPa (second column), and 25 kPa (third column) for times of 0.4 ms (top row), 0.9 ms (middle row), and 1.4 ms (right row) after the initiation of the acoustic radiation force excitation. Note that the color bars, representing axial displacement (in micrometers), are different for each image and have not been normalized.

Fig. 8.

Normalized maximum displacements (a) and time-to-peak displacements (b) after initiation of acoustic radiation force excitation as a function of tissue stiffness for the full-aperture and F1.0 focal configurations at 6 MHz in an $\alpha = 2.0 \text{ dB} \cdot \text{cm}^{-1} \cdot \text{MHz}^{-1}$ attenuation medium, focused at 7 mm. The 1 to 25 kPa stiffness range represents that expected for the cervix as it ripens throughout pregnancy.

Fig. 9.

(a) Particle velocities 3.5 mm offset from the center of the excitation for stiffnesses of 5 to 25 kPa using an F/1.0 focal configuration in an $\alpha = 2.0 \text{ dB} \cdot \text{cm}^{-1} \cdot \text{MHz}^{-1}$ medium. (b) The corresponding spectral content of these velocity profiles. (c) The center frequencies and half-power bandwidths (error bars) of these velocity spectra. Note how the center frequencies and bandwidths of the particle velocities both increase as a function of increasing stiffness.

TABLE I

Acoustic Radiation Force Excitation Focal Configuration and Transducer Parameters Used in These Simulation Studies for the Prototype Intracavity Array Transducer.

	Prototype Intracavity Array Transducer
Maximum elements (full aperture)	128
Element width (mm)	0.09
Element kerf (mm)	0.02
Element height (mm)	2.5
Focal configuration	Full Aperture, F/1, F/2
Focal depth (mm)	{5, 6, 7,..., 20}
Frequency (MHz)	{4, 5, 6, ..., 12}
Attenuation (α , $\text{dB}\cdot\text{cm}^{-1}\cdot\text{MHz}^{-1}$)	{1.3, 1.7, 2.0}
Center frequency (MHz)	7.2
Fractional bandwidth	0.5
Elevation focus (mm)	None
Excitation duration (cycles)	400

Note that this array lacks a lens and does not have any elevation dimension focusing.

TABLE II

Depths of Field for the Acoustic Radiation Force Fields Shown in Fig. 5, Estimated as the Full-Width at Half-Maximum of a Gaussian Fit of the Envelope of Each Profile.

	1.3 dB·cm⁻¹. MHz⁻¹	1.7 dB·cm⁻¹. MHz⁻¹	2.0 dB·cm⁻¹. MHz⁻¹
Full aperture (mm)	0.5	0.5	0.5
F/1.0 (mm)	2.3	2.0	2.1
F/2.0 (mm)	5.9	5.1	6.3

Note that for the F/2.0 focal configuration, the width is greater than that of the main peak in energy because the envelope includes the near-field intensity that contains appreciable energy relative to the peak in the axial distribution.

TABLE III

Summary of Spectral Center Frequencies and Half-Power Bandwidths for the Full-Aperture, F/1, and F/2 Apertures as a Function of Young's Modulus.

Young's modulus (kPa)	Focal configuration	Center frequency (kHz)	Half-power bandwidth (kHz)
5	Full aperture	0.8	1.5
	F/1	0.8	1.7
	F/2	0.8	1.6
10	Full aperture	1.2	2.4
	F/1	1.2	2.4
	F/2	1.2	2.1
15	Full aperture	1.4	3.0
	F/1	1.4	3.0
	F/2	1.4	2.5
20	Full aperture	1.6	3.2
	F/1	1.6	3.0
	F/2	1.6	2.9
25	Full aperture	2.0	3.3
	F/1	2.0	3.3
	F/2	1.8	3.1