DETECTION OF INTRAOCULAR PRESSURE CHANGE IN A HUMAN EYE MODEL USING SONOELASTIC DOPPLER ULTRASOUND

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Abstract- We propose to use Sonoelasticity for clinical measurement of intraocular pressure (IOP). We postulate that the change of fluid pressure in a fluid-filled elastic shell, such as the eye, would change the apparent elasticity of the shell, which in turn would change its frequency of resonance. If a standardized normal range of resonance for the human eye can be found, any abnormality in the resonance can then be used to detect abnormalities in the IOP.

Preliminary experiments have been done using eviscerated and enucleated human and pig eyes. IOP has been varied by insertion of a saline fluid filled catheter with a hydrostatic pressure. Doppler ultrasound has been used to find the amplitude of vibration of the sclera. As little as 4 mm Hg change in IOP has been detected.

I. INTRODUCTION

All current methods for clinical determination of IOP, except for digital palpation, require deformation of the cornea and measurement of the amount of deformation or the force required to obtain the deformation [1]. Digital palpation, however, is inevitably subjective and physician dependent. The use of resonance to detect change in IOP had independently been proposed by Hamelink and Cloud [2] using laser velocimetry. Sonoelasticity [3] would offer a much simpler technique for the same purpose, and may be of great use for measuring IOP in presence of significant corneal disease or through a closed lid. Further application of this alternative method might be in 24-hour monitoring of IOP.

In our study, the eviscerated eyes were subjected to low frequency vibrations. Pulsed Doppler ultrasound was used to evaluate the vibration amplitude within the sclera by counting the number of Bessel side bands. IOP was controlled using an infusion of normal saline solution under hydrostatic pressure.

Two different types of experiments were performed. In one case, the frequency of vibration was fixed, and the IOP was varied in the range 6-100 mm Hg. This produced variation in the vibration amplitude with change in IOP. In the other case, IOP was fixed somewhere between 6-65 mm Hg. Then the vibration frequency was changed to determine the frequency response and resonances.

II. THEORY

A. Resonance of the Eye:

It is difficult to obtain an expression for the resonance of a spherical fluid-filled elastic sphere. However, a more simplified model can be used for this case. The spherical fluid-filled elastic shell can be roughly approximated as a hollow thin spherical shell. The pressure change can be accounted for by changing the elastic constant[1]. The vibration of a thin spherical shell has been investigated by H. Lamb [4]. All the modes of vibration are extensional, and they fall in two classes which are characterized by the absence of the radial component of displacement and the absence of the radial component of the rotation. In any mode of either class the displacement is expressible in terms of spherical surface harmonics of a single internal degree. In the case of vibrations of the first class the frequency $\omega\ (=2\pi f)$ is related with the degree n of the harmonics by the equation

$$\frac{\omega^2 a^2 \rho}{\mu} = (n-1)(n+2) \qquad (1)$$

where a is the radius of the sphere, ρ is the density of the shell, μ is rigidity, and σ is the Poisson's ratio.

In case of vibrations of the second class, which is of our main interest here, the frequency can be determined from the following equation,

$$\frac{\omega^4 a^4 \rho^2}{\mu^2} - \frac{\omega^2 a^2 \rho}{\mu}\left[(n^2+n+4)\frac{1+\sigma}{1-\sigma} + (n^2+n-2)\right] + 4(n^2+n-2)\frac{1+\sigma}{1-\sigma} = 0 \qquad (2)$$

where each resonance mode (n=1,2,3,·····) corresponds to two positive real resonance frequencies that are solutions to equation (2).

For scleral tissues, we used,

$a = 2.5$ cm, $\rho = 1$ gm/cc,

$\sigma = .5$, $\mu \approx 1000$ kPa.

For the normal human eye, scleral Young's modulus, $E_s = 2700$ kPa [5]. μ is approximately a third of the Young's modulus, and thus, for the sclera, $\mu_s \approx 1000$ kPa. To see how the resonance shifts with changing rigidity, we evaluated the resonance frequencies for μ = 300, 1000, and 3000 kPa (lower than, at, and higher than the normal value). The resonance is seen to shift significantly for these changes as shown in the accompanying figures which give the larger real positive frequency solutions to the first six resonance modes (n = 1,2,···6). Fig 2 shows the shift in the resonance when rigidity changes to 1000 kPa from 300 kPa. Only 6 resonance pairs are shown. The lowest resonance at 300 kPa is at 93 Hz, which goes up to 169.7 Hz at 1000 kPa. We can see that the lowest resonance is appreciably changed when

[1] This would not, however, account for the viscous damping of the fluid media.

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the rigidity increases. Fig 3 shows how the resonance would change when rigidity changes from 1000 kPa to 3000 kPa. In this case, first resonance increases to 293.9 Hz from 169.7 Hz as rigidity changes from 1000 KPa to 3000 kPa. Thus, we can expect to see the resonance frequencies to shift upward as IOP increases.

B. Estimating the Vibration Amplitude:

When a scattering object vibrates in a manner to produce a wavelength much larger than the geometric dimensions of the scatterer itself, the Doppler spectrum of the signals returning from sinusoidally oscillating structures is similar to that of a pure tone frequency modulation process [6].

The received or scattered wave can be written as [6,7]:

$$s_r(t) = A \cos\left\{2\pi f_o t + \frac{\Delta f_m}{f_L} \sin(2\pi f_L t + \varphi)\right\} \qquad (3)$$

where

A is the amplitude of $s_r(t)$,

f_o is the center frequency,

f_L is the vibration frequency,

φ is the vibration phase,

and,

$$\Delta f_m = \frac{2\, v_m f_o \cos\theta}{c_o} \qquad (4)$$

where

$v_m = 2\pi f_L \xi_m$ is the vibration amplitude of the velocity field,

ξ_m is the vibration amplitude,

c_o is the propagation speed of the wave,

and, θ is the angle between the wave propagation and the vibration vectors.

Using trigonometric identities:

$$s_r(t) = A \sum_{n=-\infty}^{n=\infty} J_n(\beta) \cos[2\pi f_o t + n(2\pi f_L t + \varphi)] \qquad (5)$$

where the modulation index β is directly related to the vibration amplitude as follows:

$$\beta \equiv \frac{\Delta f_m}{f_L} = 4\pi \frac{\xi_m}{\lambda_o} \cos\theta \qquad (6)$$

λ_o = wavelength.

For the experiments, it was necessary to estimate the vibration amplitude (or β) from the available Doppler data. Several techniques had been proposed in the past [6, 8-10] to estimate the vibrational parameters. We used a simple approach proposed by Holen, et al [8], who measured the vibration amplitude of oscillating heart valves by counting the number of significant harmonics over a certain threshold. This procedure, albeit relatively course, is sufficient for resolving sub-millimeter vibration amplitudes, and is related to the fact that, for small β, the Doppler bandwidth is proportional to the modulation parameter β, or amplitude of oscillation. Future studies may, however, use a more accurate estimator, such as the standard deviation or spectral spread of the power spectrum [9].

III. Laboratory Studies

Eviscerated human and pig eyes were used for the experiment. Two infusion ports, 90° apart at the equator, were created. One port was connected to collapsible bottle of normal saline; the other port to an open-ended, elevated tube to be used as a manometer. Bottle height was varied to change IOP, as measured by the fluid level in the manometer. Leakage from incision sites were minimal.

Eyes thus prepared were immersed in ultrasound gel and held in place by stand-off rings to prevent contact between the Doppler ultrasound transducer (5 MHz and 7.5 MHz) and the eye. They were vibrated using standard 4" speakers. The speakers were coupled to the experimental apparatus and excited so as to produce uniform vibration amplitude between 200 Hz and 900 Hz. The sample volumes used for Doppler measurement were typically 2-3 mm. The experimental set up is shown in Fig. 1.

IV. Results

From the experimental data, two graphs are enclosed. Fig. 4 shows the response of one eye at IOP's of 8 mm, and 32 mm Hg. The upward shift in the resonance as the IOP increases is quite visible. For another eye (fig. 5), response is plotted at 8 mm, 18 mm, and 30 mm Hg. For this eye, we can see more than one resonances at the two higher IOP's. The shifts are clearly visible here also.

Further experiments will be required to determine if these effects have a useful clinical application in vivo where complex boundary conditions and natural variations in the eye are encountered.

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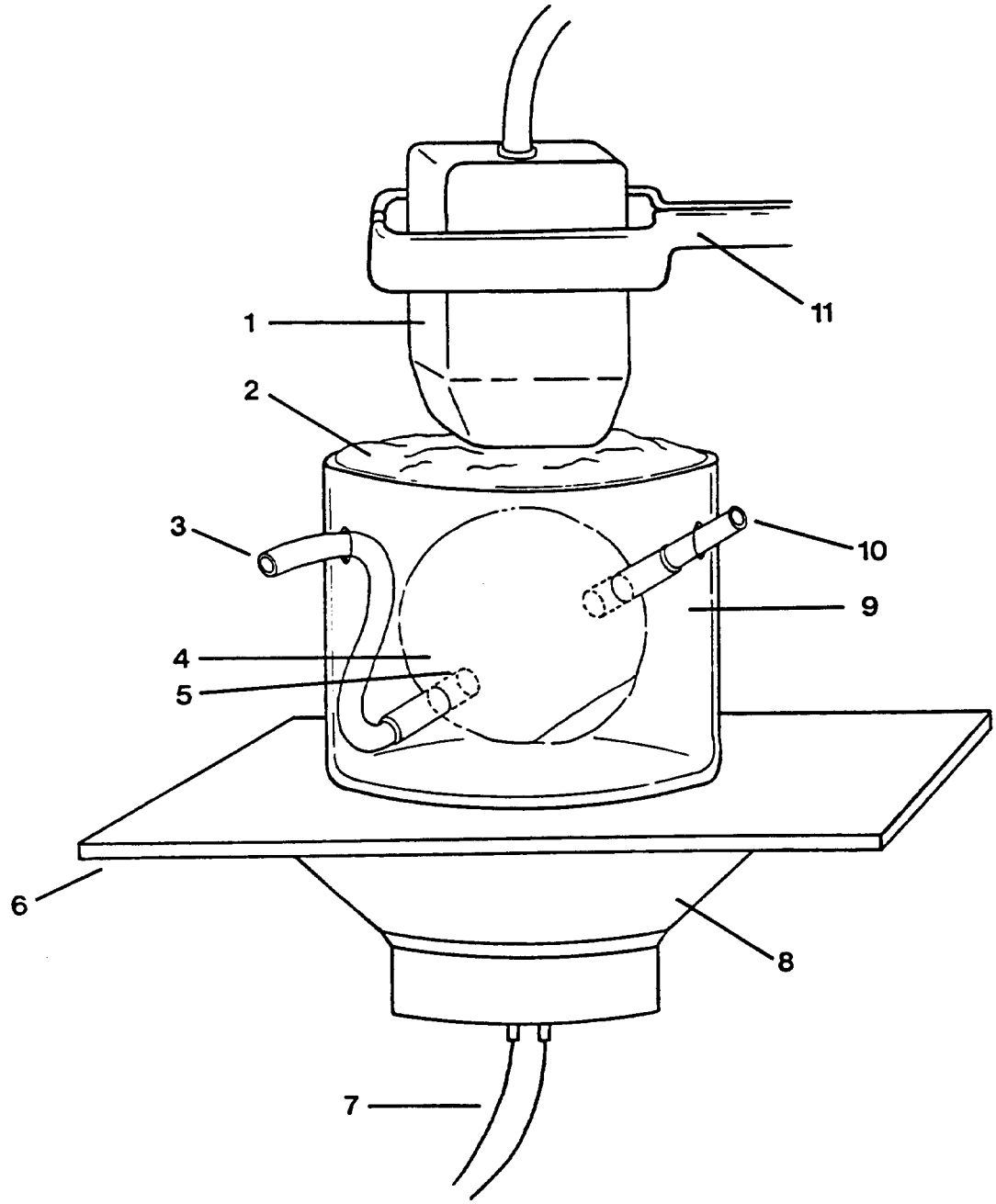

Fig. 1: Schematic diagram of experimental set up.
1) Doppler ultrasound transducer, 2) Ultrasonic gel, 3) To manometer, 4) Eviscerated eye, 5) Vitreous infusion port, 6) Isolation plate, 7) Audio signal input, 8) Speaker, 9) Specimen holder, 10) Normal saline infusion, 11) Adjustable clamp.

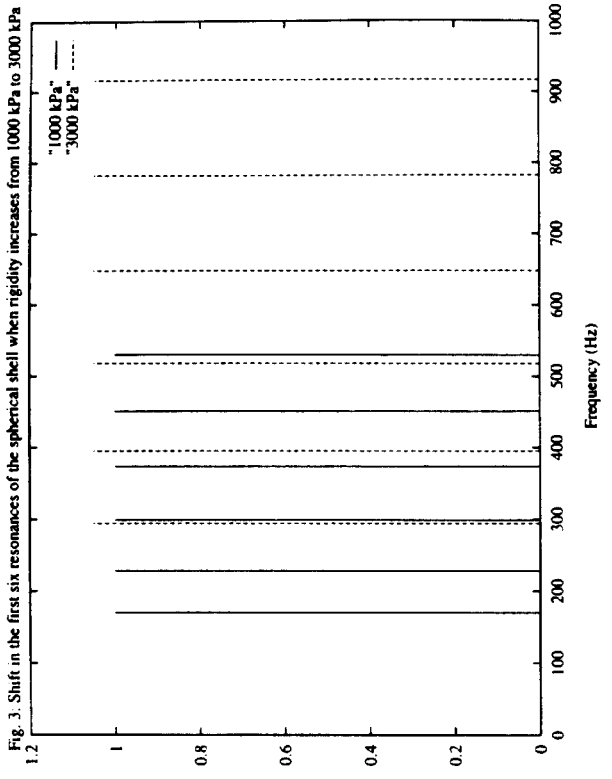
Fig. 3. Shift in the first six resonances of the spherical shell when rigidity increases from 1000 kPa to 3000 kPa
"1000 kPa"
"3000 kPa"
Normalized Amplitude
Frequency (Hz)

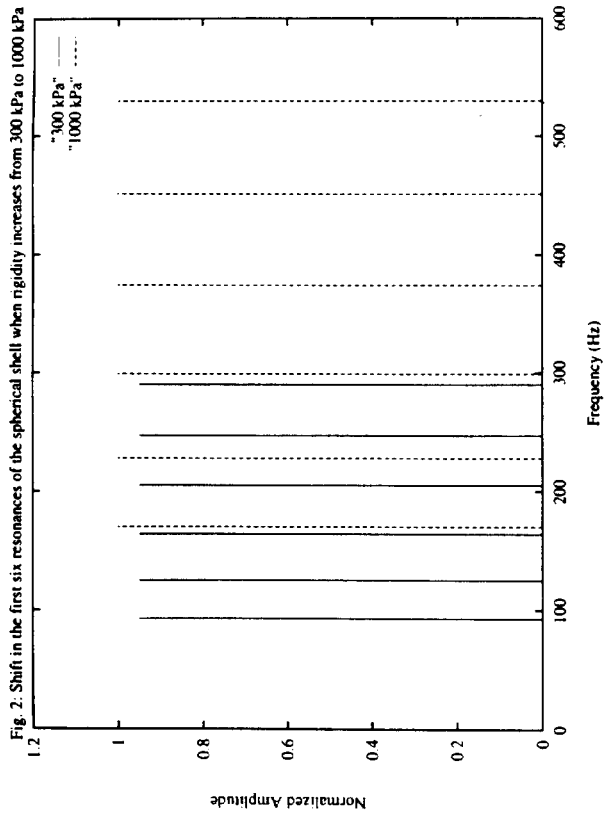
Fig. 2. Shift in the first six resonances of the spherical shell when rigidity increases from 300 kPa to 1000 kPa
"300 kPa"
"1000 kPa"
Normalized Amplitude
Frequency (Hz)

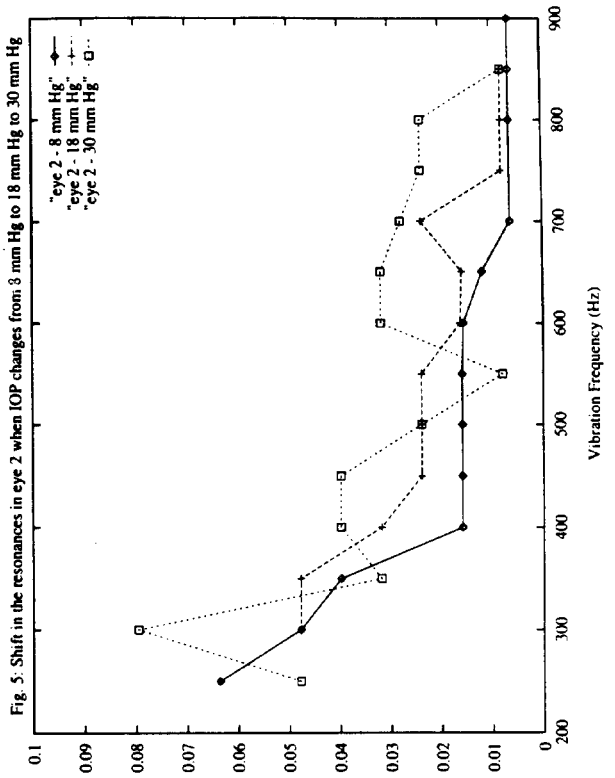
Fig. 5: Shift in the resonances in eye 2 when IOP changes from 3 mm Hg to 18 mm Hg to 30 mm Hg
"eye 2 - 8 mm Hg"
"eye 2 - 18 mm Hg"
"eye 2 - 30 mm Hg"
Peak Displacement (mm)
Vibration Frequency (Hz)

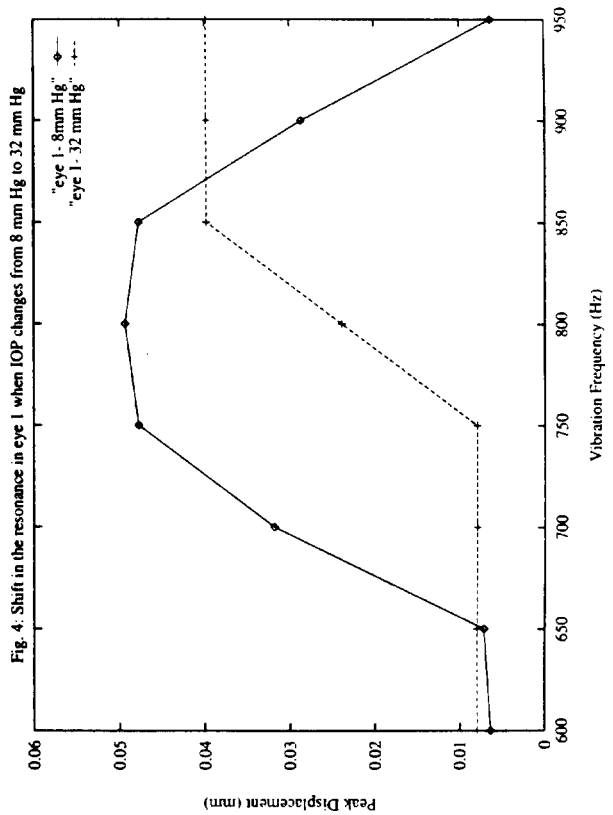
Fig. 4: Shift in the resonance in eye 1 when IOP changes from 8 mm Hg to 32 mm Hg
"eye 1- 8mm Hg"
"eye 1- 32 mm Hg"
Peak Displacement (mm)
Vibration Frequency (Hz)