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COMPUTER SIMULATION OF ULTRASONIC M-MODE IMAGE FORMATION

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INTRODUCTION

Study of the response of internal tissue structures to motion has long been recognised as a major source of information in diagnosis of various pathologies. Recent attempts to monitor this effect by recording the spatial location of ultrasonic echoes have demonstrated the potential of quantitative studies of tissue motion by ultrasound. Nicholas (1) reported patterns of temporal variations of backscattering intensity from a small ($30mm^3$) region of interest, defined by the beam shape and the receiving gate length. This approach was based on an earlier diffraction technique (2), used in a study of tissue structure. An alternative method, adopted by Dickinson and Hill (3), used temporal variations of correlation between consecutive pairs of A-scans as a measure of movement. This technique has been developed (4) to clinical implementation in the study of liver. Wilson and Robinson (5) reported results of in vivo echo tracing, using the RF A-scans.

Although all of the above studies showed differences in patterns of motion between various normal and pathologically altered tissues (mainly liver), in order to select the most effective technique and, more importantly, to optimise the analysis and the choice of variables selected, a number of practical measurements are necessary. In practice, many of the parameters in analysis are the function of equipment characteristics and, as such, cannot be altered easily. A similar study, however, can be carried out easily and more effectively by theoretical simulation, which would mimic the changes in spatial location of ultrasonic echoes in a medium of known mechanical characteristics, undergoing movement. This paper is a work in progress and reports on development of one such simulation; the results of computer modelling are compared with those obtained in a series of laboratory experiments.

THEORETICAL SIMULATION

Mechanical considerations

Soft biological tissues can be modelled as linear visco-elastic media. The equation of motion governing the displacement in such a medium from the position of equilibrium has been discussed previously (6). In a living tissue, when voluntary motion (such as limb movement) is stopped and respiration is suspended, the remaining motion results mainly from cardiovascular activity, hence we are dealing with very low frequencies (~1Hz). Considering the results reported by Truong et al (7) for frog sartorious muscle and Hutchison (8) for canine carotid artery, we have accepted Dickinson and Hill's (3) assumption that at low frequencies soft biological tissues can be considered as elastic media. Thus, the simplified version of Nyborg equation (6) for elastic medium is given as:

$$\rho \frac{d^2U}{dt^2} - E \frac{d^2U}{dZ^2} = 0 \qquad (1)$$

where ρ is the density, U(z,t) is the displacement, E is the elasticity and z the axis of the wave propagation.

Let us consider a simple geometry (Fig.1), where P(t) is the pressure distorting the tissue with elasticity E and density ρ. The rigid wall z=L represents a structure tethering the tissue (in liver, the diaphragm).

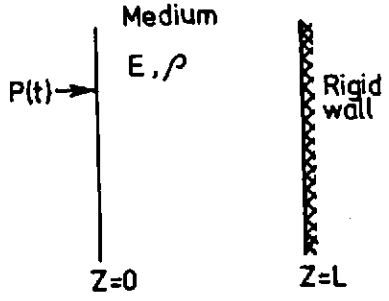

Figure 1

The lowest resonance frequency in the above model, f=c/2L (where c is the velocity of wave propagation) is, for L=10cm and c=15 ms^{-1}, 75Hz, which is much larger than cardiac frequency. Thus, the elastic response of the above system can be considered quasi-statistic and the following can be accepted as a simple solution.

$$U(z,t) = \frac{P(t) \cdot (L - z)}{E} \qquad (2)$$

The above equation gives the displacement of a point at z at time t.

A-scan simulation

Acoustic structure of soft tissues and interaction of sound with soft tissue is a very complex phenomenon (9,10). In this report, several assumptions regarding interaction of ultrasound with tissues are being made, some of which might not be accurate. However, they simplify computations and yield results which provide a fair agreement with experimental measurements.

We assumed that the scattering structure consists of a large number of isotropic rigid point scatterers of random strength, randomly distributed in three dimensions, in a homogeneous medium of density ρ and elasticity E. We further assumed that the scattering pressure of each scatterer is much smaller than the pressure of the incident wave, and that the superposition holds. Finally we assumed that the tissue attenuation and scattering of sound velocity are negligible.

The mathematical development of the computer model used to simulate A-scans was similar to the model developed by Foster et al (11). In this model, the RF A-scan is given as

$$V_{out}(t) = P(t) * \sum_{i} W_i \cdot H(r,t) \qquad (3)$$

where p(t) is the acoustic pulse (which can be measured experimentally by placing a small target on the axis of the transducer at the focal plane, in front of the transducer), H(r,t) is the acoustic response of the transducer, which is either derived analytically from the geometry of the transducer (12, 13) or measured experimentally; W is the scattering strength of the scatterer i. Thus, H(r,t) is the summation of the acoustic response of all scatterers in a region located at a distance r(ct1/2ct2) from the centre of the transducer, weighted for their strength.*

(*indicates a convolution)

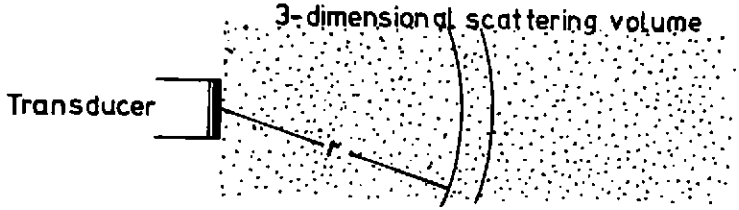

Figure 2

The RF A-scan V(t) was rectified and filtered to produce a video A-scan.

RESULTS

Experimental measurements

Tissue phantoms used in laboratory experiments were 8x8x5cm gelatine-amalgamate blocs of known elasticity, with a random suspension of spherical glass beads of a diameter 10-50μm. Oscillatory movement was induced by means of a simple mechanical device, with a maximum displacement of 4mm. A commercial real-time scanner, equipped with a circular transducer of 3.5MHz. spherically focused over 3-8cm, was used for scanning the phantoms.

(a)

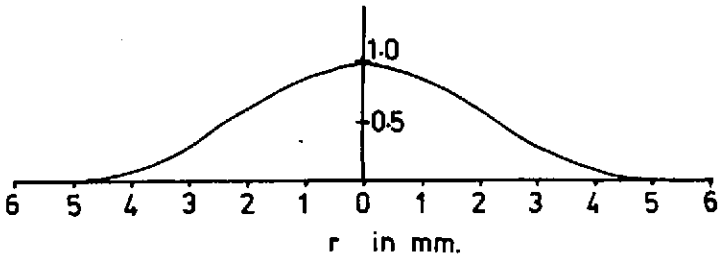

Figure 3. (a) The directivity of the transducer at the region of interest.

(b)

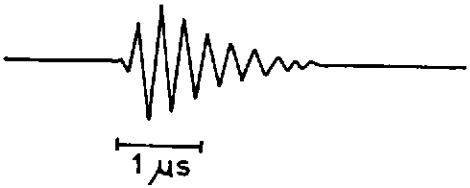

Figure 3. (b) The acoustic pulse at the region of interest.

The pulse and beam shape were measured using a small scattering wire phantom; the results are shown in figure 3. The geometry of measurements is illustrated in figure 3, showing the two positions used, i.e. movement along the ultrasonic beam (fig. 4a) or across the beam (fig. 4b).

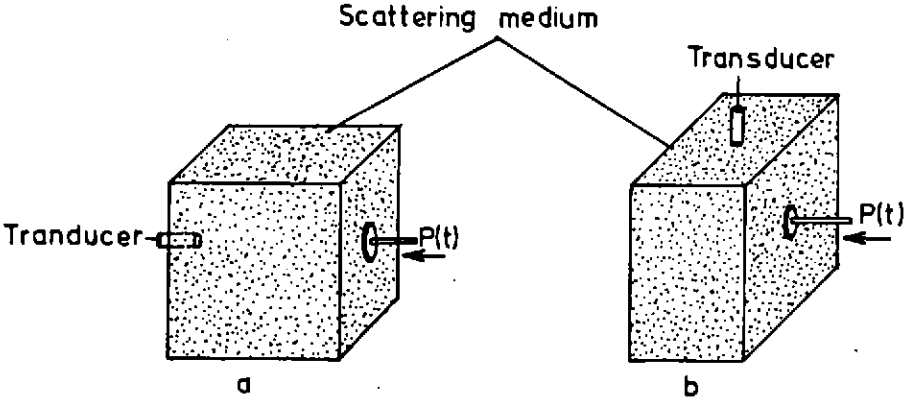

Figure 4. Experimental tissue movement.
(a) along the beam, (b) across the beam.

In all measurements, B-scans were collected first, to align the transducer (either parallel or perpendicular) to the direction of the movement vector. Following that, M-mode images were recorded, examples of which are shown in figure 5.

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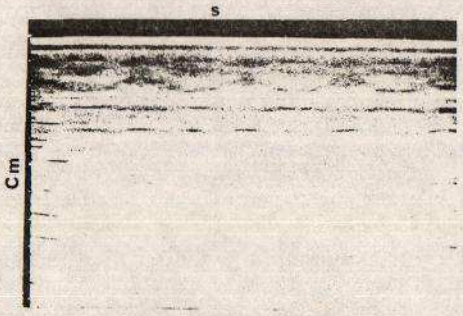

Figure 5. Experimental M-mode digitised to 512x512

Simulation of M-mode images

The region of interest was located in a geometry corresponding to that used in experimental setting. The model was a 2.56 x 2.6 x 2.56cm of isotropic medium of known density ρ and elasticity E. The chosen scattering density was $3400cm^{-3}$ and scatterers of random strength were assumed to be randomly distributed in this volume. Positions of the scatterers were modelled to change with time, along or across the ultrasonic beam, as appropriate. Every 1/128sec the positions were computed from equation 2 and a new video A-scan was evaluated, until a 256 x 256 image was obtained. Figure 6 shows an example of M-mode simulation for 2.56cm of tissue, moving along the beam.

Figure 6. Theoretically simulated M-mode (256x256)

CORRELATION METHOD OF DATA ANALYSIS

Visual comparison of the M-mode waveforms, obtained in experimental and simulated studies, was assisted by comparison of correlation patterns computed in statistical analysis. This analysis was based on an assumption that, when a sample of tissue moves, an A-scan collected from the interrogated region of interest will accordingly change. The degree of A-scan changes with time can be quantified using a correlation coefficient R, defined as

$$R' = 1.0 - \sum_{i=1} \left[y_i(t) - \bar{y}(t) \right] \left[y_i(t+\tau) - \bar{y}(t+\tau) \right] \Big/ \sigma_y(t) \cdot \sigma_y(t+\tau) \qquad (4)$$

where yi(t) and yi(t+τ), i=1,N, are the sampled aplitudes of two A-scans, each consisting of N pixels yi, collected at times t and t+τ, respectively; $\bar{y}(t)$ and $\sigma y(t)$ represent the mean and standard deviation of the discrete representation of the A-scan y(t).

R depends monotonically on the rate of movement, such that R (t=0)=0 (stationary case) and R (t>0)>0 (4).

Digitised M-mode displays obtained experimentally, as well as the simulated images, were analysed as follows. The coefficient R', as defined in equation (4), was calculated between gated portions of pairs of scans, corresponding to depth in tissue of d-1.28cm, separated by t=T/10, where T is the period of oscillation. The obtained correlation patterns, i.e. variations of the coefficient R' as a function of time, were normalised to the length of two consecutive cycles of movement, in order to facilitate comparison.

An example, calculated for experimental and simulated movement along the ultrasonic beam, is shown in figure 6. In this example, correlation patterns were calculated at a distance close to the source of movement (fig. 7.1a, 7.2a) and at a farther distance (fig. 7.1b, 7.2b).

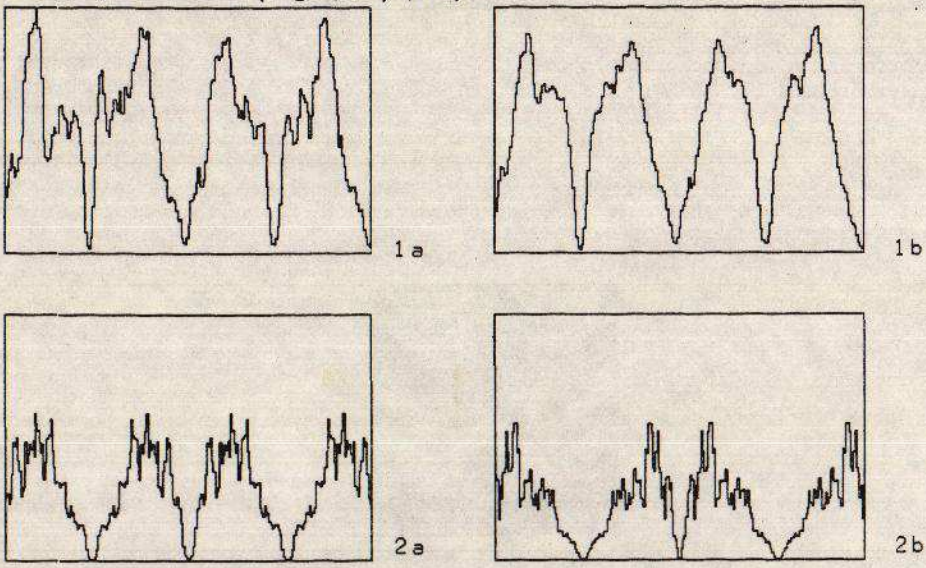

Figure 7. Patterns of correlation (movement along the beam):
1a, 1b - experiment, a, 2b - simulation;

Vertical axes: correlation (arbitrary units, same scale);
Horizontal axes: time (normalised to two periods).

DISCUSSION AND CONCLUSION

Preliminary results of M-mode simulation indication, that, for a simple oscillatory movement and a random scattering medium, a fair degree of similiarity between simulated and experimental results is obtained, assuming linear wave propagation in an elastic medium. The pattern of waveforms in sumulated images shows the right kind of 'trend' (allowing for imprecision of such description) for both axial and lateral displacement, when compared with experimental displays. This is further confirmed in statistical analysis,

where patterns of correlation computed for experimental and simulated data reveal a similiarity of structure, corresponding to maxima in the rate of movement. The differences, however, remain significant; these can be attributed, firstly, to oversimplifications in theoretical modelling, and secondly, to lack of proper control in laboratory experiments. These factors are currently being taken into account; the results from phantom studies are used to derive better theoretical models and, vice versa, more correct modelling offers suggestions as to better control of experimental studies.

Comparison of experimental results of ultrasonic movement analysis in tissue phantoms with computer simulation of this phenomenon offers a means of critical assessment of the method, while providing a physical basis for understanding of the observed regularities. The theoretical model is being developed and the next stage of simulation will include analysis of movement in three dimensions, as well as movement originating from more than one source (e.g. in liver, motion results from the direct 'shock wave' from the heart, followed by the pressure pulse from the aorta). On the basis of the present preliminary analysis it can be predicted that computer simulation of tissue movement, as seen by ultrasound, will provide results useful in clinical applications of this method and will serve a role in explaining the physiology of soft tissue movement in vivo.

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