CHAPTER IV
CHARACTERIZATION OF BONE MATERIALS AS ULTRASONIC TRANSDUCER

IV.1 INTRODUCTION

The biological significance of piezoelectricity has been described with its relevance to bone-growth and remodelling in Chapters I, II and III. The demonstration of piezoelectric effect in 'in vivo' bone has also been performed with the help of coupled transducer (Chapter III) which is suggestive of the fact that 'in vivo' bone is capable of generating electric current in response to mechanical waves. Observations in Chapter II show that physical (bioelectrical) parameters are related to physiological processes and play an important role in controlling bone behaviour. In this connection, a significant amount of data has been obtained on various electrical and mechanical properties in 'in vitro' bone, though the mechanisms of underlying phenomena still remain obscure. Mechanical properties have a definite bias towards estimating the stiffness of bone and its application in understanding of locomotion under stress (Bonfield and Dutta, 1974; Yoon and Katz, 1976; Maeda et al. 1976; Berme et al. 1977) while bioelectrical activities under dc and low-frequency stimulation have been studied largely to understand remodelling and growth pattern in bone as in
Chapter II (Hassler et al. 1977; Klapper and Stellard, 1974; Behari et al. 1979; Behari and Andrabi, 1980; 1981). However, it still remains a major task to fit in these data into a single phenomenon pertaining to the overall bone behaviour.

Piezoelectricity in bone is suggestive of the fact that it may behave as ultrasonic transducers and respond to dynamic loading. A systematic survey of piezoelectricity in bone was initiated by Fukada and Yasuda (1957), wherein, the value of electromechanical coupling coefficient 'k', charge constant 'd' and voltage constant 'g' are estimated under dynamic loading. Since then many other investigators have reported on the static as well as dynamic piezoelectricity in bone (Cochran et al. 1968; Cochran, 1974; Marvin et al. 1980; Kulas and Saha, 1978; Korostoff, 1979; Pfeiffer, 1977) and have emphasized its importance in bone electrical activities (Bassett, 1968; Gjelsvik, 1973; Güzelsu, 1978; Korostoff, 1979). However, the data on various physical, dielectric, piezoelectric and electromechanical parameters needed for characterization of bone materials as ultrasonic transducer are largely missing and poorly understood. Also, there is scarcity of data on various physical parameters in high-frequency range ( > 100 KHz).

An additional support to the study of bone behaviour in high-frequency range is obtained from the fact that ultrasound has found increasing use in medical diagnosis
and therapy (Lutz and Petzold, 1976; Woodcock, 1980; Wilson and Hood, 1980; Wells, 1980; Schwan, 1980; Lelé, 1980; Fry and Stephan, 1980; Filipczynski, 1978; Chenani and Guha, 1980; Greguss and Berloynyi, 1976). Probably an impetus to medical applications of ultrasound was obtained from the fact that transducers in the frequency range (1-10 MHz) are available possessing wide power handling capacity. However, no serious prima facie consideration seems to have been given to the role of high-frequency ultrasound in the control of biological systems. This may be attributed to the non-availability of high-frequency biocompatible transducer for biological use and the consideration that high-frequency (>10 MHz) ultrasound may not have much penetration because of high absorption at these frequencies. A number of piezoelectric transducers are available in the frequency around 1 MHz but as the frequency is approached in the vicinity of 10 MHz, their response shows a sharp decline and hence are not of much use at higher frequencies. With a view to bridge this gap and to throw open the possible role of high-frequency ultrasound in biological processes, an attempt is made to identify bone materials, e.g. bone and its two major components collagen and apatite as ultrasonic transducers by measuring various physical (resistivity, density, Curie-temperature), dielectric (dielectric constant, dielectric-loss), piezoelectric (charge-constant and
voltage constant) and electromechanical (frequency constant and coupling coefficient) parameters. These measurements are based on standard techniques (Zaffe et al. 1971). For electrical characterization of these materials in high-frequency range (1-70 MHz), variation of resistivity (ohm-cm), dielectric constant ($K_T$), inductance (L) and quality factor 'Q' have been examined. To have an assessment of this material, an ultrasonic as transducer, a test piece has been mounted in a shielded metal case with proper backing and loading media to obtain it in standard configuration. Thus obtained bone-transducer has been used to study its behaviour in high-frequency range, by examining the variation of electrical impedance, phase-angle and relative voltage (0.5-108 MHz). From the impedance phase-angle measurements and the information regarding the frequency response curve of bone, it has been observed that it has sufficient transduction capacity of electrical power in the frequency range ($>10$ MHz). The peaks in the frequency-response curve are found to fall around 56, 112 and 168 MHz, indicating the presence of first, second and third harmonic respectively. For the purpose of operating the bone transducer in high-frequency, it has been used to study velocity dispersion in bone materials in the frequency-range (1-108 MHz). Obtaining the crystal into standard transducer shape helps in drawing an equivalent electrical circuit and provides a quan-
tative comparison of parameters with other known Ceramics and quartz transducers. With the knowledge of resonant frequency, charge-constant and density, the parameters of equivalent electrical circuit have been calculated. It is hoped that these data will find use in explaining the mechanism contributing to bone piezoelectric behaviour and related phenomena in bone-growth and remodelling.

IV.2 MATERIALS AND METHODS

IV.2.1 Specimen Preparation

Experiments have been performed on human and animal bones (goats, pigs, bovine and buffaloes). Collagen and apatite have been obtained from full bone adopting the standard techniques as mentioned elsewhere (Becker and Brown, 1965). Bone materials so obtained, have been used to prepare test pieces in standard shape. The test pieces were washed in running tap water and then dried at room temperature. The test pieces of the materials obtained were having the dimensions: diameter = 10.0 mm and thickness = 1.0 ± 0.01 mm. Test pieces are provided electrical contact on either sides by painting colloidal silver paste.

IV.2.2 Poling

Poling has been done by applying a strong electric field to the electroded test pieces. To establish desired properties under optimum poling conditions for
bone and its two major constituents, electric field of different strengths have been applied for stipulated period. It has been found that for optimum poling, electric field of 4.5 KV/mm, 4.0 KV/mm and 3.5 KV/mm should be applied for 45 minutes for apatite, bone and collagen respectively at an elevated temperature. Test pieces were immersed in silicon oil-bath to sustain high electric field strength applied across them.

IV.2.3 Measurements

Measurements for various physical, dielectric, piezoelectric and electromechanical properties for bone materials are carried out using standard techniques (Measurements of piezoelectric ceramics, IRE techniques, 1961). The variation of impedance, phase-angle and relative voltage with frequency in range (0.5-108 MHz) is measured. Also the behaviour of quality factor 'Q', dielectric constant (K_T), resistivity (Ohm-cm), and inductance (L) in the frequency range (1-70 MHz) for bone materials are observed on unpoled and poled test pieces. Measurements on poled pieces have been conducted one week after poling. This was done to ensure stabilization of the test pieces.

IV.2.3(i) Low-field Measurements

Dielectric constant (K_T), dielectric-loss (tan δ) and resistivity (ohm-cm) measurements have been performed on the unpoled and poled electroded test pieces of
bone, collagen and apatite with the help of radiofrequency bridge (Model B601, Wynekerr, England) at 1 KHz. Voltage constant (g) has been measured by 'drop-test' method. In this, a known mass is dropped from a definite height on the test piece which is connected to the electrometer (Model 616, Keithley Digital Electrometer) to measure the voltage generated. Voltage constant (g) is calculated by taking the ratio of the product of voltage and thickness of the sample to the force at the time of striking it. Charge constant (d) has been calculated from the measured voltage constant and free dielectric constant using the following formula:

\[ g = \frac{d}{k_0 K^T} \]

where, \( k_0 \) is the permittivity of the free space.

IV.2.3(ii) **Measurements with Temperature**

Variation of capacitance and dielectric-loss (\( \tan \delta \)) with temperature has been measured by immersing the mounted test pieces in silicon oil. The oil-bath has been heated to various temperatures in range (25-225°C). The test piece is connected to the radiofrequency bridge through the leads to note changes in capacitance and dielectric-loss at 1 KHz due to changing oil-bath temperature. The temperature corresponding to the maximum value of capacitance is noted separately for unpoled and poled pieces of bone, collagen and apatite and is termed as Curie-temperature for these materials.
IV.2.3(iii) High-Frequency Measurements

High-frequency measurements consist of two parts. In the first phase, electrical properties of bone materials and PCB-6 have been measured in the frequency range (1-70 MHz). Second part deals with the performance of standard bone transducer in the frequency range (0.5-106 MHz). Standard bone transducer has been designed taking into consideration its characteristic impedance at the resonance frequency. The choice of backing and loading media are governed by their characteristic impedances so as to have minimum reflection and maximum transmission.

PART I

Measurements of variation of electrical properties i.e. quality factor, resistivity and dielectric constant for bone materials and the corresponding parameters for PCB-6 have been performed on Q meter (Model 4342A, H.P). Electroded test pieces were connected to Q meter through wire contacts. Zero adjustment operation for measurements of above mentioned parameters has been done adopting standard method.

The calculation of coupling coefficient 'k' has been performed by finding the values of clamped dielectric constant \( K^S \) and free dielectric constant \( K^T \) with the help of the formula:

\[
K^S = K^T (1-k^2)
\]
PART II

This part deals with the operational assessment of bone transducer (Lootronix, India) in the standard configuration in the frequency range (0.5-108 MHz). Bone transducer has been designed by mounting it in a shielded metal case and with due consideration to its characteristic impedance \( Z \) and quality factor 'Q' in the frequency range of operation. The transducer has been connected to measuring apparatus through a standard 50 ohm coaxial wire having B and C connectors. Measurements of variation of electrical impedance and phase-angle of bone transducer in frequency range (0.5-108 MHz) has been done with the help of RF vector impedance meter (Model 4815 A, H.P). The apparatus displays both electrical impedance and phase-angle simultaneously. Frequency of choice has been adjusted on the frequency panel before observations are taken. Frequency corresponding to minimum impedance and reversal in polarity has been noted and is termed as resonant frequency. The product of the resonant frequency and thickness of the test pieces used in characterizing standard transducer is termed as frequency constant of the bone materials. The determination of the frequency response of the bone transducer and piezoelectric ceramic transducer (PCB-6, Piezoelectric Ceramics, India) has been carried out by vector voltmeter (Model 4805A, H.P) and signal generator (Model 8601A, H.P). Two identical
transducers acting as transmitting and receiving systems are coupled directly with the help of a coupling medium. The variation of relative voltage has been measured on vector voltmeter with change of frequency of the exciting signal obtained from standard signal generator. The ratio of the direct signal to the one through the transmitting and receiving system has been measured for various frequency in the range (0.5-108 MHz). It has been found that there are distinct peaks corresponding to frequencies at 56, 112 and 168 MHz. An independent check to the peak positions has been done by connecting the receiving transducer to the dual-beam oscilloscope (Model R7844, Tektronix) and to the spectrum analyser (Model 8565A, H.P). It has been found that for this experimental set up, peaks correspond to the positions as mentioned above. The same experiments have been repeated in identical manner with FCB-6 transducer to have a comparison with these observations.

PART III

The measurements of the ultrasonic velocity in the frequency range (1-108 MHz) is based on the transmission technique using two identical transducers coupled to either sides of the samples of bone, collagen and apatite of various thickness, as mentioned in Chapter III.
IV.3 RESULTS AND DISCUSSION

Table IV.1 shows the data for bone materials to characterize their solid state: density (kg/m$^3$), resistivity (\(\text{ohtm-cm}\)), dielectric constant (\(K_\text{T}\)), dielectric-loss (tan \(\delta\)), charge constant (\(d\)), voltage constant (\(g\)), electromechanical coupling coefficient (\(k\)), frequency constant (\(N_\text{T}\)) and Curie-temperature (\(^\circ\text{C}\)). These parameters have been measured by the methods mentioned above. These data extended for more than fifty samples, show an average of the calculated values and are in good agreement with the reported results (Maeda et al. 1976; Yasuda and Fukada, 1957; Shames and Lavine, 1964). This confirms that bone materials under study are similar to those of the materials used by other workers. After ascertaining the characterization of the physical properties, behaviour of these materials is probed in the high-frequency region. Also, the temperature dependance of capacitance and dielectric-loss have been performed and results are presented below.

IV.3.1 Physical Properties

From Table IV.1, it is clear that collagen is found to have lowest while apatite has highest values for resistivity and dielectric constant. For unpoled pieces of bone materials, resistivity is found in the range $10^7$-$10^9$ \(\text{ohtm-cm}\) and increases by a factor of $10^2$-$10^4$ for poled bone materials. The details of poling conditions
<table>
<thead>
<tr>
<th>PARAMETERS</th>
<th>UNPOLED PIECES AT ROOM TEMPERATURE (30°C)</th>
<th>POLED PIECES WITH CONDITIONS OF POLING</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>BONE</td>
<td>APATITE</td>
</tr>
<tr>
<td>Density (Kg/m³)</td>
<td>1824</td>
<td>1894</td>
</tr>
<tr>
<td>Resistivity (Ω·cm)</td>
<td>~10^8</td>
<td>~10^9</td>
</tr>
<tr>
<td>Dissipation-factor (tan δ)</td>
<td>0.042</td>
<td>0.052</td>
</tr>
<tr>
<td>Charge constant 'd' (C/N) 10^{-12}</td>
<td>2.63</td>
<td>3.36</td>
</tr>
<tr>
<td>Voltage constant 'g' (V·m/N) 10^{-3}</td>
<td>10.25</td>
<td>11.88</td>
</tr>
<tr>
<td>Dielectric constant (K')</td>
<td>29</td>
<td>32</td>
</tr>
<tr>
<td>Frequency constant (Hz·meter) 10^4</td>
<td>5</td>
<td>4.5</td>
</tr>
<tr>
<td>Coupling coefficient (k)</td>
<td>0.42</td>
<td>0.38</td>
</tr>
<tr>
<td>Curie-Temperature (°C)</td>
<td>120</td>
<td>145</td>
</tr>
</tbody>
</table>
are given in Table IV.1. It has been found that resistivity value is sensitive to the method of preparation and state of hydration. Out of large number of observations on test pieces, only those values have been considered which fall in the particular region. These data are consistent with the reported results (Shamos and Lavine, 1964). The value of density for bone, collagen and apatite lies in the range 1824-1897 kg/m³ and there has been observed no apparent change in it after test pieces are poled. In our case, density for apatite is found slightly lower than what it has been reported (Gilmore and Katz, 1968). This discrepancy in density may be attributed to the presence of pores in the sample which are introduced during the extraction of apatite from compact bone as also observed by Katz (1971).

From Fig. IV.1, it is clear that temperature corresponding to maximum value of capacitance are 120°C, 95°C and 145°C for bone, collagen and apatite respectively and is referred as Curie-temperature for these materials.

IV.3.2 Dielectric Properties

The values of dielectric constant (κT) at 1 KHz for bone, collagen and apatite, are found 29, 6 and 32 respectively and do not change significantly after the samples are poled. The present values of dielectric constant are higher than reported values (Shamos and
Capacitance vs temperature for Bone, apatite and collagen in disc shape of dimension: dia=10 mm and thickness = 0 mm.

- Apatite
- Bone
- Collagen

Fig. IV.1
Lavine, 1968). The difference in the value of dielectric constant may be due to frequency at which it has been measured. It may be pointed out that dielectric constant decreases with frequency (Geddes, 1972)

The variation of capacitance and loss-factor with temperature (25-225°C) are presented in Fig. IV.1 and Fig. IV.2, respectively. From Fig. IV.1, it is clear that capacitance for apatite rises to a peak value of 440 pF corresponding to a temperature of 145°C and thereafter a sharp decline follows. Identical behaviour is observed for all the three materials. The peak position shifts towards the higher value in the order collagen, bone and apatite. Corresponding to these, highest value of capacitance has been found for apatite.

An identical pattern is also observed for tan δ vs. temperature variation for the three materials under study (Fig. IV.2). Maximum dielectric-loss has been found in the case of apatite followed by bone and collagen (Table IV.1). Our earlier investigations (Behari et al. 1974) lead us to conclude that conductivity increases with the rise of temperature. An early increase in loss-factor (tan δ = ω CR) with temperature may be interpreted by saying that increase in 'C' is faster than a decrease in 'R'. At a temperature near 100°C, the difference in their rate of variation is maximum and thereafter 'C' decreases while 'R' becomes almost constant for the rise of temperature under investigation. As
Loss-factor ($\tan \delta$) vs temp. for Bone, apatite and collagen in disc shape of dimension: dia=40 mm, thickness=1.0 mm.

--- Collagen
--- Apatite
--- Bone

Fig. IV.2 Temperature (°C)
explained earlier (Behari et al. 1974), a decrease in resistivity is due to an additional number of protons available for charge transport. It may be suggested that at Curie-temperature this phenomena is at its optimum and dipole layers formed in the process (of charge migration) starts contributing to the increased capacitance. The maximum value of capacitance at Curie-temperature for apatite follows from the fact that out of the three materials under investigation, its resistivity is highest. In terms of the molecular structure of the collagen, it may be stated that it has the lowest resistivity which accounts for its low value of dielectric constant and dielectric-loss among all the three materials (Table IV.1, Figs. IV.1 and IV.2).

IV.3.3 Piezoelectric Properties

The measured values of voltage-constant have been found 10.25, 15.7 and 11.88 and 13.80, 18.80 and 13.38 (V-m/N) x 10^{-3} for unpoled and poled bone, collagen and apatite respectively. The corresponding values of charge-constant are 2.63, 0.84 and 3.36 and 3.54, 0.99 and 3.79 (C/N) x 10^{-12} for unpoled and poled bone, collagen and apatite respectively (Table IV.1). From these data it is inferred that bone materials have better utility as receiving-transducer material as these have higher values of voltage constant as compared to their respective charge constant. Collagen is having lowest
charge constant inspite of having highest voltage constant. This is probably due to its low dielectric constant.

Present data for charge constant are found higher by a factor of 10 for bone as compared to reported results (Fukada and Yasuda, 1957; Maeda et al. 1976). However, it is difficult to compare these results to reported findings due to the difference involved in measurement, techniques and sample preparations. Also, present observations show that piezoelectric properties for apatite are at variance with the findings of others (Liboff and Shamos, 1973; Braden et al. 1966; Shamos et al. 1963). It has been observed that apatite has the highest charge constant $3.36 \times 10^{-12}$ C/N and $3.79 \times 10^{-12}$ C/N for unpoled and poled pieces respectively which is indicative of the fact that apatite is the best possible transducer material among three bone materials e.g. bone, collagen and apatite.

The piezoelectricity in apatite may be attributed to the fact that asymmetry in crystal is produced during its extraction from compact bone and some inherent lattice defects. In small crystals containing a few units, a charge imbalance may exist between the columns of hydroxyl ions, oriented in any directions. As a result, the crystal as a whole may be non-centrosymmetric and thus piezoelectric (Key et al. 1964). Also, the possible electric properties of apatite must be considered
in the light of recent development in the study of crystal faults as electronic devices (Paterson, 1966). Since a number of lattice defects are supposed to be present in crystal of hydroxyapatite (Neumann, 1957), it is possible that these crystals may display electronic conductivity. Furthermore, the reported electret phenomenon in apatite provides an additional support to the observed piezoelectric effect in apatite in the present investigation (Andrabi and Behari, 1981). A similar anomalous behaviour has been observed in apatite while studying Hall-effect in bone materials. Hall-voltage for apatite has been found to vary with magnetic field while it is not so for bone and collagen under identical conditions (Andrabi and Behari, 1979).

IV.3.4 High-Frequency Measurements

PART I

To study the electrical behaviour of bone-materials, in the high-frequency range, the variation of resistivity, capacitance, inductance and the related 'Q' factor have been examined (1-70 MHz). Figs. IV.3 and IV.4 depict these observations for bone materials and PCB-6. From Fig. IV.4 it is clear that the slope for variation of dielectric constant with frequency is least for collagen while it is maximum for apatite and is intermediate for bone. This may be attributed to relatively larger symmetry in the structure at the molecular level.
Resistivity vs frequency for apatite, bone and collagen. Inductance vs frequency for PCB-6 in disc shape of dimensions; dia = 10.00 mm and thickness = 1.0 mm.

- Bone
- Apatite
- Collagen
- PCB-6

Fig. IV. 3
Dielectric-constant and quality factor vs frequency for Apatite, bone, collagen and PCB-6 in disc shape of dimensions: dia=10.00 mm and thickness=1.0 mm.

Fig. IV. 4
in collagen resulting in the lesser number of free dipoles. In apatite there are voids in it giving rise to the existence of ion-vacancies and offering sites for impurities. This leads to higher dependence of dielectric constant on frequency for apatite. An intermediate state can be visualized to exist in bone. As referred in Section IV.3.2 that dielectric constant decreases with increasing exciting signal frequency. This may be suggested due to relaxation-phenomenon (Schwan, 1968). However, for good insulator, dielectric constant does not vary significantly from dc to microwave frequency. In the case of piezoelectric materials, there is strong dependence of dielectric constant on frequency (Zaffe et al. 1971).

**PART II**

Variation of phase-angle with frequency in the range (0.5-108 MHz) is depicted in Fig. IV.5. It can be seen that phase-angle is around -90° over a frequency range (0.5-56 MHz). In proximity to 56 MHz, there is reversal in the polarity and hereafter, the phase-angle approaches fast towards +90°. It is reflected in Fig. IV.5. From the observations of impedance (Fig.6) and phase-angle taken on RF vector impedance meter it may be concluded that the material is highly capacitive. Owing to its capacitive nature, the impedance variation with frequency is strongly dependent upon the method of
Phase-angle (tan\(\lambda\)) vs frequency for Bone, apatite, collagen & PCB-6 in disc shape of dimension: dia = 10 mm, thickness = 1.0 mm.

---

Fig IV.5
Impedance vs frequency for Bone, apatite, collagen and PCB-6 in disc shape of dimensions:

dia=10.00 mm and thickness=1.0 mm.

\[
(Z)\min (\text{Bone}) = 40 \, \text{mS}
\]
\[
(Z)\min (\text{Collagen}) = 7 \, \text{mS}
\]
\[
(Z)\min (\text{Apatite}) = 15 \, \text{mS}
\]

---

**FIG. IV6**

Impedance vs frequency graph for Bone, Apatite, Collagen, and PCB-6.
its mounting in an electrical circuit. An independent support of its capacitive nature is provided by 'Q' measurements. Capacitance of bone samples, obtained with 'Q' meter using a standard inductor yields $C_1 > C_2$. On the other hand, PCB-6 data reveals that its phase-angle is always positive and hence behaves as inductor in the region of its applicability. Frequency response curve for bone materials (Fig. IV.7) shows that peak positions correspond to frequencies at 56, 112 and 168 MHz indicating first, second and third harmonic. Similar results have been observed in the variation of electrical impedance with frequency (Fig. IV.6). In view of the reported values of the elastic constant (Lang, 1970; Subyoon and Katz, 1979), using about 5 MHz ultrasonic pulse, it may be suggested that the phenomenon under observation is related to the microscopic nature of the material and the resonance peak positions of the crystal are the excited states of the molecules therein. Impedance minima correspond to 56, 112 and 168 MHz. It may be mentioned that relaxation processes in vivo bone system fall in the MHz region (Bchari and Singh, 1981; Coakely and Dunn, 1972; O'Brien, 1972). In dry bone, besides being loss of water several additional factors come into picture, e.g. formation of dipoles, existence of interstitial-vacancies combination and impurities migration in response to ultrasonic field (Lakes and Katz, 1979). It may be suggested that these factors
Relative amplitude vs Frequency for Bone, apatite, Collagen and PCB-6, in disc shape of diameter = 10.0 mm and thickness = 1.0 m.

- --- Bone
- --- Collagen
- --- Apatite
- --- PCB-6
pertaining to the dry bone are responsible for the shift in resonant peak towards the high frequency side. Hence, it may be concluded that resonance peak positions of the crystal is related to the properties of the materials at the molecular level.

An independent corollary of above is contained in the velocity-dispersion curve (Fig. IV.8). Apatite being the most stiff out of the three shows least dispersion. The collagen molecules being the most elastic, there is probability of generation of acoustic modes of large amplitudes and hence the mutual coupling coefficient is maximum in this case as also proposed earlier (Lakes and Katz, 1979; Lakes et al. 1977; 1979). This leads to higher scattering of ultrasound with the excited modes which result into velocity dispersion. These findings are consistent with the reported results (Subyoon and Katz, 1979), wherein, bone has been found to behave as viscoelastic medium in the frequency range (1-10 MHz). Furthermore, in the frequency range (40-108 MHz), the curve for ultrasonic velocity in collagen vs frequency is almost linear. A similar trend is observed in the bone in the frequency region extending from 20 to 80 MHz. For apatite, the slope is negligibly small as compared to the other two materials. For bone velocity increases from 4725 m/s at 1 MHz to 9235 m/s at 108 MHz while for collagen it varies from 2273 m/s at 1 MHz to 9073 m/s at 108 MHz. For apatite the variation is almost
Ultrasonic velocity vs frequency for Bone, collagen and apatite.

- --- Bone
- Apatite
- Collagen

Fig. IV.8

Ultrasonic velocity (m/sec.)

Frequency (MHz)

2000  4000  6000  8000  10000
2  4  8  16  32  64  128  256  512  1024
small. This follows from the fact that bone is a viscoelastic material and so are its two major constituents. In the bone, the stress is distributed non-uniformly over the entire medium owing to its biphasic and anisotropic structure. Since collagen has a lower modulus of elasticity, the strain is more effective in this medium and dispersion curve shows a higher slope in it. For the same reason, in apatite the slope is less while for bone it lies in between the two (Fig. IV.8).

IV.3.5 Equivalent Electrical Circuit Parameters of Bone Transducer

The electrical characteristics of transducer materials are influenced by their mechanical properties and by the type of mechanical loading to which they are subjected. These quantities may be represented by an equivalent circuit of the transducer. It is assumed that strain-stress curve is linear and radiation takes place from only one face (air packed) and the impressed stress is in phase with the impressed voltage. Under these assumptions, ultrasonic transducer may be represented by an equivalent electronic circuit of four terminals network with two electrical input and two mechanical output as shown in Fig. IV.9a and IV.9b. The values of the various constants of equivalent electrical circuit have been calculated for bone using the standard formulae (Kinsley and Fry, 1962), and are presented along with those for PCB-6 and quartz in Table IV.2.
Fig(IV.9.a) The four terminal network representing the equivalent circuit of a transducer.

Fig(IV.9.b) The equivalent electrical circuit for a piezoelectric plate transducer operating near resonance where mass, elastic compliance and mechanical damping are transformed into equivalent electrical elements \( L, C, R \) by piezoelectric effect. \( C_0 \) is the capacitance of the transducer at the resonant frequency under static conditions.
<table>
<thead>
<tr>
<th>No.</th>
<th>Parameters</th>
<th>PCB-6</th>
<th>Quartz</th>
<th>Bone</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>$C_0 \ (pF)$</td>
<td>1042</td>
<td>3.15</td>
<td>20.2</td>
</tr>
<tr>
<td>2</td>
<td>$C \ (pF)$</td>
<td>778</td>
<td>0.0268</td>
<td>10.0</td>
</tr>
<tr>
<td>3</td>
<td>$R \ (\Omega)$</td>
<td>152</td>
<td>4750</td>
<td>6.2</td>
</tr>
<tr>
<td>4</td>
<td>$L \ (Henery)$</td>
<td>$20 \times 10^{-6}$</td>
<td>457</td>
<td>$0.84 \times 10^{-6}$</td>
</tr>
<tr>
<td>5</td>
<td>$f \ (Hz)$</td>
<td>$1.27 \times 10^6$</td>
<td>$45.5 \times 10^3$</td>
<td>$56 \times 10^6$</td>
</tr>
<tr>
<td>6</td>
<td>$C_E \ (N/m^2) \times 10^{10}$</td>
<td>8.72</td>
<td>86.05</td>
<td>128.8</td>
</tr>
<tr>
<td>7</td>
<td>$e \ (C/m^2)$</td>
<td>24.39</td>
<td>0.17</td>
<td>59.39</td>
</tr>
<tr>
<td>8</td>
<td>$\varepsilon \ (C/m)$</td>
<td>3.84</td>
<td>0.03</td>
<td>9.35</td>
</tr>
<tr>
<td>9</td>
<td>$(LC) \times 10^{-12}$</td>
<td>$15.6 \times 10^{-3}$</td>
<td>12.24</td>
<td>$8.4 \times 10^{-6}$</td>
</tr>
</tbody>
</table>
Fig. IV.9a and Fig. IV.9b show the four terminals network for bone transducers wherein capacitor $C_0$ is connected in parallel with a series circuit containing resistance, inductance and capacitance. 'C' is defined as the capacitance of the ultrasonic transducer at resonant frequency under static conditions, whereas $L$, $C$ and $R$ are equivalent electrical constants to mass, elastic compliance and mechanical damping respectively under dynamic state. This transformation of mechanical constants into electrical constants is mediated through piezoelectric effect. From Table IV.2, it is apparent that PCB-6 has highest dielectric constant among the known materials used as transducers. However, impedance is maximum for Quartz. The product of inductance and capacitance is less for bone by a factor of $10^6$ and $10^3$ as compared to quartz and PCB-6 respectively. This corresponds to the fact that resonance frequency is highest for bone. At this frequency (56 MHz) bone transducer delivers power larger by a factor of $10^3$ as compared to other conventional transducers.

The information regarding the performance of the transducer can be obtained with the knowledge of $C/C_0$ and piezoelectric stress coefficient ($e$). The ratio $C/C_0$ is independent of crystal dimensions and modes of vibration and is related to electromechanical activity by the following relation:

$$\frac{C}{C_0} = \frac{8k^2}{\pi^2(1-k^2)}$$
where \( k \) is the electromechanical coupling coefficient and its value is presented in Table IV.2. The values of ratio \( C/C_0 \) for various materials are given below:

\[
\frac{C}{C_0} \text{ Quartz} = .008 \\
\frac{C}{C_0} \text{ PCB-6} = .756 \\
\frac{C}{C_0} \text{ Bone} = .494
\]

From this it may be concluded that coupling coefficient for bone materials falls in between those of PCB-6 and quartz. The value of the piezoelectric stress coefficient \( 'e' \) for various piezoelectric materials provides an estimate regarding the input power required to produce the same acoustic output for different materials. From Table IV.2 it is clear that piezoelectric stress coefficient \( 'e' \) for bone is higher as compared to quartz is indicative of the fact that less input power will be needed in case of bone to produce the same output as in quartz.