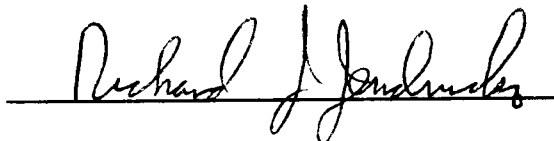


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DETERMINATION OF POTENTIAL NOISE SOURCES IN THE
INSTRUMENTATION USED FOR MEASUREMENT OF THE
PIEZOELECTRIC EFFECT IN BONE

A Thesis
Presented for the
Master of Science
Degree
The University of Tennessee, Knoxville

Gary Lynn Chamberlain

March 1986

ABSTRACT

Measurement of the electrical properties of mechanically stressed bone has been done since 1957. The inconsistency of the results thus far, however, indicate a need not only for the development of a more refined method of measuring these properties, but also for determination of potential noise sources which can mask the desired signals. In the past, three amplification methods have been used to measure the electrical response of bone: current amplification, voltage amplification, and charge amplification. Of the three, the voltage amplifier has been the most popular, mainly because of its ease of use. In an earlier thesis, Simpson (1980) studied the properties of all three types of amplifiers and then developed voltage and current amplifiers suitable for the task of measuring the electrical response of mechanically stressed bone.

For this study, prototypes were built and tested, and a suitable system was developed to incorporate both a current and voltage amplifier for the measurement and study of the electrical response of mechanically stressed bone samples subject to cantilever beam bending. The impact loading system used a weighted plunger to provide end loading of the bone samples. This system was also able to measure the mechanical deflection of the bone directly using an optical probe, allowing for the study of the correlation of voltage

and current generation with the mechanical deflection of the bone. The results were recorded and studied on a Hewlett-Packard 5420B Digital Signal Analyzer. Testing was conducted to determine potential noise sources, isolate the causes, and reduce or eliminate them.

Three sources of noise were found: 1) 60 Hz, 2) static electricity, and 3) mechanical vibration. Grounding removed the 60 Hz noise. Static electricity and mechanically induced vibration produced signals that were typical of the signals produced by normal bending of bone.

Static electricity was produced by motion of the plunger within the plunger guide just before impact with the bone samples. Static electricity may be reduced by avoiding the use of sliding surfaces for plunger guides near the bone samples.

Vibration induced signals were caused mostly by movement of the cable between the electrode leads from the bone sample and the inputs to the amplifiers. It was found that these noise signals can be reduced by designing a more rigid mounting for the bone samples, and either reducing the length of the cabling between the bone samples and amplifier or designing a "hybrid" amplifier that can be mounted adjacent to the bone samples.

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CHAPTER I

INTRODUCTION

The need for the detection of small signals (such as those produced by mechanically stressed bone) in high impedance circuits has made it necessary to develop improved measurement methods. Two measurement systems, (voltage and current) were studied for their susceptibility to electrical noise, mechanical vibration, and static charge. Measurement of piezoelectric response in bone was chosen as an example because of the difficulty of measurement and the interest among many researchers in studying this phenomenon.

The fact that mechanically stressed bone would produce an electrical response was first introduced by Yasuda (1954), with later work (Fukada and Yasuda, 1957) indicating that the observed signals being produced were very similar in nature to signals produced by a classical piezoelectric effect similar to many crystalline materials (Cady, 1964). Continued study further defined the electrical properties associated with mechanical stress in bone.

Once the electrical response had been established, research branched out in two primary directions involving: (1) the measurement of electrical response on mechanically stressed bone; and (2) the use of electrical stimulation in live animals to enhance clinical treatment of bone injuries (i.e. stimulation of bone growth and healing). Work in the

last ten years in particular has centered around electrical stimulation as a clinical tool (Friedenberg et al., 1971; McElhany et al., 1968) and in vivo studies of whole bone subjected to externally applied electrical fields (McElhany et al., 1968). The greatest benefit has been the promotion of bone growth in the treatment of non-unions (Friedenberg et al., 1971). Also being tested are the use of electrical stimulation to promote more rapid ingrowth of new bone into porous prosthetic implants (Weinstein et al., 1976), the use of porous prosthetics made of piezoelectric material (Park et al., 1980, 1981), and the use of porous prosthetics with a permanent (electret) charge to encourage bone ingrowth. While considerable research has been done to determine the characteristics of the electrical response of mechanically stressed bone, a clear, consistent model of the process has not yet been presented. Before efficient clinical applications can be developed, this effect should be more thoroughly understood.

There are three major discrepancies in the literature on this subject. These are: (1) moisture and humidity alter the magnitude of the piezoelectric effect (Erikson, 1976), (2) whether the signal is a monophasic or a biphasic signal (Cochran et al., 1968; Shamos and Lavine, 1964; Williams and Breger, 1975), and (3) whether there is a sustained low level D.C. current (Bassett and Becker, 1962; Cochran et al., 1968; Harms et al., 1970; Steinberg et al.,

1968; Williams and Breger, 1975). Improved measurement methods can help answer one or more of these questions.

Earlier work on instrumentation amplifiers by Simpson (1980) produced a detailed theoretical characterization of the two transduction methods currently used, voltage and charge amplifiers, as well as a detailed study involving a current amplifier. Demonstration prototypes of the voltage and current amplifiers were then developed and built, to overcome many of the problems associated with earlier systems. Furthermore, a system utilizing both amplifiers and the mechanical loading apparatus into a small signal measurement system to test bone samples was designed and constructed. Performance of the developed system was evaluated and significant limitations were identified.

CHAPTER II

LITERATURE REVIEW

The following is a summary of research done on measuring the electrical response of mechanically stressed bone and a review of some examples of clinical applications. The review covers experimental results to date, several theories concerning possible mechanisms of the electrical response, and theories as to its usefulness in a living organism.

I. EARLY WORK

Wolf's Law has been recognized since 1892 as an accurate description of the remodeling process of bone (Bassett, 1965). However, not until 1953 (Yasuda, 1954), was it first discovered that mechanically stressed bone would produce an electrical response. This work, along with later work by Fukada and Yasuda (1957), led these researchers to the conclusion that since this response was produced only upon the application of mechanical stress to the bone, the mechanism involved was classical piezoelectricity.

Apparently unaware of the work of Fukada and Yasuda, Bassett (in an unpublished 1956 work) also observed the piezoelectric effect, but attributed the effect to "artefact" because of the stainless steel electrodes he was

using and because he observed a similar effect in wood (Bassett, 1968). It was not until the work of Bassett and Becker in 1962 that the electrical response of bone gained attention in the United States. Bassett and Becker conducted experiments on bone strips (30 x 2 x 1.2 mm) subjected to cantilever bending loads. They found that the concave portion developed a negative charge and the convex portion developed a positive charge. Step loading produced a voltage pulse which died out, and removal of the load produced an opposite pulse of approximately equal magnitude.

Subsequent investigations (Bassett, 1968; Shamos et al., 1963) have not only verified the work of Bassett and Becker, but have also demonstrated the effect on bone taken from different animals and from different anatomical sites. Shamos and Lavine have even suggested that the piezoelectric effect is a property of biological material (1967). Furthermore, preparation of the bone samples for testing, such as oven drying (up to 129°C), air drying, freezing, aging, boiling in water, and amount of hydration did not significantly alter the nature of the response (Bur, 1976; Maeda and Fukada, 1982; Marino and Becker, 1975; Reinisch and Nowick, 1975). (However, the magnitude of the response was sensitive to the hydration state (Steinberg et al., 1968).) It became an accepted fact that mechanically stressed bone produced an electrical response, and that piezoelectricity,

if not the sole mechanism, could describe much of the observed effect (Bassett, 1968). Later work, however cast some doubt as to whether classic piezoelectricity fully describes the electrical response of bone (Anderson and Eriksson, 1970; Gross and Williams, 1982; Harms et al., 1970; Johnson et al., 1980; Reinish and Nowick, 1975).

II. RECENT WORK

In an attempt to describe more accurately the response of bone to mechanical stress, more precise and refined methods for measuring the electrical response of bone have been developed.

Three types of bone specimens have generally been used. Fukada and Yasuda used cubes of bone under uniaxial compression (Fukada and Yasuda, 1957), Bassett and Becker have used long bone strips subjected to cantilever bending (1962), and whole bones have been tested both in vivo and in vitro (Fukada and Yasuda, 1957; Harms et al., 1970; McElhaney, 1967; Shamos et al., 1963). In all cases, when a rapid impulse force was applied, a sharp voltage spike was produced. A number of factors affected the amplitude of this spike, yet results were fairly consistent. In most cases, the pulse was negative on the concave or compressive side of the bone with respect to the convex side of the bone.

Methods used for loading specimens have become more sophisticated over the years. Lever mechanisms (Fukada and Yasuda, 1957) and weights dropped from strings (Bassett and Becker, 1962) gave way to electromagnetic load cells and modified Instron Testing Devices, which allowed for precise control over the nature of the applied load and better repeatability (Bur, 1976; McElhaney, 1967). Such methods allowed for step, ramp, sinusoidal, and other types of loads to be applied to the bone so that the response of bone could be examined and studied for nonlinear, fatigue, speed of loading, and memory effects.

Considerable attention has also been given to the hydration state of the bone samples tested in vitro. Increased moisture content reduced the voltage potential (Erikson, 1976; Bur, 1976; Maeda and Fukada, 1982; Marino and Becker, 1975; Reinisch and Nowick, 1975). Since hydration has been seen to play such a significant role in the magnitude of the electrical response of bone, a great deal of research has concentrated on how moisture affects the characteristics of the piezoelectric response (Bur, 1976; Cochran et al., 1968). Early studies were performed on dry bone (Fukada and Yasuda, 1964; Fukada and Yasuda, 1957; McElhaney, 1976; Shamos and Lavine, 1964; Shamos et al., 1963). However, since bone in vivo is not dry, it was questioned whether such results were meaningful. So far this question has not been fully answered.

The question of just how the electrical response changes as a function of the hydration state is still being studied. However, the results have shown that even though the hydration level affected the results, the method used to achieve the level of hydration apparently did not. It was reported that "Samples which cannot be used immediately can be stored under controlled conditions of temperature and humidity to retain physiological status. Specimens can be partially or totally dehydrated by air or oven drying with or without the use of chemical desiccation" (Jendrucko, 1979). Even freezing had little if any effect on specimens (Marino and Becker, 1975). The only process which seemed to significantly affect the bone samples was excessive heat (greater than 120°C), which can irreversibly alter the chemical and electrical characteristics of bone tissue.

The factors that affect the electrical response of bone include moisture content, curing temperature (Bur, 1976), specimen size and shape (Bassett and Becker, 1962), electrode type and placement, rate of loading (Bur, 1976; Cochran et al., 1968), and frequency of applied force (Bur, 1976). Hence, for any research to be meaningful (and in particular, for results obtained in different laboratories to be compared), an accurate description of the testing procedure is essential.

III. INTERPRETATIONS

It is now assumed that the electrical response to mechanical stress in bone is part of a feedback mechanism to stimulate bone reformation in response to changing stress conditions. A number of mechanisms have been proposed not only for how the electrical response stimulates bone reformation, but also for how the electrical potentials are initially generated. An excellent review of this latter question has already been performed (Williams, 1974; Williams and Breger, 1975) and will only be summarized here. The most popular theories are:

1. Distortion of H-bonds in collagen and/or its cross linkages;
2. Stress-sensitive p-n junctions;
3. Reorientation of water molecules due to stress;
4. Streaming potentials;
5. Imperfections in the hydroxy-apatite mineral structure that lowers symmetry;
6. Liberation of hydrogen ions by the rupture of cross linkages;
7. Changes in stress by the spontaneous polarization of collagen.

Since none of these theories completely describes the response of bone to mechanical stress, there is still considerable question as to the exact mechanism. The

possibility exists that any or all of the above theories may contribute to the electrical response of bone.

IV. CLINICAL APPLICATIONS

One of the main characteristics of piezoelectricity is that not only does it produce an electrical response to stress, but conversely, if an electrical potential is applied, then strain (or displacement) is produced in the piezoelectric material. The biological significance of this was considered early in the initial investigations (Bassett, 1965; Bassett, 1968; Becker et al., 1964; Becker, 1961). The idea that the electrical response of mechanically stressed bone was a feedback mechanism for the remodeling of bone was being established. This led to the possibility that the converse might occur, that electrical stimulation could result in bone remodeling in vivo. This theory was first tested by implanting electrodes powered by a small battery pack (Bassett, 1968) into dog femurs. After several months, histological examination of the femurs did reveal hypertrophy at the cathode and atrophy at the anode.

Once enhancement of bone growth was demonstrated, the principle was applied to the clinical setting. Several treatment settings have been studied, but the most effective use has been in the healing of non-unions (Friedenburg, 1971). Even after many months of failure to heal, electrical stimulation can produce positive progress in

these cases. The need for more effective treatment in the clinical setting has motivated continuing research in the naturally occurring signals produced by bone in vitro.

V. PRELIMINARY DESIGN

For his thesis in 1980, Mark Simpson designed and analyzed voltage and current amplifiers for the measurement of the electrical response of bone (Simpson, 1980). His goal was to improve systems that were being used at that time for the measurement of the electrical response of bone, and his findings provide the basis for the work described here. Readers may refer to Appendix A for a summary of details of Simpson's analysis and are encouraged to refer to Simpson's work as important supplemental material to this thesis.

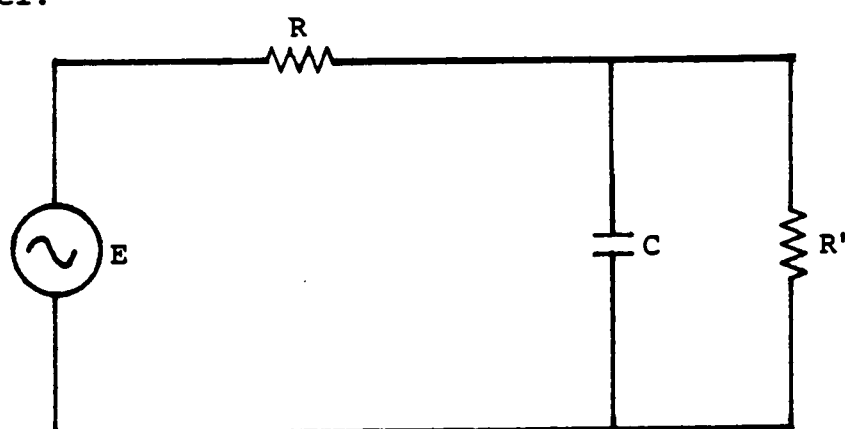
Previous investigations of the electrical response of bone to mechanical deformation have most often employed voltage or charge amplifiers to measure the small signals generated. Although these systems can be easily constructed in the laboratory or obtained commercially, they suffer a number of serious limitations in use, such as: (1) Frequency response limitations associated with the high output impedance of dry bone samples, (2) Difficulty in determining whether the signal generated in bone is monophasic or biphasic, and (3) Baseline shift limitations which occur after loading and which may be due, at least in part, to artefact.

With these limitations in mind, Simpson's goal was to develop and explore the utility of current amplification (since current amplification has improved frequency response), for characterizing strain-induced electrical signals in bone. The system he developed could measure both current and voltage so that a comparison could be made of their waveform characteristics.

For the purpose of design and analysis, mechanically stressed bone was modeled as a signal source that behaves electrically like a resistor in parallel with a capacitor (Stephan et al., 1976) (Figure 1). The bone can be considered as either a voltage source or a current source. Either source can be theoretically interchanged by representing a Norton current source as a Thevenin voltage source and visa versa. The bone was treated as a current source to simplify the subsequent analysis.

Standard models for voltage, charge, and current amplifiers were then examined in detail. A number of factors had to be considered. For a voltage amplifier, the input impedance should be an order of magnitude greater than the source impedance of the bone to reduce signal attenuation. Most investigators reported that the source impedance of bone was typically about 10^9 ohms across the electrode leads for bone samples of the size used here (Shamos et al., 1963; Stephan et al., 1976). Thus, the input impedance of the amplifier needed to be at least 10^{10} ohms.

Model:



R = Resistance of Amplifier

R' = Resistance of Bone

C = Capacitance of Bone + Amplifier

Figure 1. Electrical Bone Model.

The magnitude of voltage from test samples of bone was usually reported in the range of 2 to 37 millivolts. From both a theoretical and experimental analysis, it was found that the frequency response of the voltage amplifier was severely limited by a high impedance source with a finite value of capacitance. Therefore, the system gain was chosen to be 11 volts/volt to maximize the frequency response of the voltage amplifier. Since a high input impedance is a primary requirement, an Analog Devices AD 515 instrumentation amplifier was selected which allowed for an amplifier design with 10^{13} ohms input impedance. This more than met the 10^{10} ohm input impedance requirement and also met the other requirements for sensitivity, noise, and gain.

For a current amplifier, however, the input impedance had to be as low as possible because the current signals being measured were extremely small (typically in the range of 1 to 100 picoamps). For the current amplifier, the frequency response, unlike the voltage amplifier, is essentially independent of the value of the bone source impedance. (The calculated value of rise time for the voltage amplifier was 4 milliseconds, compared with 0.017 milliseconds for the current amplifier.) A gain of 1 volt/nanoamp was used. Again, the AD 515 was used but was configured as a current amplifier.

Once the requirements for the voltage and current amplifiers had been determined and the device meeting the

design requirements had been selected, final design and analysis were performed. More detailed performance specifications can be found in Appendix A.

A prototype of both amplifiers was built and tested. The testing was performed on human femoral bone milled into small rectangular strips 30 mm long, and about 2 to 3 mm wide and 1 to 2 mm thick. To verify the work performed previously by other researchers, the samples were dried by two different methods: oven and air drying. Three types of testing were performed: loading and unloading to examine "typical" waveforms; progressive or increased loading to determine some degree of linearity; and fatigue loading to look at any memory effects or deviation from expected results as compared to a single test.

Cantilever beam bending was used to test the bone samples. In this configuration, a bone sample was held firmly in a clamp at one end, and the other end was either suddenly or slowly loaded with a weight, depending on what type of behavior was to be measured. The weight on the end of the bone sample will bend the bone just enough to produce the expected waveforms but not enough to fracture or change the structure of the bone. Electrodes were placed on the top and bottom surfaces of the bone and connected to the amplifier to measure the voltage (or current) produced by the bone. Since it is necessary to prevent leakage of the signal from the bone to ground, the

holding clamp for the bone is made of plexiglas, a very good insulator.

The hardware was then tested. Voltage waveforms were typical of those obtained elsewhere in the literature. When loaded, the convex portion of the bone became positive and the concave portion became negative with respect to each other; removal of the load produced the opposite effect. Current waveforms produced both positive and negative waveforms and showed a much higher frequency content in the signals than was expected.

Fatigue loading showed little effect on the signals. Similarly the drying method had little if any effect on the signals (as had been reported elsewhere). (Moisture content, however did tend to reduce the magnitude of the signal.)

The experimental results can be summarized as follows:

- 1) The voltage waveform and magnitude were typical of results obtained from other researchers, and the signals were well within the range of the equipment used to measure them.

- 2) As predicted, the current amplifier displayed a much higher frequency response than the voltage amplifier. This should contribute to an understanding of the electrical response of bone.

- 3) As predicted from piezoelectric theory, magnitudes of the signals for both voltage and current increased as the

applied load was increased, adding evidence that the signals are not artefact.

4) Negative currents from the bone were detected upon removal of the load, which indicated that the signal was biphasic and that the pulse observed due to removal of the load from the bone was not artefact.

5) Within the resolution of the current amplifier used, no low level D.C. currents were detected upon sustained loading.

In view of these findings on the electrical response of bone to mechanical stress, in particular the signal detail provided by current measurements, it was suggested that in subsequent research both voltage and current waveforms should be measured in order to obtain a more detailed characterization and understanding of the underlying electromechanical piezoelectric phenomena. Based on the above data, it was felt that a detailed analysis of the performance limitations of the amplifiers should be done describing limitations of the system such as:

1. Overall noise of the system as used for testing,
2. Resistance to external noise sources,
3. Sensitivity to static electricity,
4. Sensitivity to external vibration, and
5. Constraints on physical layout.

If more accurate measurements of the voltage and current signals can be made, then these results can be compared to

actual displacements of the bone or to actual measured forces applied to the bone, to get a better understanding of the electrical response of bone. In order to meet, at least in part, the above goals, the presently described work was conducted.

CHAPTER III

EXPERIMENTAL DESIGN

This chapter describes the development of the hardware (amplifiers, testing jigs for holding bone samples, packaging of the system) used for measuring the electrical response of bone, the preparation of the bone samples for testing, and finally, the testing procedure.

I. INSTRUMENTATION

The voltage and current amplifier circuits that Simpson developed were used in the construction of the amplifiers for this project. A high quality, low noise electrometer operational amplifier (op amp), the AD 515, was selected for the amplifiers to measure both voltage and current.

Benefits from using this device are:

1. High input impedance (10^{13} ohms);
2. Low cost (under \$20 in 1985);
3. Low offset voltage and temperature drift;
4. External null offset to neutralize any electrode polarization.

The manufacturer's specifications for the AD 515 op amp are listed in Table 1. The schematic for both circuits is shown in Figure 2.

Because of the high sensitivity of the system, associated with the small signals measured, special care was

Table 1. Manufacturer's specifications for the AD 515.

MODEL	AD515J
OPEN LOOP GAIN¹	
$V_{OUT} = \pm 10V$, $R_L \geq 2k\Omega$	20,000V/V min
$R_L \geq 10k\Omega$	40,000V/V min
$T_A = \text{min to max}$, $R_L \geq 2k\Omega$	15,000V/V min
OUTPUT CHARACTERISTICS	
Voltage @ $R_L = 2k\Omega$, $T_A = \text{min to max}$	$\pm 10V$ min ($\pm 12V$ typ)
@ $R_L = 10k\Omega$, $T_A = \text{min to max}$	$\pm 12V$ min ($\pm 13V$ typ)
Load Capacitance ²	1000pF
Short Circuit Current	10mA min (25mA typ)
FREQUENCY RESPONSE	
Unity Gain, Small Signal	350kHz
Full Power Response	5kHz min (16kHz typ)
Slew Rate Inverting Unity Gain	0.3V/ μs min (1.0V/ μs typ)
Overload Recovery Inverting Unity Gain	100 μs max (16 μs typ)
INPUT OFFSET VOLTAGE³	
vs. Temperature, $T_A = \text{min to max}$	3.0mV max (0.4mV typ)
vs. Supply, $T_A = \text{min to max}$	50 $\mu V/^{\circ}C$ max
	400 $\mu V/V$ max (50 $\mu V/V$ typ)
INPUT BIAS CURRENT	
Either Input ⁴	300fA max
INPUT IMPEDANCE	
Differential	1.6pF $10^{13}\Omega$
Common Mode	0.8pF $10^{15}\Omega$
INPUT NOISE	
Voltage, 0.1Hz to 10Hz	4.0 μV (p-p)
$f = 10Hz$	75nV/ $\sqrt{Hz}$
$f = 100Hz$	55nV/ $\sqrt{Hz}$
$f = 1kHz$	50nV/ $\sqrt{Hz}$
Current, 0.1 to 10Hz	0.003pA (p-p)
10Hz to 10kHz	0.01pA rms
INPUT VOLTAGE RANGE	
Differential	$\pm 20V$ min
Common Mode, $T_A = \text{min to max}$	$\pm 10V$ min ($\pm 12V$ typ)
Common Mode Rejection, $V_{IN} = \pm 10V$	66dB min (94dB typ)
Maximum Safe Input Voltage ⁵	$\pm V_S$
POWER SUPPLY	
Rated Performance	$\pm 15V$ typ
Operating	$\pm 5V$ min ($\pm 18V$ max)
Quiescent Current	1.5mA max (0.8mA typ)
TEMPERATURE	
Operating, Rated Performance	0 to $+70^{\circ}C$
Storage	$-65^{\circ}C$ to $+150^{\circ}C$

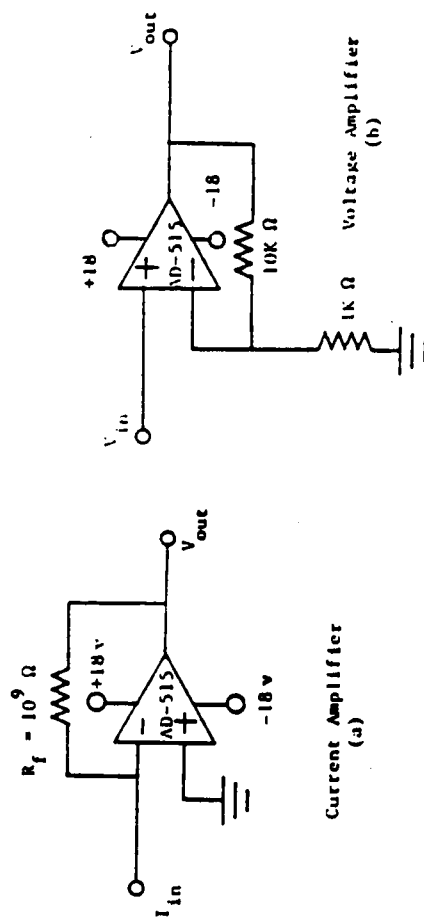


Figure 2. Current and Voltage Amplifiers.

given to the layout of the circuit board. First, the circuit board had to be grounded to prevent small changes in ground potential since these would show up as a baseline shift; second, the very small currents involved (less than 100 pA) had to be shielded from outside noise sources to prevent noise masking of the signal to be measured; third, because of the sensitivity of both the voltage and the current amplifiers, stray capacitance in the circuit had to be as low as possible to prevent changes in capacitance from showing up as changes in voltage or current.

To reduce noise problems, a circuit board was designed with a large ground plane to which all grounding points were attached; most of the circuit board consisted of copper clad ground, with a 3 cm border around the circuit board. The two amplifiers were mounted on the same board but were separated from each other by a portion of the ground plane to reduce circuit crosstalk. Interference between output and input was reduced by keeping the input and output sides of the circuits separated. To reduce external power supply noise sources, the circuit was powered by four 9-volt batteries mounted on the circuit board. To eliminate any noise that might be introduced between the batteries and the AD 515, power supply lines were also filtered before going to the AD 515. The final design of the circuit board can be seen in the photograph presented in Figure 3.

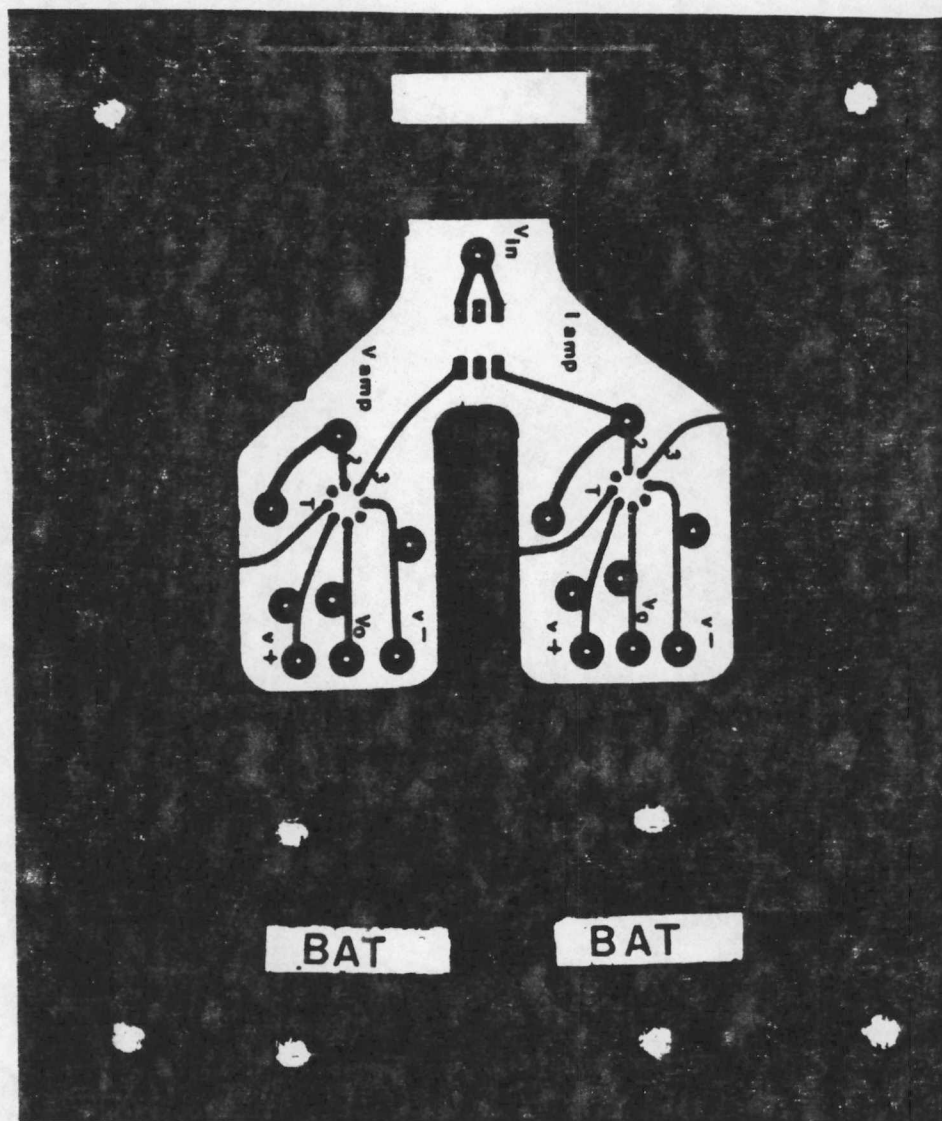


Figure 3. Printed Circuit Board Layout.

To reduce interference even further, the leads from the bone to the amplifiers had to be made as short as possible and shielded from noise. Reducing lead length is particularly critical with the high impedance voltage amplifier due to possible noise pickup of the leads. Therefore, the testing jig and the amplifier circuit board were placed together in a grounded and shielded steel enclosure. This arrangement kept the length of the leads from the bone to the amplifiers to a minimum and shielded both the amplifiers and the bone samples from external noise.

Even with this arrangement, considerable care had to be taken in designing the connecting lines from the bone to the amplifiers to prevent noise. Even though shielded coaxial cable (coax) was used to reduce noise, potential problems associated with coax such as cable leakage, noise, and capacitance had to be considered. Capacitance in particular had to be considered because it has the ability to store or filter the signal or both and thus to distort it. Movement of the cable could cause several problems, any of which could significantly alter the signal:

1. Frictional movement of the cable insulation and the shield could cause electrostatic charge which could be measured as signal.

2. Cable movement could cause changes in capacitance. If there is any charge in the cable, this change of capacitance would show up as a signal.

3. Cable movement could cause the middle conductor to move in relation to the shield. If the shield produces a magnetic field, then motion of the conductor within the magnetic field could produce a voltage.

The configuration used kept the cable length to about 15 cm. A steel enclosure measuring 15 x 18 x 30 cm was used. The final configuration of the testing apparatus is shown in Figure 4, and will be described below. The holding jig for the bone samples was constructed to fit within the available space. To prohibit any significant electrical leakage of the signal, a plexiglas ($\rho \approx 1 \times 10^{15}$ ohm-cm) jig (1) was made. It was designed with a 10 mm slot to hold the bone samples (2) uniformly for testing. The jig also was made adjustable to allow easy changes in testing conditions such as the drop height and weight of the plunger (3). The jig was then designed to be easily inserted and removed (4) from the enclosure for the changing of samples and the adjustment of test conditions. Finally, the jig was designed to position an optical probe (5) used to measure the actual deflection of the bone.

II. PREPARATION OF THE BONE SAMPLES

Typical testing of bone in cantilever bending is done using bone which is 30 mm long, 2 to 3 mm wide, and 1 to 2 mm thick. Sample specimens from the same stock used by Simpson (1980) were milled from human femur obtained from

Figure 4. Final Configuration of Testing Apparatus.

- (1) Plexiglas Holding Jig.
- (2) Bone Sample.
- (3) Plunger.
- (4) Wing Nuts.
- (5) Optical Probe.
- (6) Spring Loaded Release Catch.
- (7) Plunger Release Cable.
- (8) Plunger Reset Cable.
- (9) Dip Switch.
- (10) BNC Connectors.
- (11) Alligator Clips.
- (12) Connecting Cable.
- (13) On-off Switch.
- (14) Batteries.
- (15) Printed Circuit Board.
- (16) Ground Wires.
- (17) Optical Probe Cable.
- (18) Optical Probe Holding Jig.

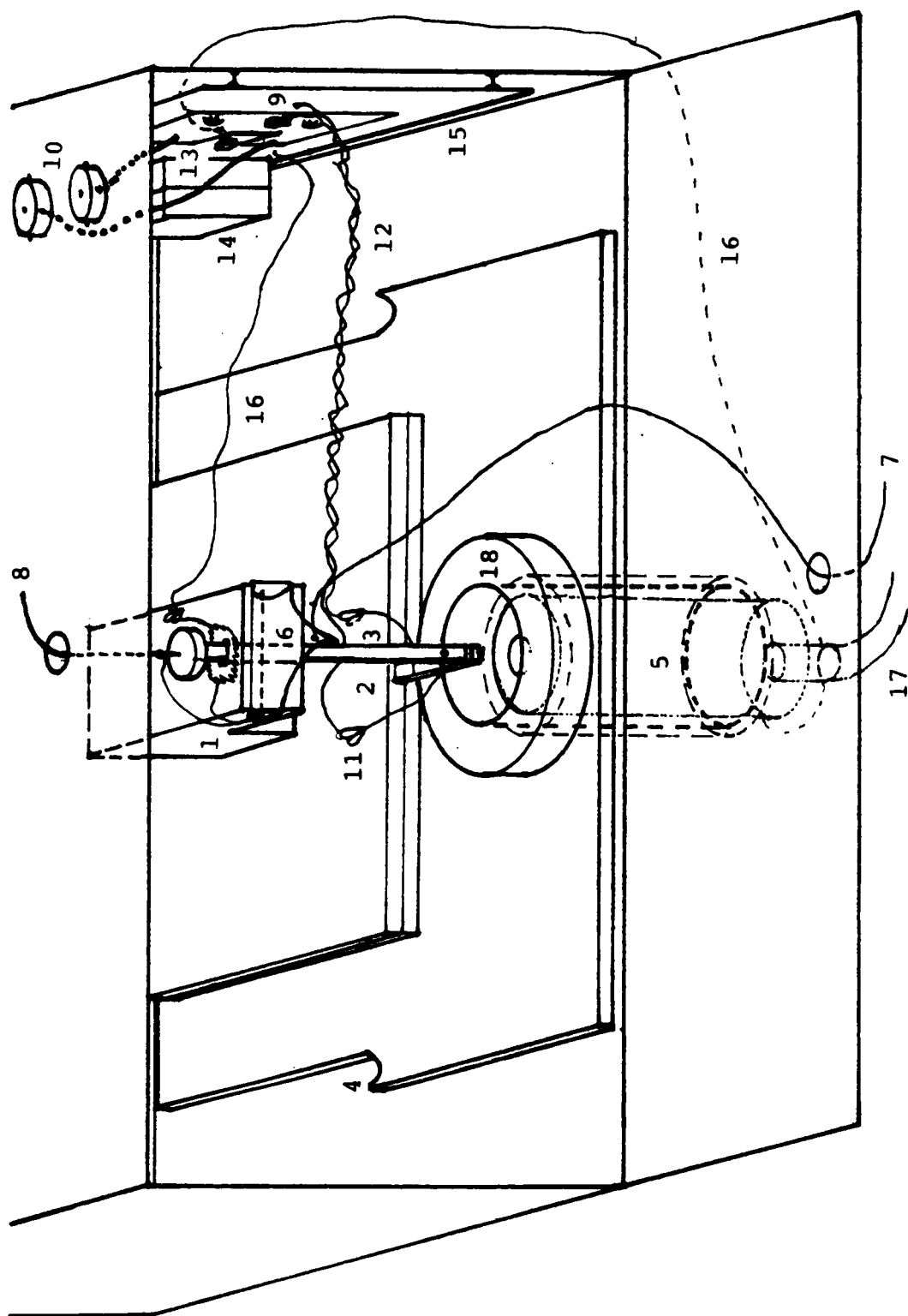


Figure 4.

The University of Tennessee School of Dentistry, Department of Dental Anatomy, Memphis, Tennessee. Because information about the donors was unavailable, and to insure consistency, Simpson had performed some compression tests on the bone. The mechanical properties he obtained were consistent with results which had been previously reported in the literature.

The milling sequence used for obtaining bone samples is shown in Figure 5 and is described in detail by Simpson (1980). The samples were then dried.

Much attention has been given in the literature to the drying procedure and hydration state of bone samples used for testing. The three methods normally used are 1) oven drying, 2) desiccation, and 3) air drying. Since Simpson observed larger currents from air dried samples than for oven dried samples, all of the samples used for testing were air dried.

Electrodes were then attached to the prepared bone samples. Silver, gold and platinum electrode leads have all been used with silver paint or silver epoxy electrodes for measuring bone piezoelectricity. But since silver epoxy cures at 150°C , and since the bone ultrastructure starts to break down at this temperature, silver paint which dries in about 30 minutes was used for the electrodes.

Cochran et al., (1968) found that the position of the electrodes on the bone significantly altered the magnitude

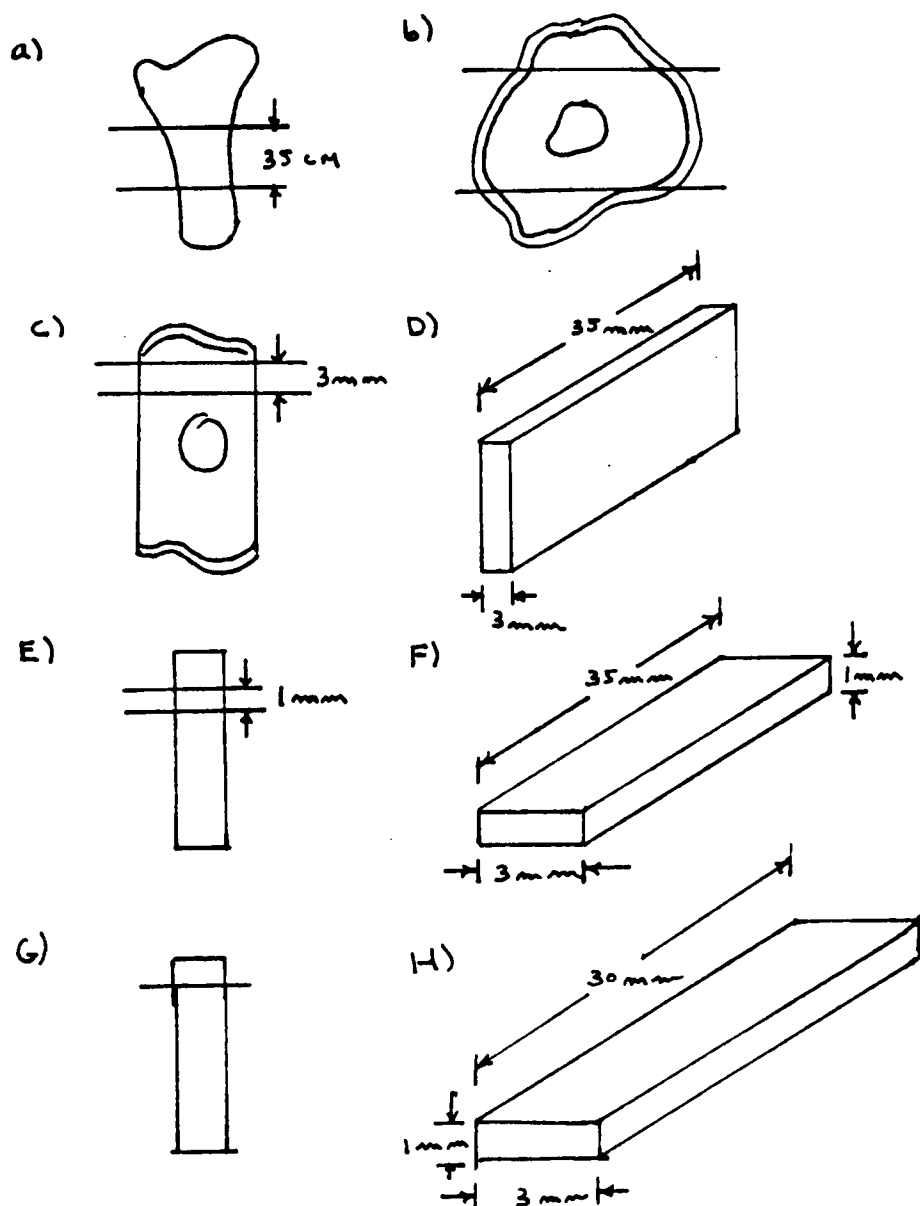


Figure 5. Milling Procedure for Sample Preparation.

Source: M. L. Simpson, "Evaluation and Development of the Instrumentation Used in the Measurement of Bone Piezoelectricity," M.S. Thesis, The University of Tennessee, Knoxville, 1980, p. 47.

of the signal from the bone. He also found that the largest signal could be obtained by attaching the electrodes 2.5 mm from the load on the convex or free end and 2.5 mm from the clamp on the concave or clamped end. Following this convention, silver paint electrodes, 1 to 2 mm wide, were painted across the entire width of the bone. (Ten mm of the bone was then placed in the holding clamp with 20 mm of exposed bone subject to cantilever bending.) Platinum electrode leads were used, because of platinum's excellent electrical properties, and attached to the silver paint electrodes. Since little stress is placed on the electrode leads, they were attached only with the silver paint.

III. SYSTEM OPERATIONAL CONSIDERATIONS

An enclosure that would allow access to samples and changes in test conditions had to be selected. The most readily available solution to this problem was found to be an army surplus ammunition box. (These boxes are readily available, made of steel, and easy to open and close.)

To make testing easier, a release mechanism consisting of a spring loaded release catch (6) and matching hole in the plunger was designed into the holding jig which would allow the plunger to be consistently set and released (Figure 4). Since testing was to be done inside the enclosure, the release mechanism was operated by separate strings attached to the release catch and the plunger.

Holes were drilled in the enclosure and the strings were run through, allowing for easy loading (7) and unloading (8) of the plunger for testing without having to open the enclosure.

A dip switch (9) was installed on the circuit board to switch the input from the bone to either of the two amplifiers as needed. (Voltage and current measurements cannot be made simultaneously because the current amplifier, by design, shorts the voltage signal to ground, making voltage measurements impossible.) The output of each amplifier was attached, using coax cable, to BNC connectors (10) mounted on the enclosure.

Alligator clips (11) were attached to the holding jig, adjacent to the sight where the bone samples were held, and to the amplifier with 15 centimeters of coax (12). The electrode leads from the bone could then be easily connected to and disconnected from the alligator clips allowing ease in sample changes.

An on-off switch (13) was mounted on the circuit board near the batteries (14) with a hole cut in the enclosure to allow access to the switch. This switch was placed on the circuit board (15) to reduce lead length for noise reduction purposes and to reduce the possibility of external noise pickup due to an externally mounted switch. The switch case and the enclosure were grounded to the circuit board ground plane.

Finally, access had to be provided for the optical probe which was used to measure deflection of the bone samples. A hole was cut into the bottom of the enclosure and a sleeve was made to hold the probe head in position under the free end of the bone samples.

The electrical response of bone was measured and recorded on a Hewlett-Packard Model 5420A Digital Signal Analyzer (with a model 9872A Digital plotter used to obtain hard-copy plots). In the plots obtained, the scale of the vertical axis is in millivolts and the scale of the horizontal axis is in milliseconds (or seconds), except where noted. A negative time delay was used for recording data, so that baseline values of current and voltage from bone before loading could be established. The actual magnitude of the electrical response of bone was obtained by dividing the peak magnitude of the response by the D.C. gain of the appropriate amplifier used for making the measurement. The D.C. gain of the voltage amplifier was 11 volts/volt, and the D.C. gain of the current amplifier was 1 volt/nanoamp. The deflection of the bone was measured using a Mechanical Technology Incorporated KD-100 Optical Probe with a KD-1103X module and a KD-LS-I optical extender. The magnitude of the displacement of the bone samples was not calculated due to the difficulty in calibrating the probe to each individual sample. The waveshapes from the optical probe showed displacement characteristics of the

bone samples. The expected range of the response (based on Simpson's work) for voltage was about 2 to 40 mV and for current was about 2 to 100 pA.

CHAPTER IV

TESTING PROCEDURES FOR EVALUATING NOISE RESPONSE

I. PRELIMINARY TESTING

Before any testing of bone samples was begun, the case was grounded to the circuit board ground plane to reduce noise. The case and circuit board were also grounded to the Fourier analyzer through the BNC cables. During the initial set up, there was considerable 60 cycle noise being produced by the insertion of the optical probe. Even though the probe was an optic fiber light source, the end of the probe is metal encased and about 15 centimeters long, with only about five centimeters actually inside the enclosure. The electrical interference was caused by the probe head acting as a noise antenna when it was inserted into the enclosure. This was easily solved by grounding the probe head to the ground plane.

Further preliminary testing showed that the metal scope cart on which the equipment was located was also a source of noise, even though the testing box was positioned on a wooden shelf about one foot above the cart. This noise was eliminated by grounding the cart to the metal box.

Early testing showed that movement of the plunger within the plunger guide was producing static buildup which could cause some measurement problems. In particular, this static charge caused a baseline voltage drift when the

plunger was moved. After about one minute, the drift would return to a stable voltage. To reduce the buildup of static charge, the plexiglas plunger guide, which positions the plunger over the bone for a drop test, was grounded to the circuit board by a wire attached to the Plexiglas guide with silver paint. The metal release catch that held the plunger up before each test was also grounded to insure that the metal plunger was discharged and at ground potential except during a test. Also a ground wire was attached to the plunger so that the plunger could be grounded even during a test. To determine what effect grounding and static charge would have on testing of bone, all of these grounds were made removable.

Two plungers were made for testing. The first plunger was made of metal with an insulated tip 3mm long (and made of epoxy) to prevent bleed off of the signal to the plunger. To determine what effects, if any, charge storage of the metal conductor would have on the testing of bone, a second plunger was made of micarta (a phenol formaldehyde resin), a good insulator ($\rho \approx 3 \times 10^9$ to 2×10^{16} ohm-cm, depending on humidity and surface contamination).

II. NOISE TESTING

The first tests were made to determine if the system was sensitive to external vibration. Walking around the room or shutting the door would show up as a measurable

signal during preliminary setup. In an attempt to determine which portion of the system was sensitive to vibration, a series of uncalibrated tests was performed. The shelf which supported the testing box was uniformly tapped by hand, to test for external vibration sensitivity. Three conditions were used in this test to isolate the source of noise sensitivity. First, the coax was disconnected from the circuit board to determine sensitivity of the circuit board to mechanical vibration. Next the coax was reattached to the circuit board with no electrode leads to test for sensitivity to vibration of the coax. And finally, a bone sample was installed and the whole system was tested for sensitivity of the system to vibration of the electrode leads. This testing was done using both the voltage amplifier and the current amplifier.

III. ARTEFACT TESTING

One of the primary goals of this research was to test for the possibility of artefact signals. Unlike the work of Simpson (1980), who dropped the weight from a string attached to the bone, the weights for this experiment were dropped onto the end on the cantilever beam from a plunger that was directed to the end of the cantilever beam through a plexiglas guide. This arrangement allowed for uniform impact testing and made it easier to use the optical probe.

A release catch held the plunger above the bone and was released by a string fed through a hole in the testing box. After the drop, the plunger was unloaded from the bone and returned to the ready position by pulling the plunger up with another string fed through the top of the box. Tests similar to those done for vibration sensitivity were performed on a plexiglas "bone" chip (1.6 x 3 x 30 mm) to test for the possibility of artefact signals unrelated to bone response except that weights were dropped onto the end of the plexiglas chip. As with the vibration sensitivity testing, tests were also performed with the coax disconnected from the circuit board to determine if there was any sensitivity of the circuit board to vibration due to performing a normal test. The procedure was then performed again with the coax connected to determine if there was any sensitivity of the coax to vibration during a normal test. Finally, a normal test was performed with the plexiglas "bone" attached to the coax to test for sensitivity of the system to vibration of the electrode leads during normal testing.

Because of the potential problems associated with coax which have been discussed previously, four types of cables were tested to determine if the cabling method would have any significant effect on the results. The four types of cable used were:

1. Large coax (0.15 inch),
2. Small coax (0.085 inch),
3. Twisted pair made from 22-gauge stranded wire, and
4. Twisted pair made from 22-gauge solid wire.

Several long-term tests (up to five minutes in duration) were done to determine voltage drift, if any, of the voltage amplifier. After several minutes, voltage measurements were made to determine the steady state voltage. A test was then made with the coax attached to determine leakage of the coax. Finally, a drift test was done by observing the signal for up to 2 minutes after the amplifier was turned on.

Comparison testing of bone and plexiglas was begun once the "artefact" response from plexiglas "bones" had been determined. For normal testing, a 35 gram plunger was dropped from a height of one centimeter onto the end of the specimen. Bone and plexiglas were tested the same way and measurements were made of both current and voltage. To verify the results, additional tests were performed, using a plunger weighing 17 grams, and initial plunger heights of 1 and 1/2 centimeter for both plungers. To reduce individual sample variation, averaging was performed over 10 tests for each test condition.

Deflections were measured with the optical probe, so that a standard reference was available to compare signals from the voltage and current amplifiers. Next, a series of

tests was done on the bone to determine the frequency spectrum for the probe, voltage amplifier, and current amplifier.

To compare results of this experiment with the results of the work of Simpson, a 35 gram weight, which was connected to the bone by a string, was dropped from a height of about one centimeter. This test was done for both the bone samples and for the plexiglas "bone."

Since bone exhibits some photoelectric response (Becker and Brown, 1965), and since an optical probe was being used, tests to evaluate if photoelectric effects would influence the results of normal testing were performed. Tests were done comparing signals with and without the optical probe and with several intensities of light from the probe (which uses a high intensity halogen light source).

Also, to verify the results of others, simple tests were done to determine if there was a direct photoelectric effect by directly shining a halogen light source on the bone.

The size of the electrodes and electrode placement were chosen to correspond to that of Simpson (1980), and Cochran et al. (1968) so that results from the present study could be compared to earlier work. Simpson's method was duplicated by using weights to load the bone specimen. A 70 gram weight was tied to the end of a string which was attached to the end of the bone specimen. The weight was

then dropped from a height of one centimeter to simulate an instantaneously applied load. To verify that an artefact signal was not being produced, this test was repeated using a plexiglas "bone."

Some mechanically induced noise in the system (such as noise due to vibration of the cables or the circuit board) is a function of the applied load. To isolate this noise source from the response of the bone, some signal processing was done on the collected data. An average of 10 response signals were collected and stored. The electrode leads from the bone samples were then reversed and an average of 10 response signals were again collected and stored. This signal would contain both the noise signal from the mechanically produced noise as well as a signal from the bone of a reversed polarity. By subtracting one of these two signals from the other, the mechanically produced noise signals would cancel and the signal from the bone would be isolated. This was done for both bone and plexiglas to determine if an artefact signal would be produced.

Finally, a step function input was simulated by using a one kilogram weight dropped from a height of two millimeters and suddenly released. This would produce a steady state deflection of about two millimeters without damaging the bone sample. This procedure was done for both bone and plexiglas.

CHAPTER V

EXPERIMENTAL RESULTS

The experimental protocol is described in Chapter IV. The size of the electrodes and electrode placement corresponded to that used by Simpson (1980) and Cochran et al. (1968). Thus, their previously reported results were used to demonstrate the validity of the data obtained from this research.

I. GROUNDING

It was recognized that grounding problems would exist. Attempts were made to identify as many of these problems as possible before testing was started. As additional noise sources were discovered, they were usually able to be eliminated with little difficulty. The most significant noise problem was static electricity due to movement of the plunger within the plexiglas plunger guide which did at times produce voltage signals from the amplifiers that approached the ± 10 volt limits of the Signal Analyzer. Slow movement of the plunger produced D.C. drift, which would return to a stable voltage after several seconds. More rapid movement (which was needed for testing) produced voltage signals that appeared very much like the measured signal waveforms obtained from bone but were of significantly greater magnitude. These signals were reduced

by grounding of the plunger guide and the plunger, and finally by using a plunger made of micarta instead of metal. The grounding of the plunger (Figure 6A) reduced sustained D.C. offsets which were observed in the voltage signal of the ungrounded plunger (Figure 6B), but did not change the magnitude of the response. Figure 7 shows that grounding the plunger after an impact significantly altered the waveshape at the time at which the plunger was grounded; furthermore, grounding the plunger smoothed out some of the spikes that occurred in the waveform. This figure demonstrates that an ungrounded plunger could produce effects due to surface charge. Grounding did not significantly alter the magnitude of the response, which was 5.5 to 9.1 mV in this case. While grounding provided some benefit, these attempts were only partially successful. Quantitative evaluation of these benefits was difficult because results were not consistent, as shown in Figure 8. These figures show that a grounded plunger showed no significant change in comparison to an ungrounded plunger and plunger guide, these signals were all about 0.15 volts in magnitude. (Changes in humidity are most likely responsible for this inconsistency, since inconsistent results occurred on different days while results during any given day were fairly consistent.) In spite of the inconsistent results, grounding generally seemed to produce better results by reducing the sustained D.C. offsets which

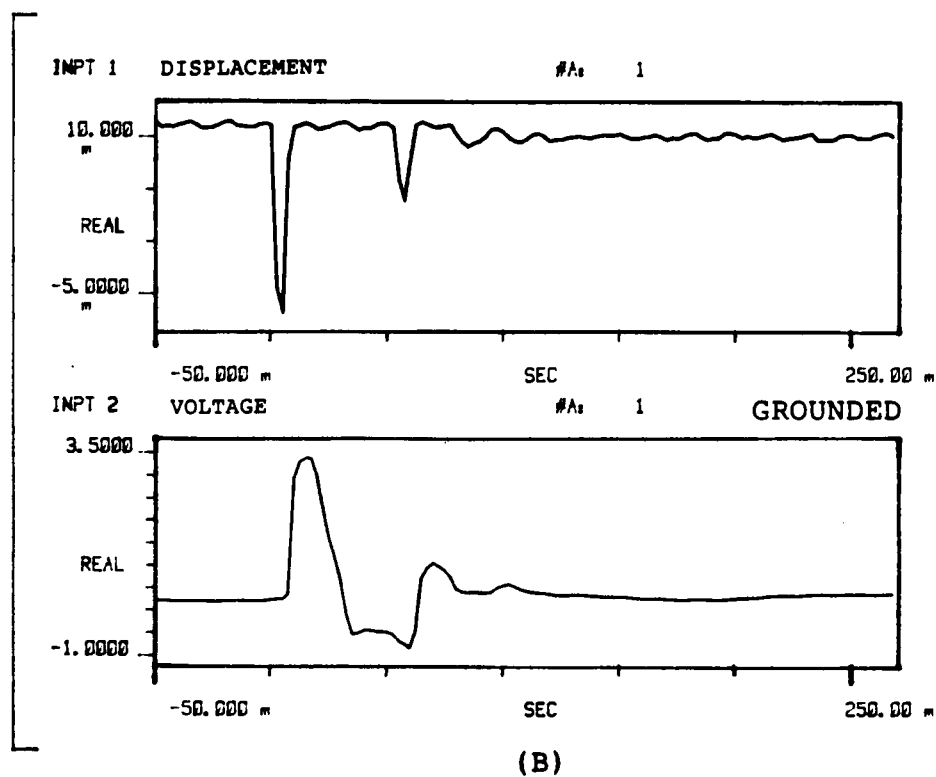
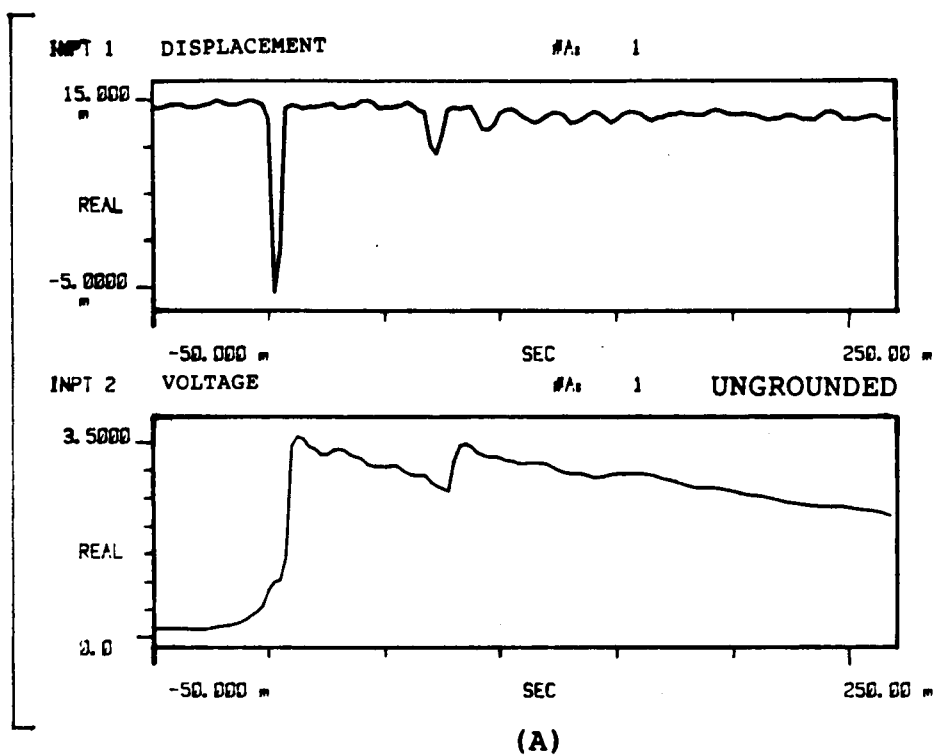


Figure 6. Comparison of Ungrounded and Grounded Plungers.

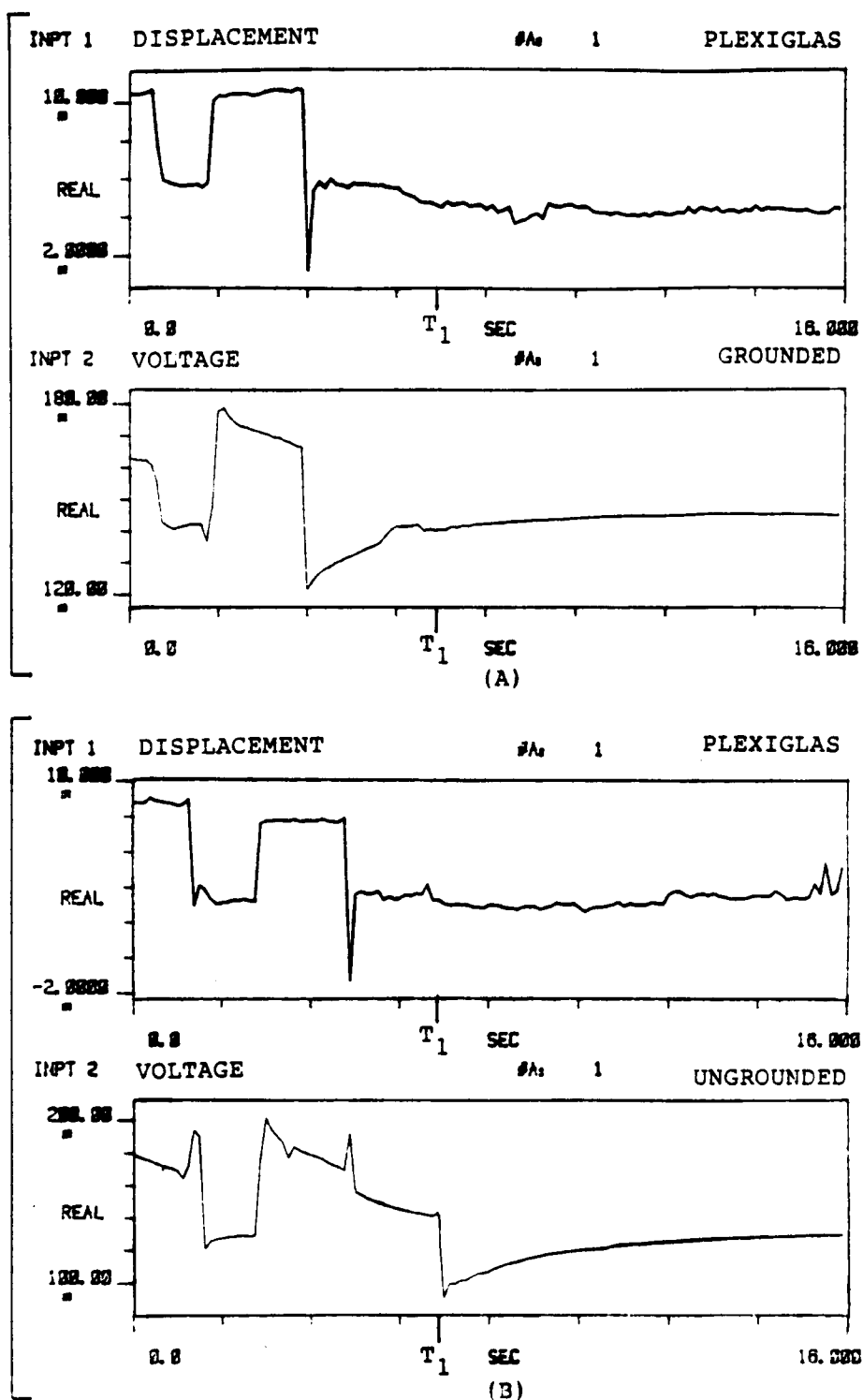
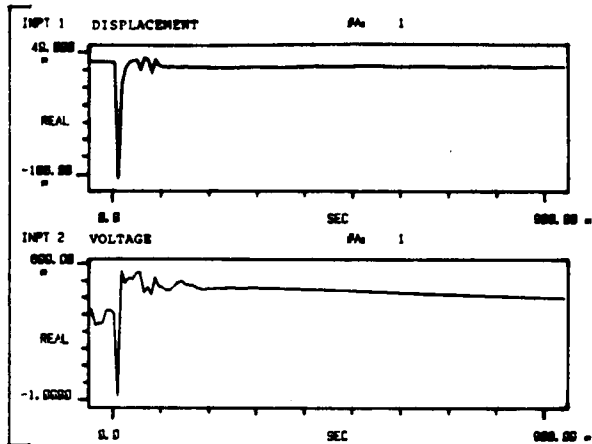


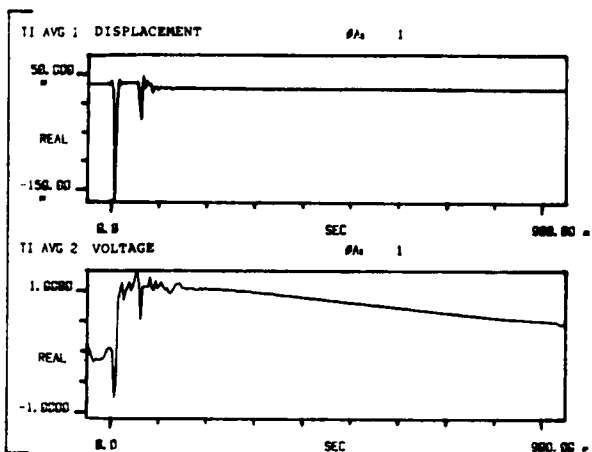
Figure 7. Effect of Grounding Plungers with Grounded Release Catch at Time (T_1) for Grounded and Ungrounded Plungers.



(A)

Ungrounded Plunger

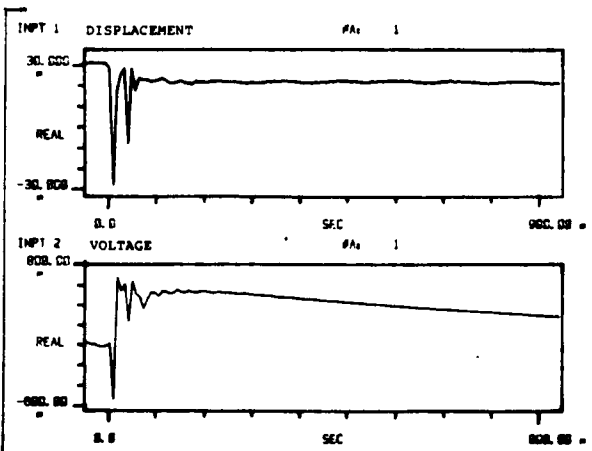
Ungrounded Plunger Guide



(B)

Ungrounded Plunger

Grounded Plunger Guide



(C)

Grounded Plunger

Grounded Plunger Guide

Figure 8. Comparison of Ungrounded Plunger, Grounded Plunger Guide, and Grounded Plunger and Plunger Guide.

were observed when the ungrounded plunger was used.

To avoid the conduction and charge storage effects of the metal conductor, the second micarta plunger was used, but this also failed to produce consistent results. Since micarta is not conductive, no tests were done with the micarta plunger grounded. The micarta plunger produced two unexpected changes in the nature of the results as shown in Figure 9. First, the micarta plunger bounced much more than did the metal plunger, and second, the average peak magnitude of the response tended to be substantially higher (0.82 volts for the micarta plunger as compared to 0.46 volts for the metal plunger). In general, the micarta plunger seemed to produce much more static charge (and grounding of the plunger guide did not seem to have any effect); therefore the micarta plunger was not used extensively for testing.

Current waveform shapes were not changed by grounding. The magnitude of the current output was increased by about a factor of two when the micarta plunger was used instead of the metal plunger. These signal magnitudes, (both current and voltage) were much higher than magnitudes previously reported (Simpson, 1980), indicating that the signals were not due to the piezoelectric response of bone but were due instead to static charge. This observation was not consistent however, for at times, the static charge was

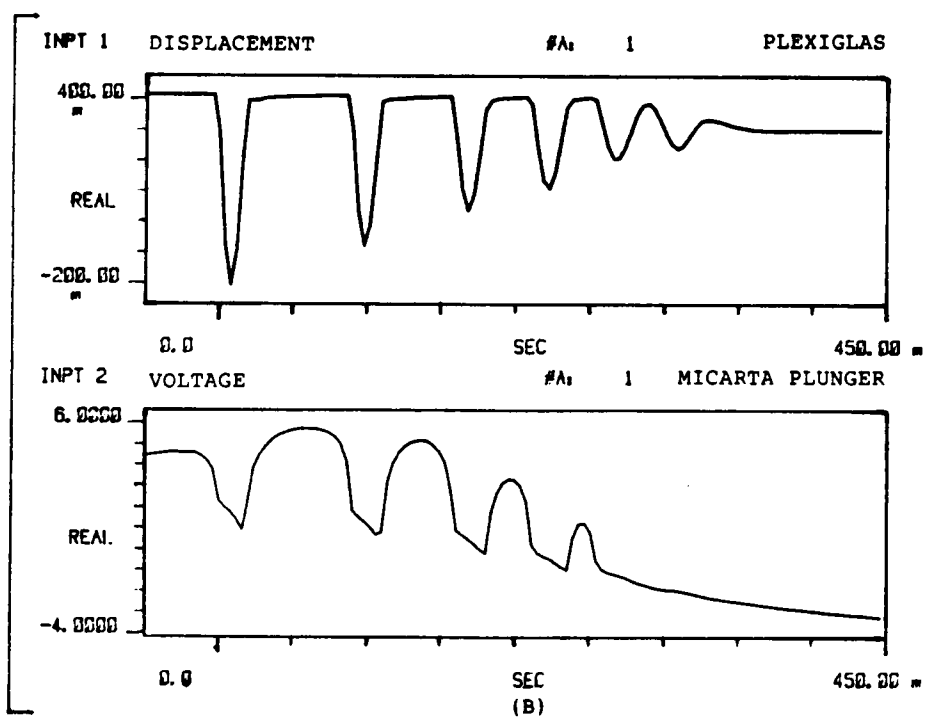
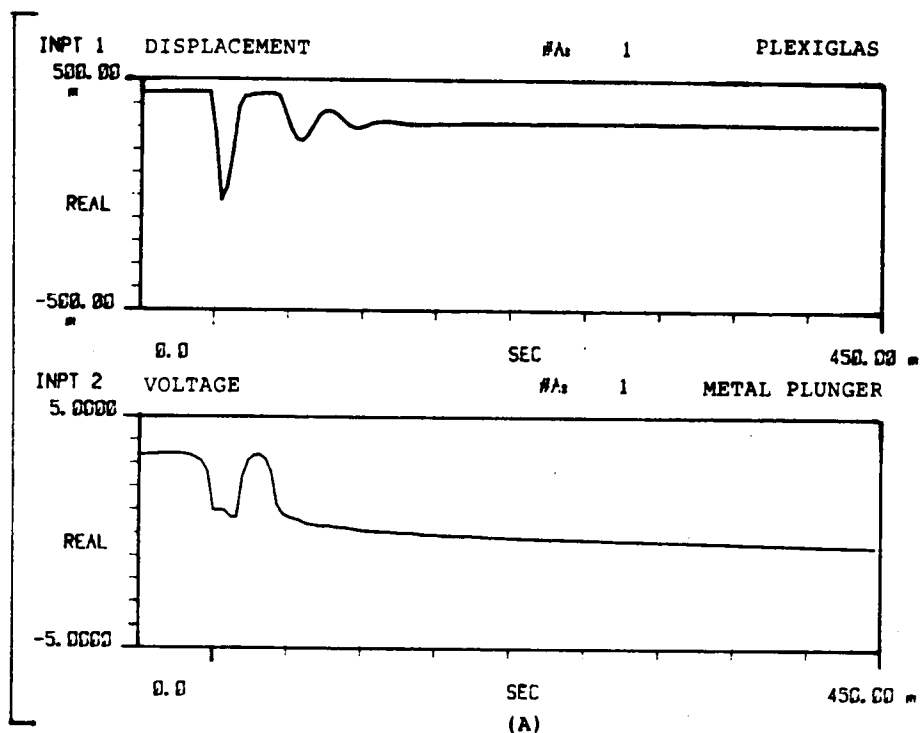


Figure 9. Comparison of Metal Plunger and Micarta Plunger.

reduced significantly and the signals were of the expected magnitude.

II. NOISE INDUCED BY VIBRATION

Early study showed that signals were generated when people walked across the laboratory floor during testing, therefore tests were performed to determine the measurement system sensitivity to external vibration. The data collected for this investigation were deflection signals from the optical probe, voltage signals from plexiglas, and current signals from plexiglas. The signals from the optical probe ranged from 0 to about 5 mV. Negligible voltage was produced due to vibration of the circuit board with no coax attached. With the coax attached, however, there were significant signals of an average magnitude of 9.1 mV from the output of the voltage amplifier. When the electrode leads from the plexiglas "bone" were attached to the coax, approximately the same signal was observed. Similar tests for current using the circuit board only, circuit board with coax, and circuit board with coax and electrodes attached to a plexiglas "bone" chip, produced signals of 60 to 100 mV from the output of the current amplifier for all three tests, showing that the circuit board and probably the coax as well were sensitive to vibration. The vibration sensitivity testing demonstrated the need for care but showed that only significant

vibration, such as heavy walking or slamming of doors, would affect testing.

III. ELECTRODE LEAD ARTEFACT TESTING

Motion of the electrode leads appeared to be the most likely source of significant artefact signals. Tests were performed similar to those performed for sensitivity to external noise (since normal testing could also produce vibration induced response). The results of this testing are shown in Figure 10 (for voltage) and Figure 11 (for current). These tests, using a plexiglas "bone" with no grounding of the plunger guide or the plunger, consisted of three parts: 1) no connections to the circuit board, 2) coax only, and 3) coax and electrode leads attached to the "bone." There was no appreciable response due to the circuit board only. However, significant response was obtained, both for current and for voltage, when the coax was connected to the circuit board, clearly demonstrating the need to

- 1) reduce motion of the coax,
- 2) find another cable connection less sensitive to vibration noise, or
- 3) move the amplifiers closer to the bone samples.

When the electrodes were connected, there was no appreciable difference between the response of the coax alone and the coax with the electrode leads attached. If artefact was

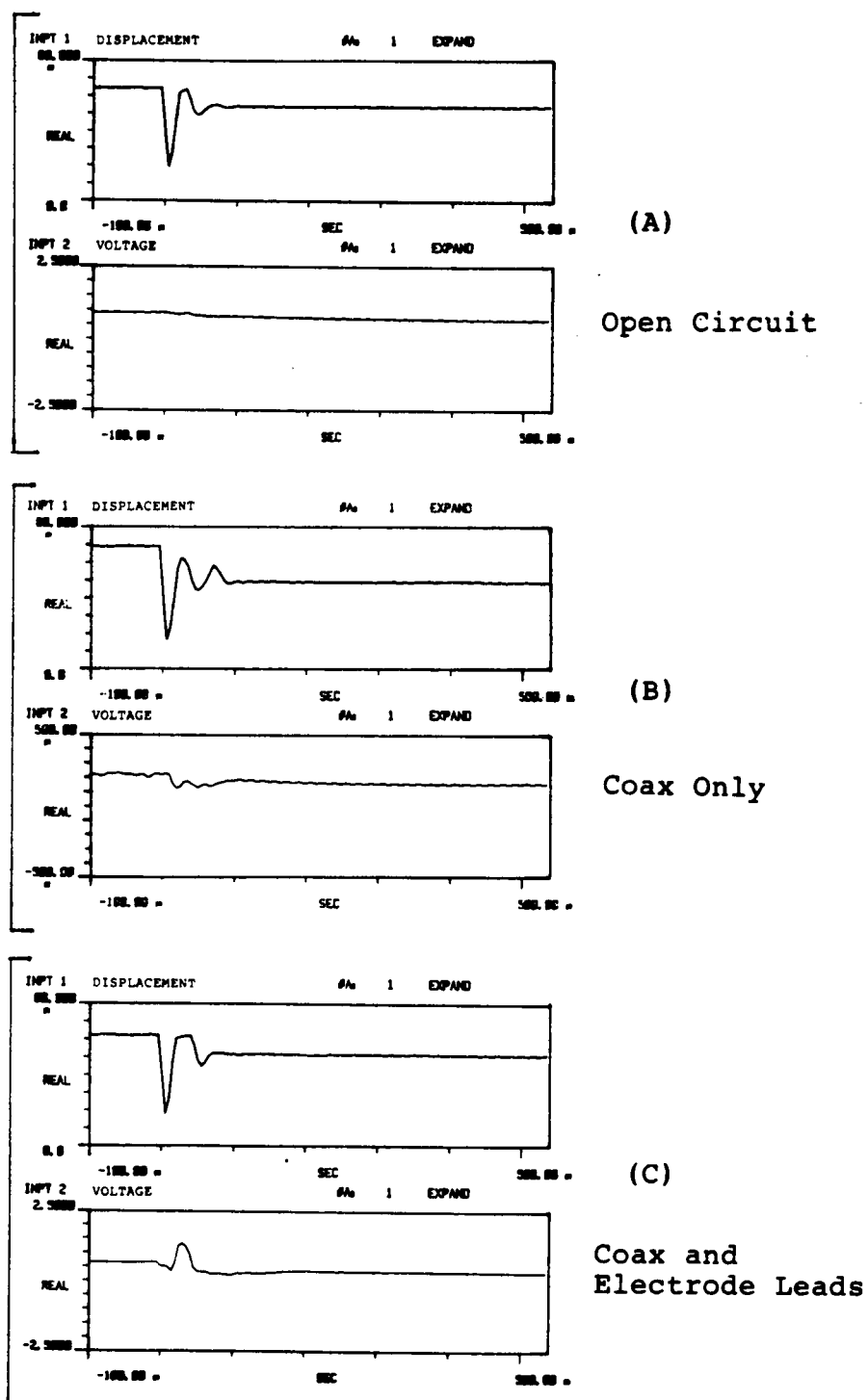


Figure 10. Voltage Response of Artefact Testing of Plexiglas.

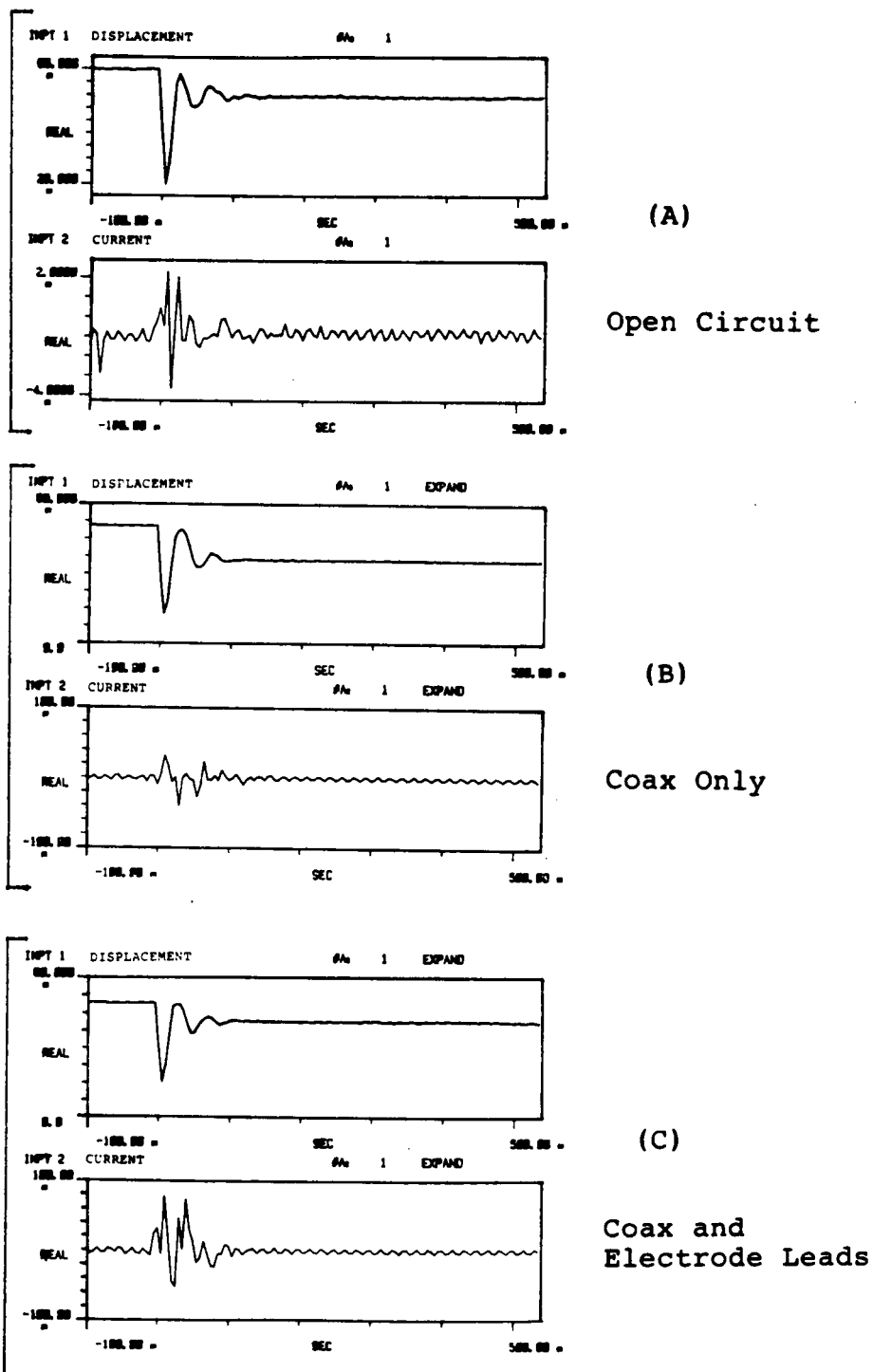


Figure 11. Current Response of Artefact Testing of Plexiglas.

present due to the electrodes, as was suspected by Bassett (1968), then the response was probably much smaller than the normal piezoelectric response from bone.

Because of the large signals obtained from the test using coax, additional tests were performed to determine if changes in the cabling used would reduce the noise problems. Four types of cables were used for comparison (see below).

CABLE COMPARISONS

Wire	Voltage	Current
stranded twisted pair	24.6 mV	120 pA
solid twisted pair	30.9 mV	100 pA
large coax (0.15 inch)	22.7 mV	80 pA
small coax (.085 inch)	15.5 mV	60 pA

Results indicated a small benefit from using twisted pair instead of coax, but the benefits were negligible. While not conclusive, the results do tend to support the conclusion that the small coax is the cable which yields the lowest artefact signal. However, the smaller signals could also have been due to cable leakage or capacitance, rather than a better cable. In the absence of complete data, additional tests are needed to determine the electrical properties of these cables, or low leakage, low capacitance coax that is resistant to vibration should be used.

Foam rubber was added around the small coax cable in an attempt to dampen coax cable motion, however the damping did not provide a consistent benefit and very often deteriorated performance. Figure 12 shows that the foam rubber damping

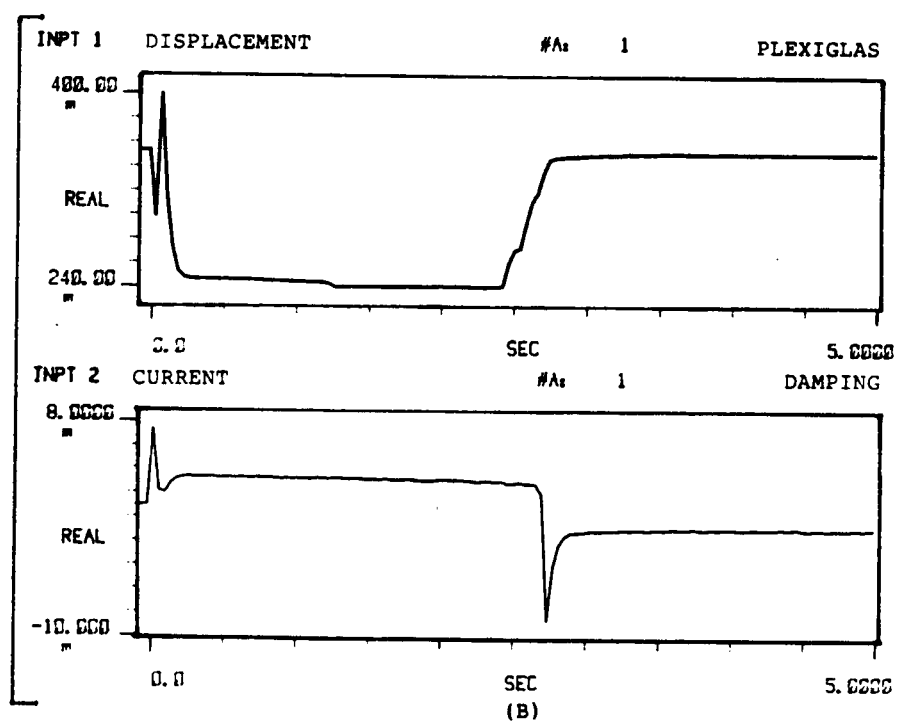
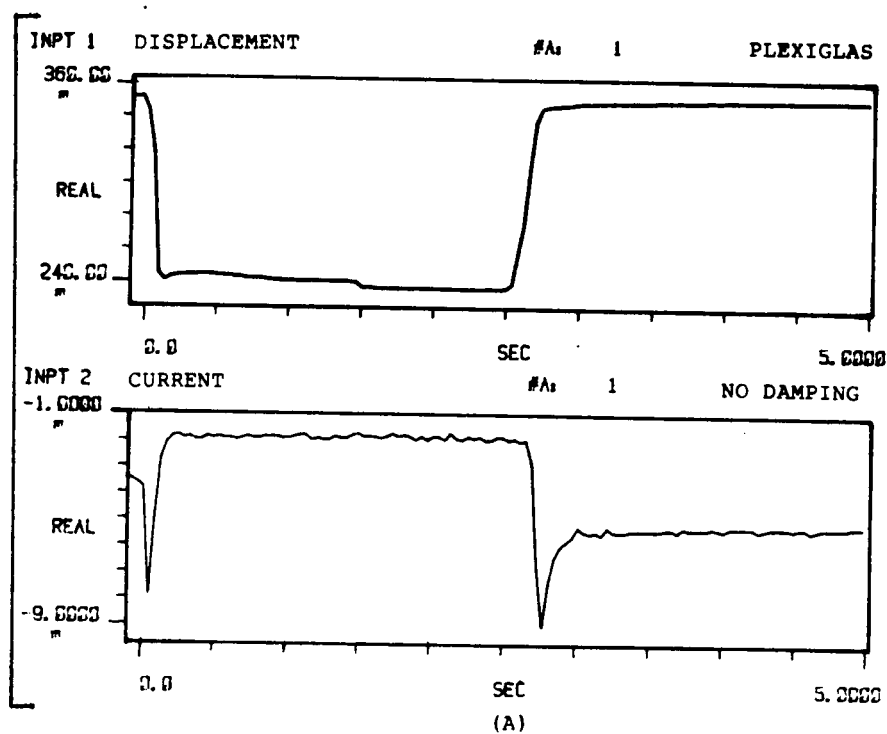


Figure 12. Effect of Foam Rubber Damping of Coax.

actually increased the magnitude of the current response from 8 pA to 18 pA with the electrode leads disconnected. Some configurations of damping material could also produce mechanically induced ringing in the cable. Based on these results, no further attempts were made at damping coax cable motion.

IV. SIGNAL DRIFT CALIBRATION

Voltage drift was tested to determine a normal baseline voltage so that results could be compared more reliably. It can be seen from the data obtained that the open circuit voltage, at the amplifier output, became stable at about 0.6 volts, closed circuit voltage using plexiglas became stable at 0.45 volts, and closed circuit voltage using bone became stable at 0.15 volts. Since the time required to return to the baseline voltage was 20 to 40 seconds, this was the time allowed between tests. (With bone attached, there was still some drift occurring even after 5 minutes, but this was a voltage that floated around 0.15 volts.)

Residual noise levels were also measured for both amplifiers after they had reached a stable output voltage. The output noise level was 150 microvolts for the current amplifier and 2 millivolts for the voltage amplifier in a normal testing configuration using a bone sample and with all leads attached.

V. SIGNIFICANCE OF SYSTEM ARTEFACT

The response of plexiglas was compared to bone to determine if the response from bone may contain significant "artefact." Figures 13 and 14 were typical and showed comparisons of the responses of bone with plexiglas. The results indicated that about 4% of the current response and 6% of the voltage response was due to "artefact" signal. (Since most of the "artefact" signal was due to motion of the cable connection between the bone and the amplifiers, moving the amplifiers closer to the bone and reducing the length of the cable connection between the bone and the amplifiers would reduce this even further. Static interference was negligible during these tests.)

VI. TESTING OF BONE

Testing of the bone samples was done by dropping weights onto the ends of the bone chips as described in Chapter IV. Three different bone samples were used along with the plexiglas, and the dimensions are given below.

BONE SAMPLES

Bone	Dimension
Bone 1	2.0 x 3.0 x 30 mm
Bone 2	2.1 x 2.7 x 30 mm
Bone 3	2.0 x 3.0 x 30 mm
Plexiglas	1.6 x 3.0 x 30 mm

Comparisons of the three bone samples showed that there was no significant difference between bone 1 and bone 3 but that

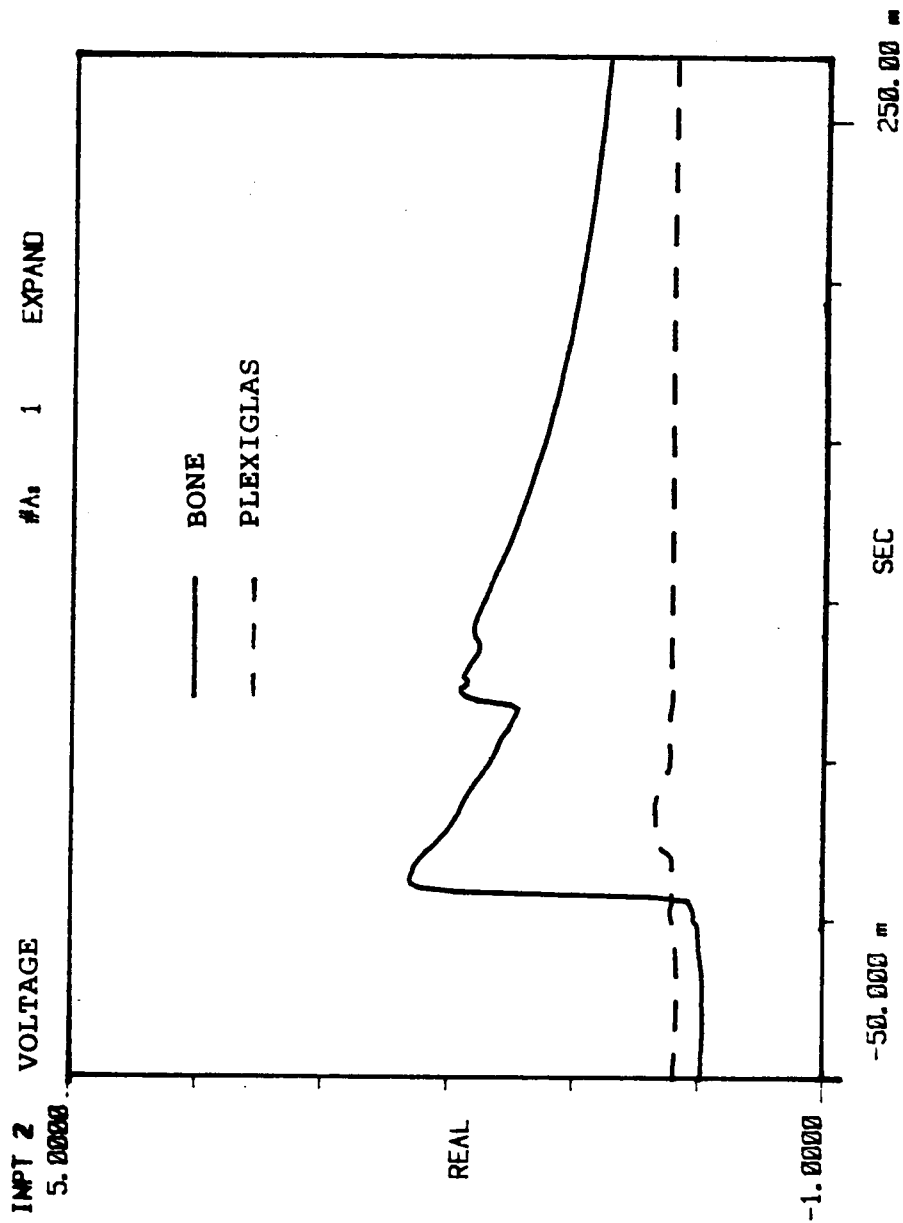


Figure 13. Comparison of Voltage Responses of Bone and Plexiglas.

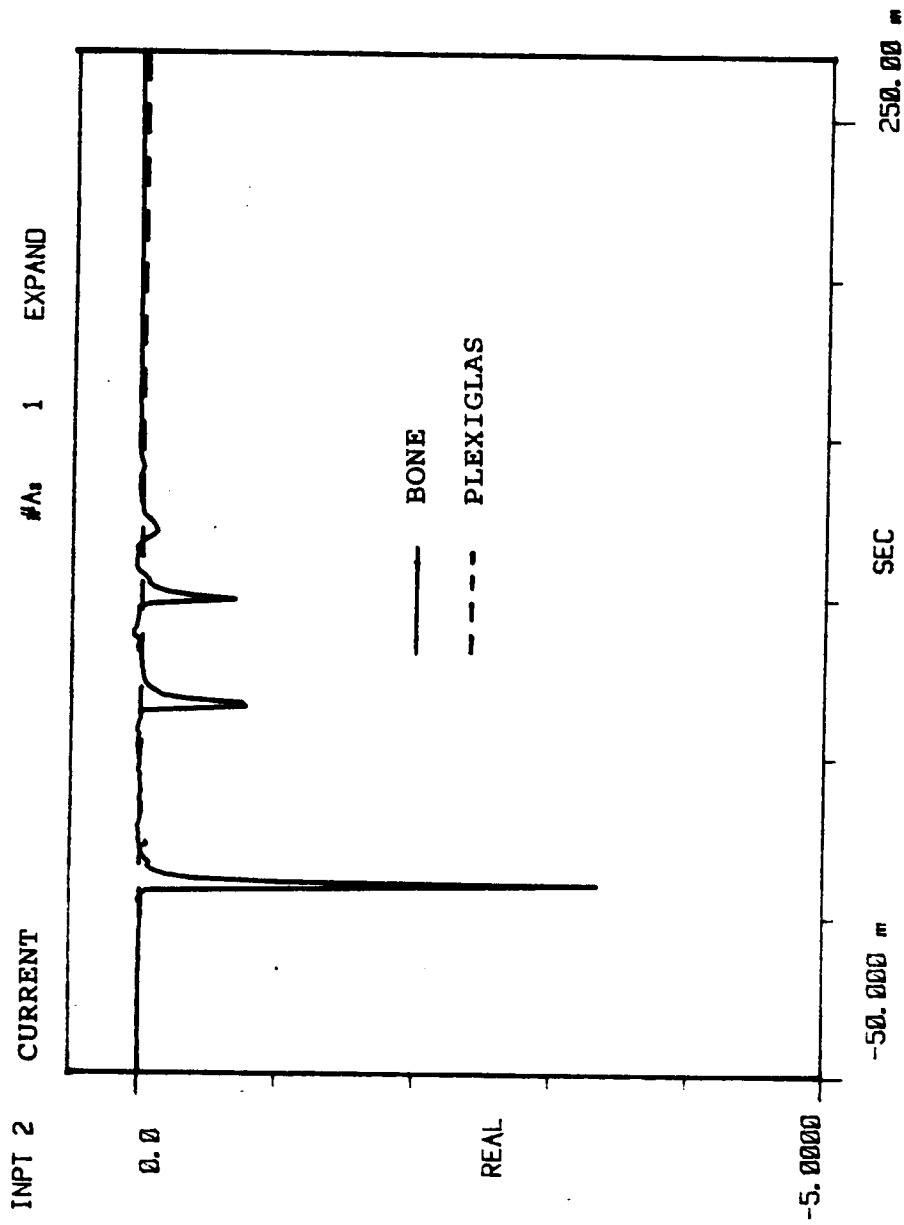


Figure 14. Comparison of Current Responses of Bone and Plexiglas.

bone 2 produced signals that were four times higher than the other two samples. This may be due to a difference in the resistance of bone 2 compared to the other two bones. A lower resistance would allow more leakage of the charge from the plunger to the adjacent electrode. Because of this difference, bone 2 was not used for subsequent testing.

Typical response for the voltage amplifier, shown in Figure 15, had a magnitude of 0.41 volts. Additional response curves are shown in Figure 16 which show magnitudes of response ranging from 0.13 to 0.23 volts. These tests were all done with the metal plunger, and with the plunger guide and the plunger grounded. The range of the voltage response for these examples is not only larger than expected, but it is also not consistent. For the consistent loading that was used, the magnitude of the results were expected to be more clustered. At times the response from the amplifier exceeded the ± 10 volt range of the signal analyzer, a response which far exceeds the expected piezoelectric response of bone based on previous measurements (Simpson, 1980). These excessive voltages are probably due to other factors such as static electricity and to a lesser extent cable motion.

Similarly, typical results for the current amplifier are shown in Figure 17, and additional curves are shown in Figure 18; their ranges are of the magnitudes of response of 300 to 5000 pA. As before, the range of the results is much

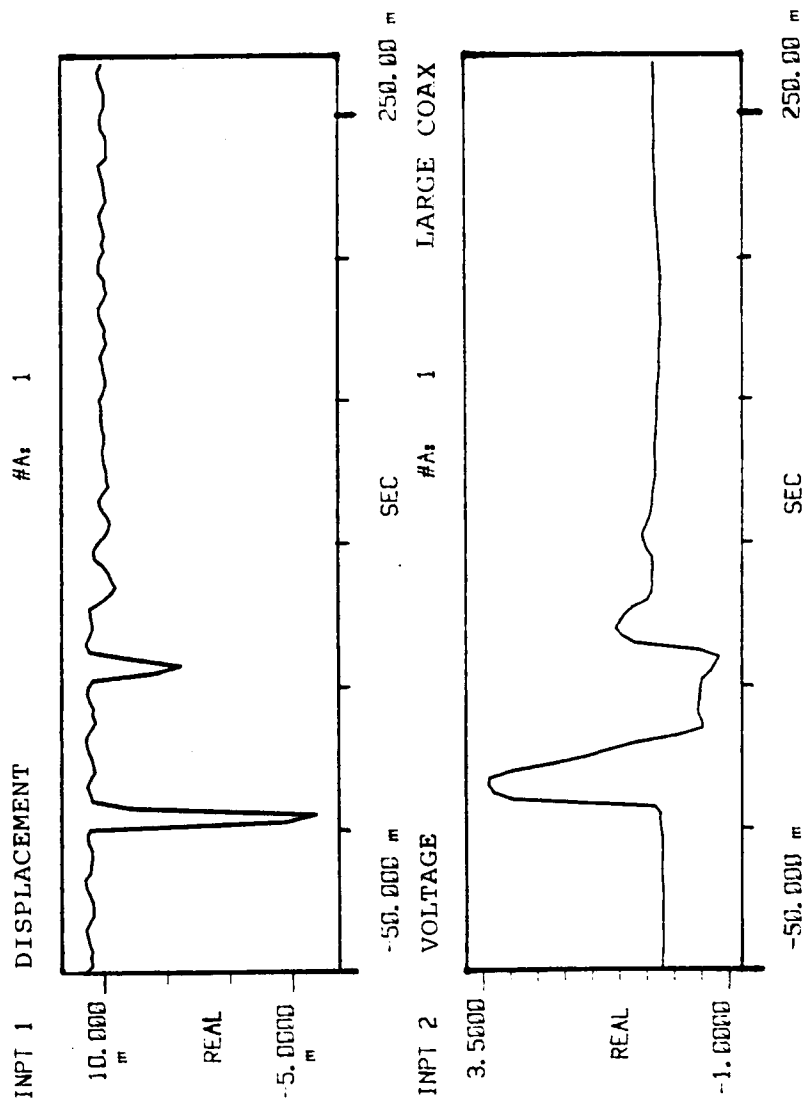


Figure 15. Sample Voltage Waveform Using Grounded Metal Plunger.

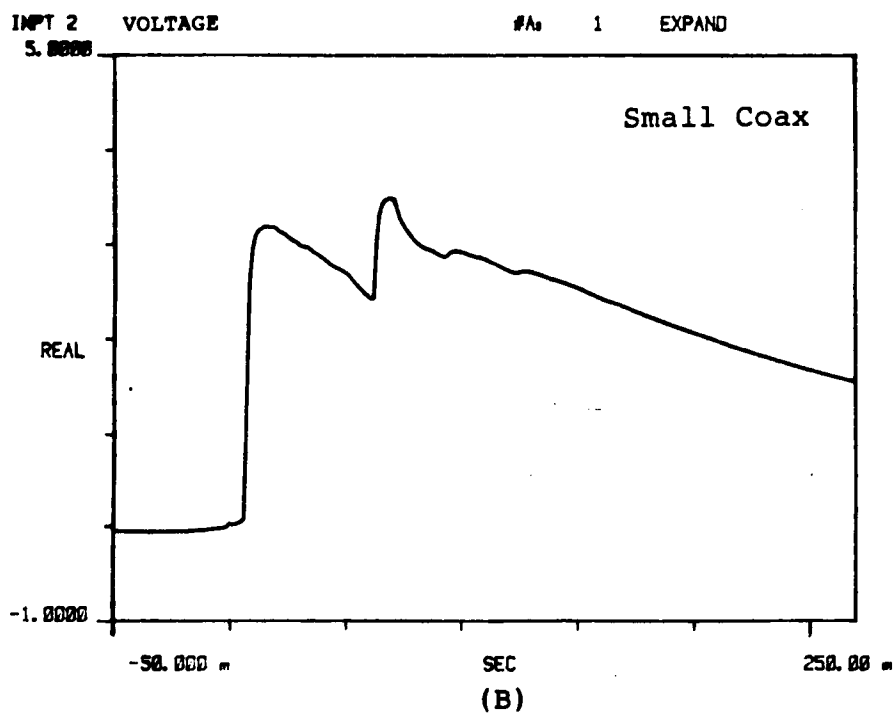
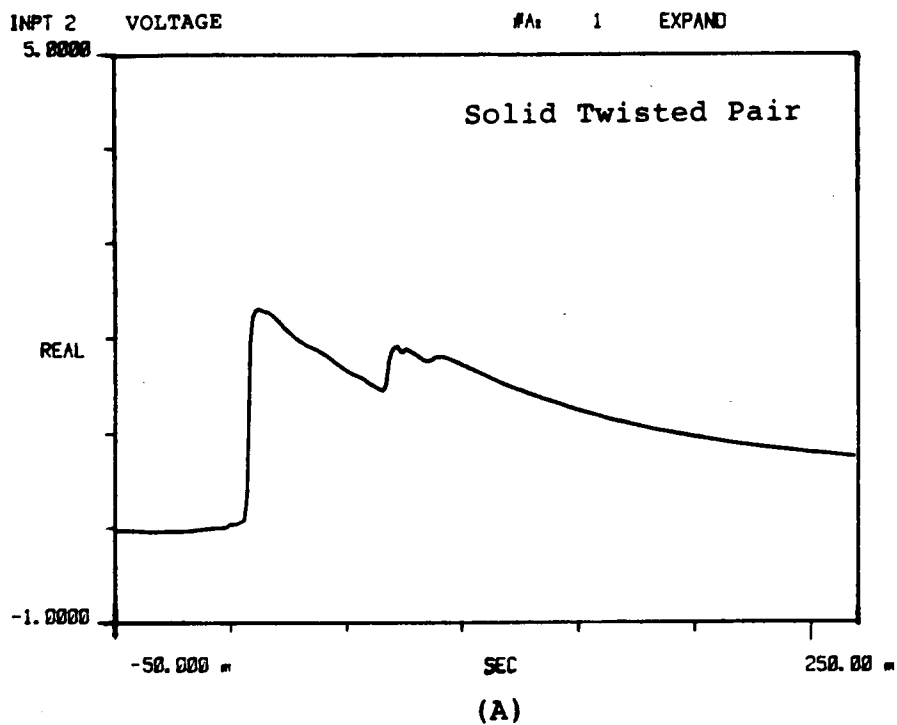


Figure 16. Sample Voltage Waveforms Using Different Connecting Cables and Grounded Metal Plunger.

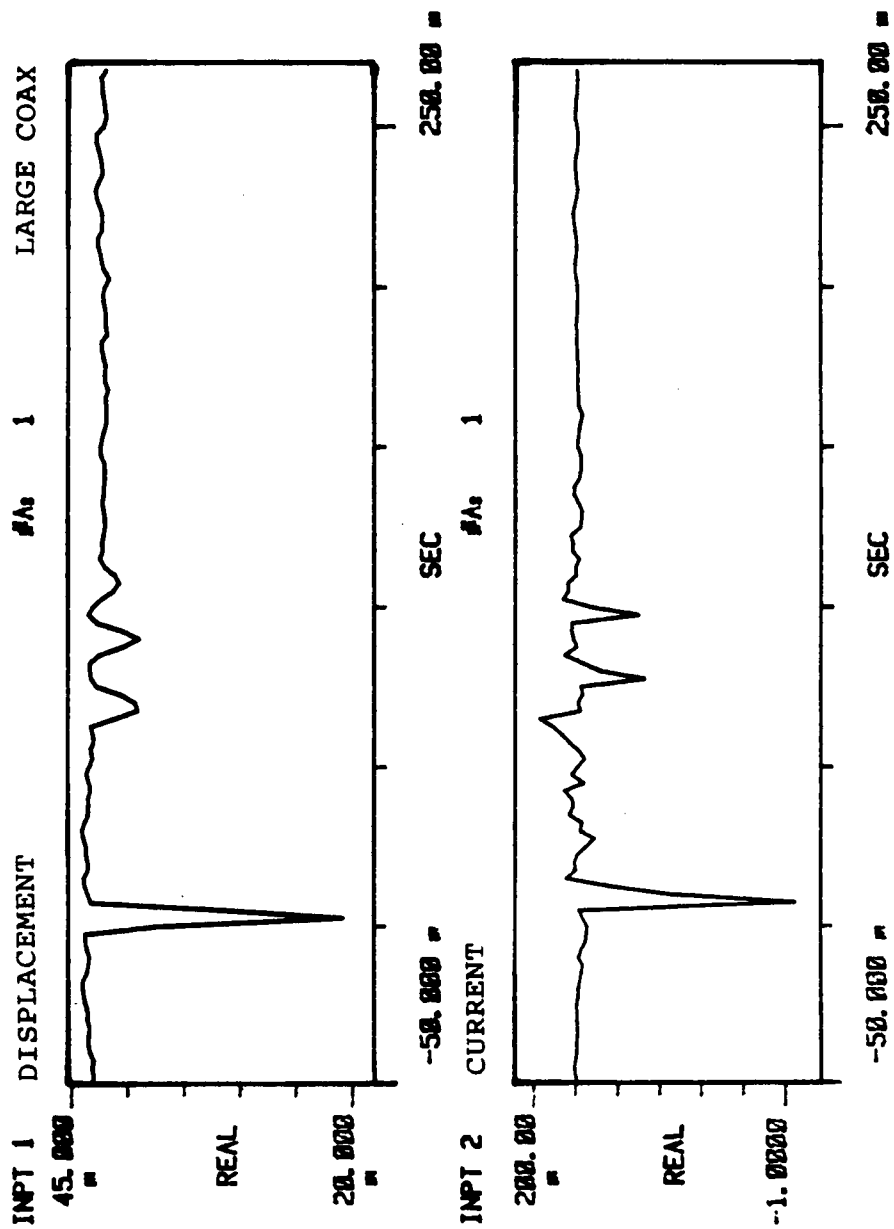
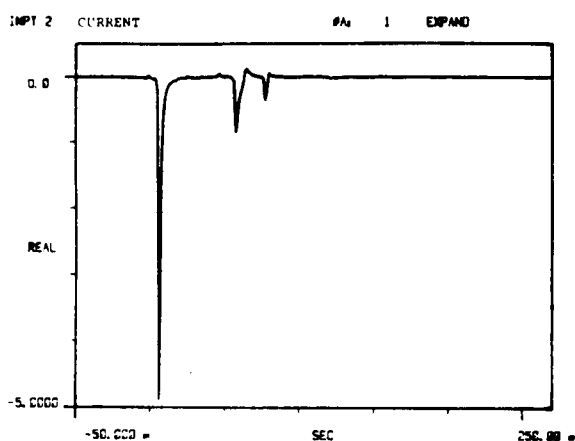


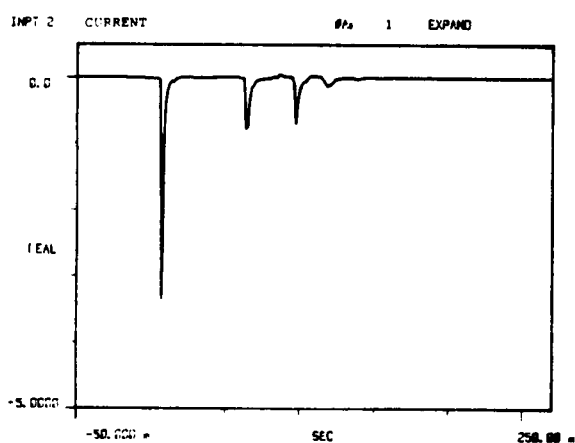
Figure 17. Sample Current Waveform Using Grounded Metal Plunger.



(A)

Plunger and
Guide Grounded

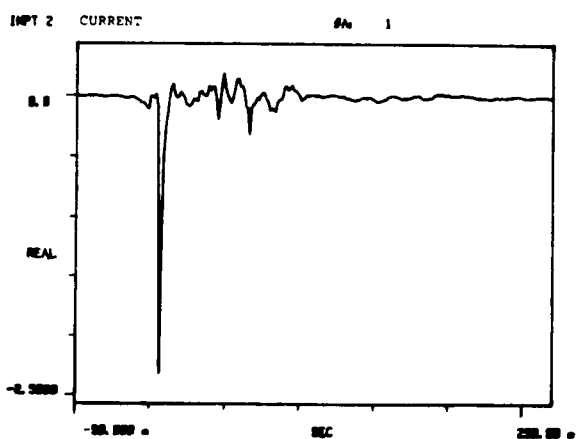
Small Coax



(B)

Plunger and
Guide Grounded

Solid Twisted Pair



(C)

Plunger and
Guide Ungrounded

Large Coax

Figure 18. Sample Current Waveforms Using Different Connecting Cables and Metal Plunger.

too inconsistent, suggesting other factors such as static electricity.

Results from tests in which the weight was reduced from 35 grams to 17 grams gave further evidence that the response seen was not due to piezoelectric response. For these tests, averaging over 10 tests was done to reduce the effects of sample to sample error. The results showed the same 0.73 volt response for both weights, and the current increased from 5000 pA for the 35 gram plunger to 7000 pA for the 17 gram plunger. This is completely inconsistent with the expected linear response from bone reported by other researchers (Cochran, 1966; Cochran et al., 1968; Guzelsu, 1978), which again lends support to the conclusion that the measured response was probably due to static electricity and not to a piezoelectric response.

VII. FREQUENCY RESPONSE

Since analysis by Simpson (1980) had shown that the frequency response of the voltage amplifier was limited by the source impedance of bone, tests were done to determine the actual measured frequency response range of the voltage amplifier, current amplifier, and optical probe in actual use. The frequency responses of all three can be seen in Figures 19, 20, and 21, where the upper trace shows the time response of the signal and the lower trace shows the linear frequency spectrum of the signal. The results show that the

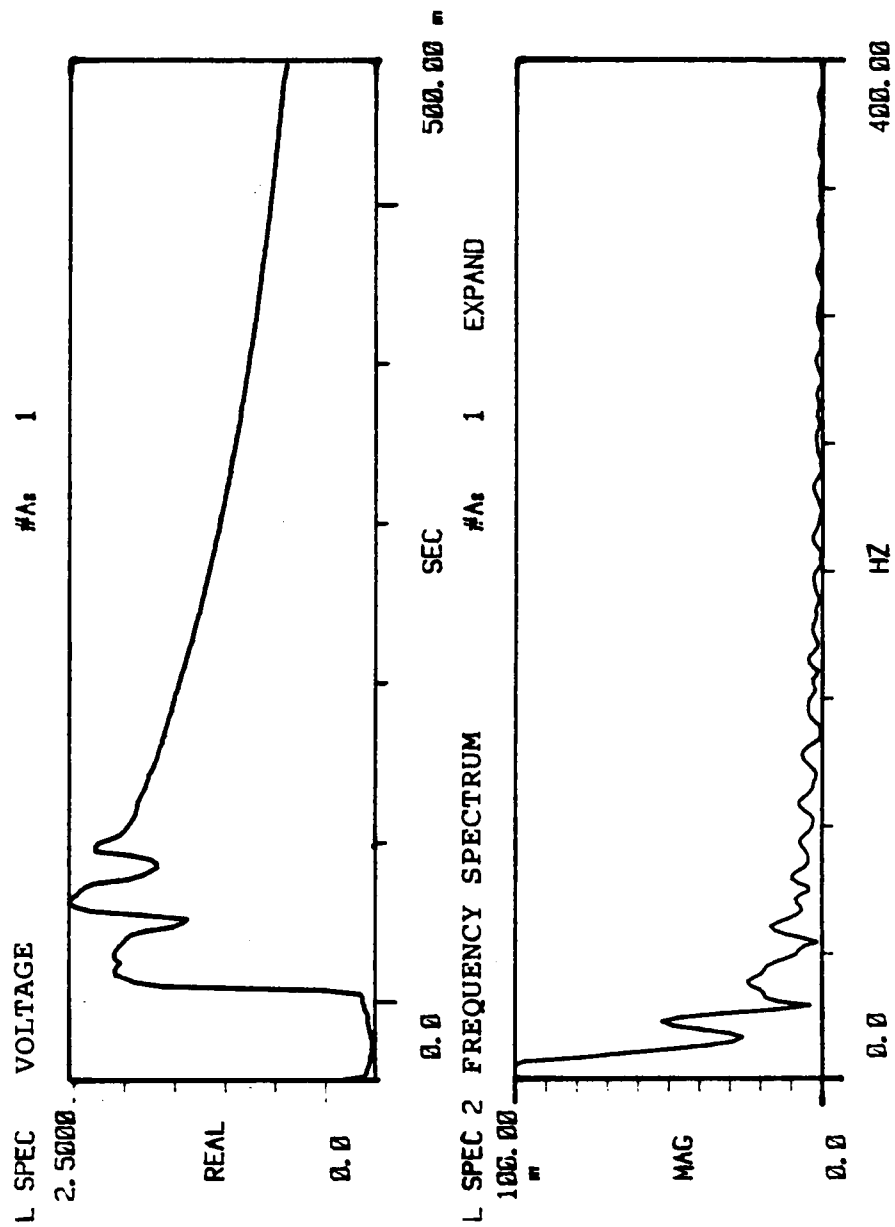


Figure 19. Frequency Response Spectrum of Voltage Amplifier During Impact Test.

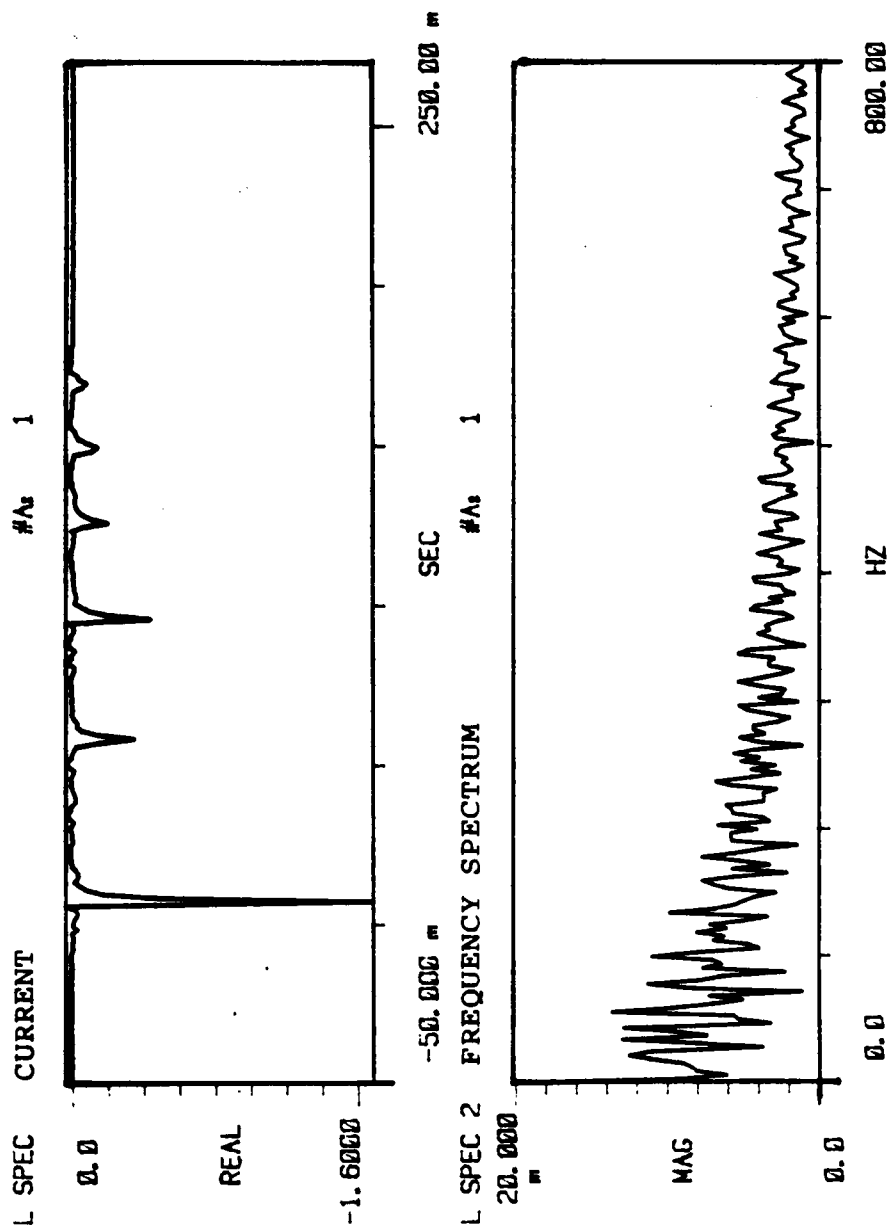


Figure 20. Frequency Response Spectrum of Current Amplifier During Impact Test.

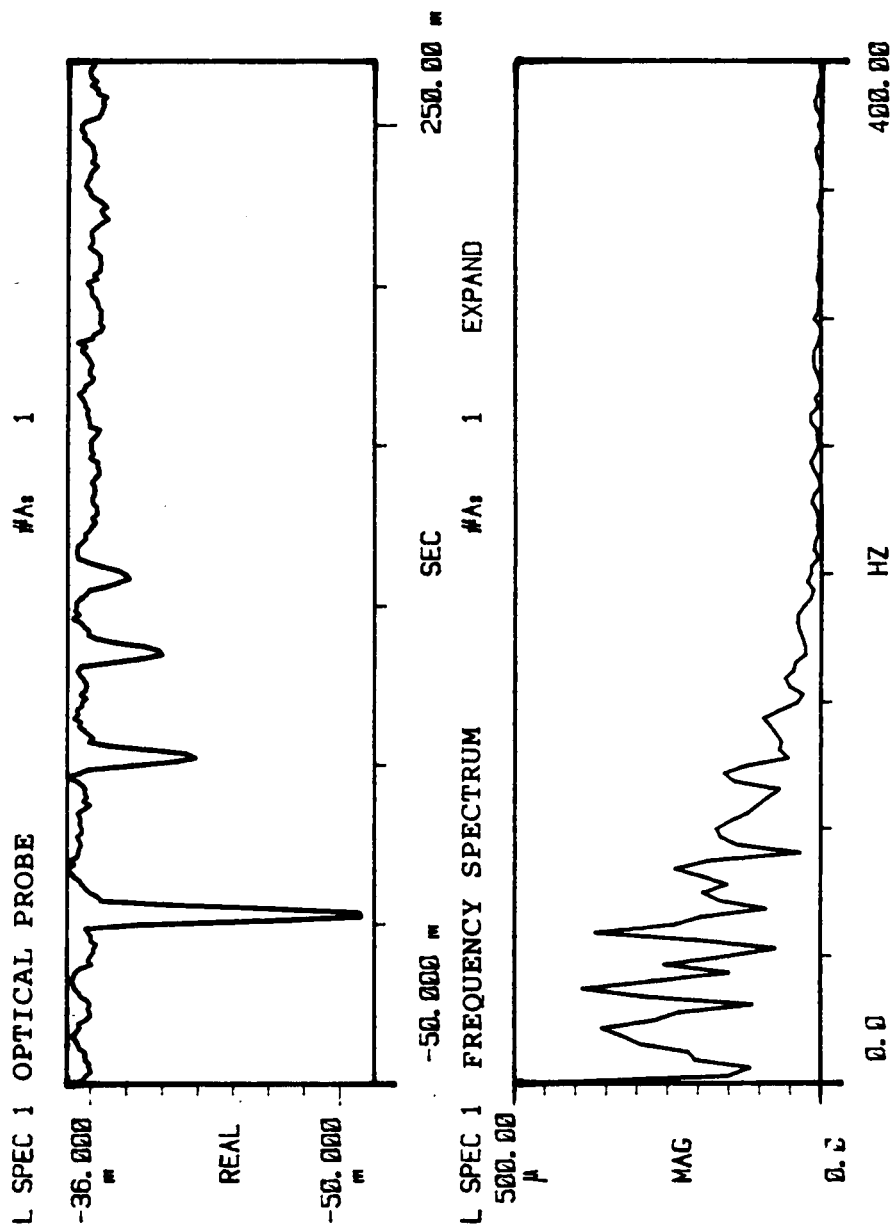


Figure 21. Frequency Response Spectrum of Optical Probe During Impact Test.

frequency range of the current amplifier is two to three times larger than the frequency range of the voltage amplifier, (up to 500 Hz for the current amplifier and up to 200 Hz for the voltage amplifier). The frequency range from the optical probe was about 200 Hz, which should have reflected the frequency response of bone, since the optical probe has a frequency response of greater than 100 kHz. Since the motion response of the bone was measured at the very tip of the free end of the bone versus the electrode position, the mass of the plunger resting on the tip may have changed the motion of the bone such that the frequency range was less than the frequency range from the current amplifier.

VIII. PHOTOELECTRIC EFFECT

Since an optical light probe was being used to measure mechanical response of bone, tests were done to determine if the use of the light from the optical probe would have any effect on the voltage or current testing. The light from the optical probe produced a 2 mm diameter spot of light on the tip of the free end of the bone in an otherwise dark box. The light was turned off and time allowed for the bone to become stable before testing (there was no change in the response of the bone during this time). The light was then turned on for two seconds, turned off for two seconds, and turned back on. There were no detectable changes in the

signal from either the current amplifier or the voltage amplifier due to this action.

To examine the possibility of a direct photoelectric effect from a more diffused light source, a high intensity halogen light source was used to illuminate the entire bone from the underside, and again, the light was turned on, off, and then on again. As with the smaller spot of light, there was no detectable changes in the signal from either the current amplifier or the voltage amplifier due to a diffused light source.

Finally, to determine if there would be any photoelectric effect on the dynamic response of the bone, normal impact tests were performed using bone with the optical probe set for maximum intensity, 20% of maximum intensity, 2% of maximum intensity, and turned off. There was no change in response of either the voltage or current amplifiers due to any intensity changes of the light, (from none to maximum power), from the optical probe.

IX. DROPPED WEIGHT IMPACT LOADING

The results of testing (see Figures 22 and 23) with weights dropped from a string (unlike the plunger tests) are typical of the results obtained by Simpson (1980). As others have reported for voltage measurements, the concave portion of the bone became negative while the convex portion became positive. Signal levels are similar to those

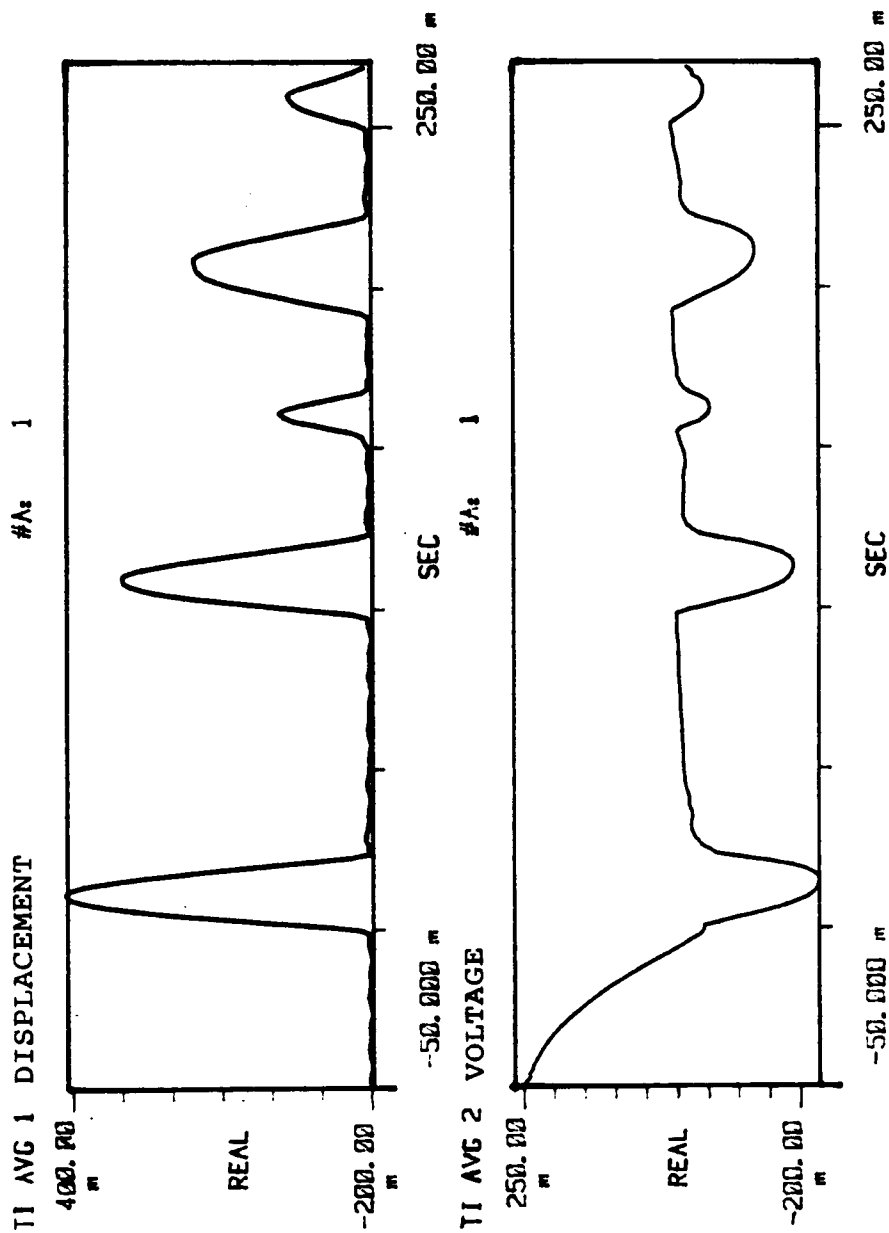


Figure 22. Sample Voltage Waveform from Drop Test Using a 35g Weight.

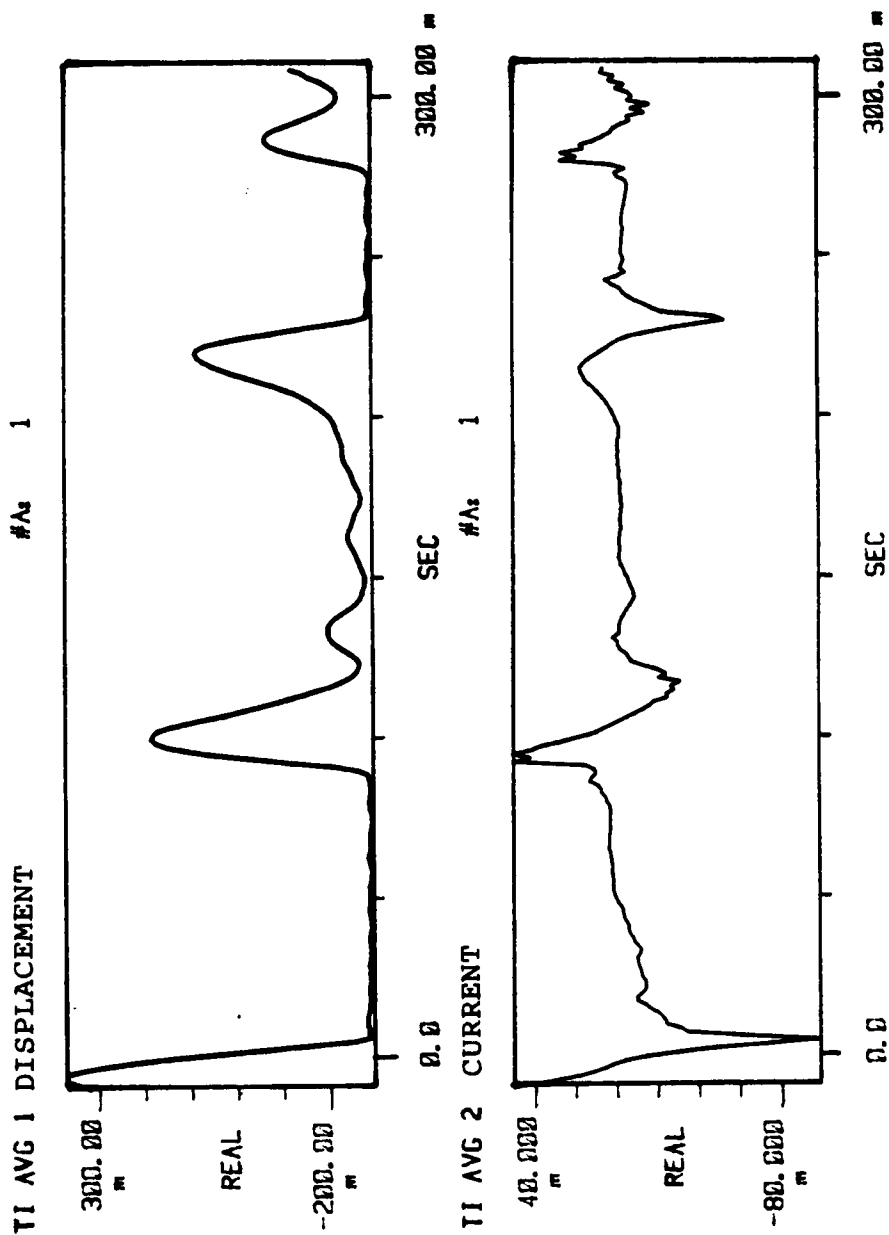


Figure 23. Sample Current Waveform from Drop Test Using a 35g Weight.

reported by Simpson and others (Cochran et al., 1968). The current waveforms are also similar to those reported by Simpson. During testing, a ringing signal was noticed during short segments of several tests (Figure 24), which may have been caused by the natural resonant vibration of the bone subject to cantilever bending. The frequency of this ringing ranged from 1000 to 2000 Hz, and occurred in both the current and voltage waveforms (though much attenuated in the voltage waveforms). The calculated resonant frequency of the bone samples is 2.2 kHz (see Appendix B). When the weights were dropped, they normally bounced several times (Figure 22). Between bounces, the bone sample were free to vibrate at resonant frequencies with only slight damping. Some damping (which will lower the resonant frequency) may have occurred due to the electrodes, the electrode leads, the attachment of the string (which was not in tension between bounces), and possible mass effects of the above.

X. STEP LOADING

The results of the step loading experiment can be seen in Figures 25 and 26. the mono-phasic nature of the voltage response and the bi-phasic nature of the current response can clearly be seen in these figures. Signal levels are consistent with previous results. However, when the same test were done with plexiglas, signals of about 1/2 the

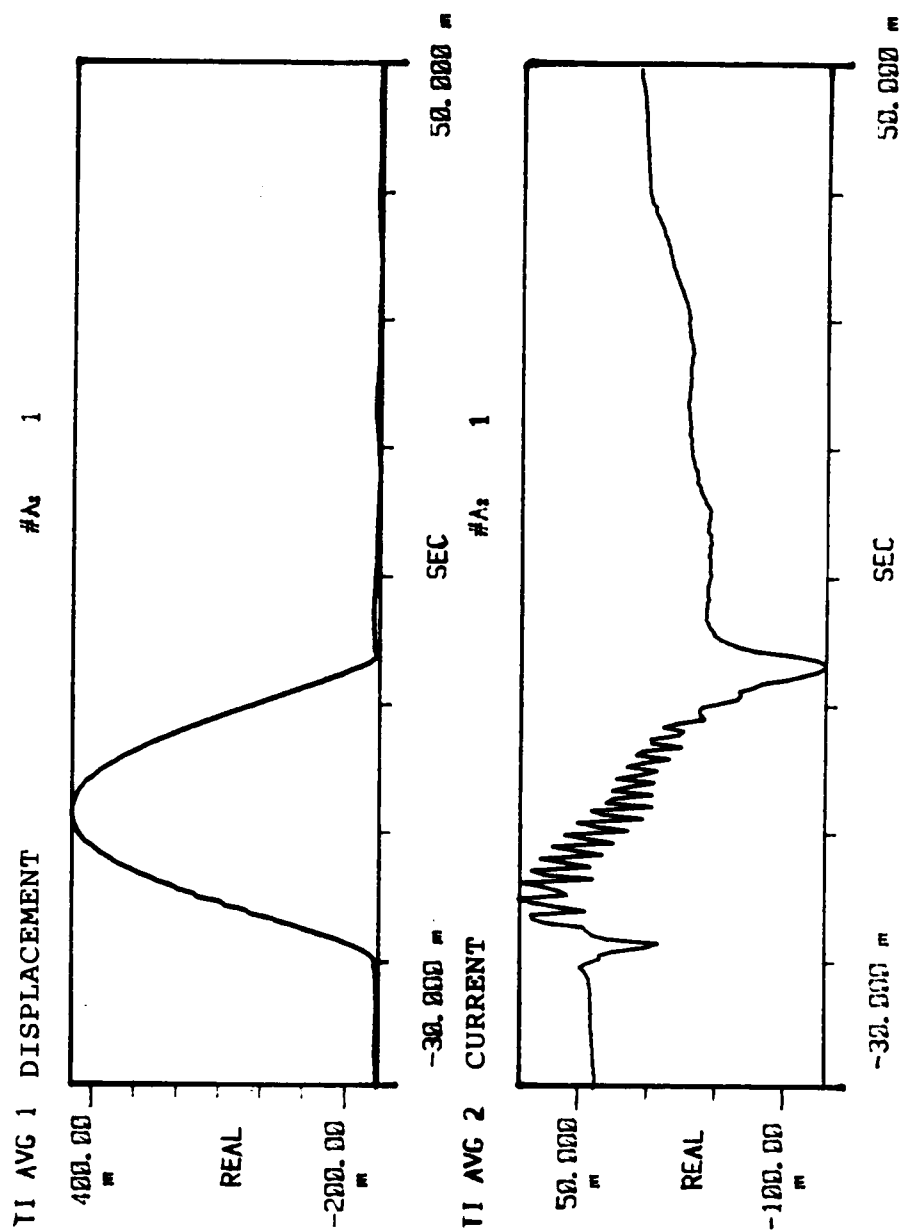


Figure 24. Sample Current Waveform from 35g Drop Test with Oscillation.

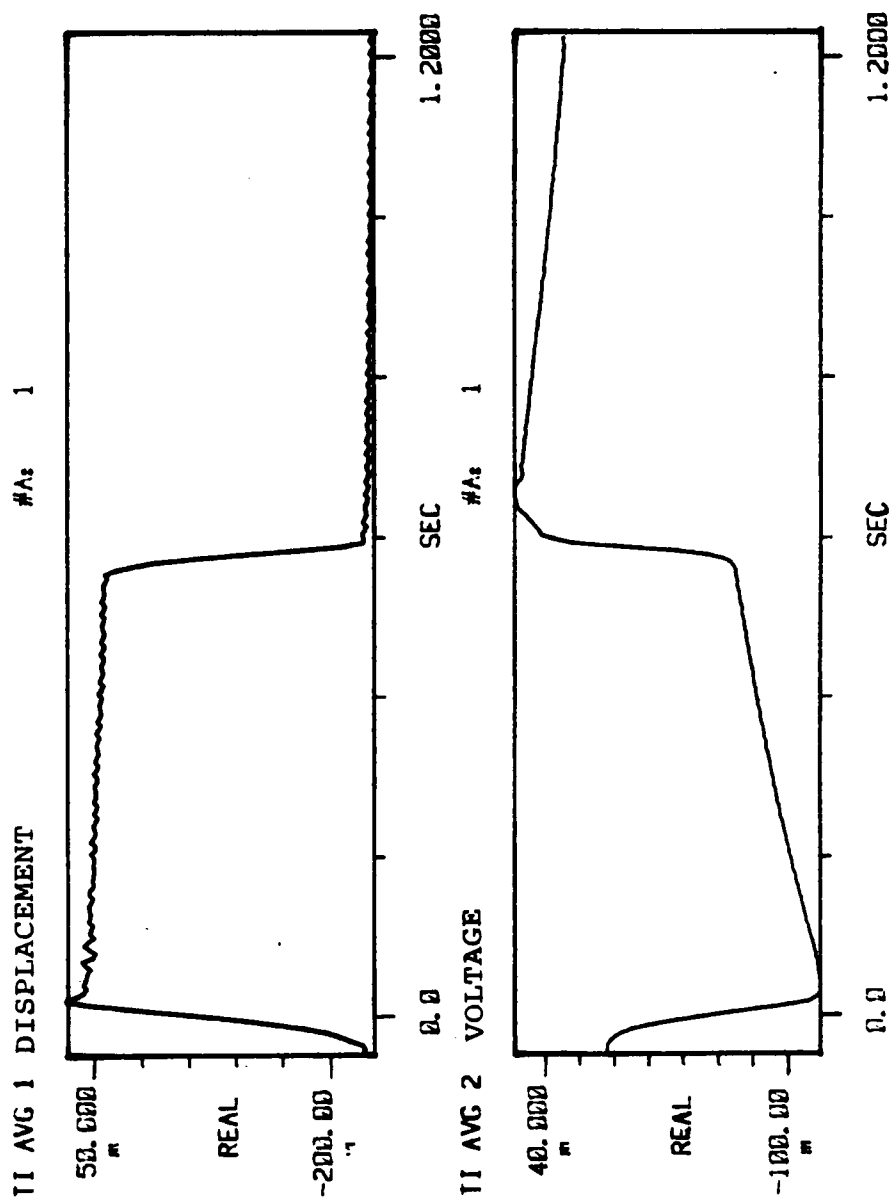


Figure 25. Sample Voltage Waveform from Bone Step Loading Experiment.

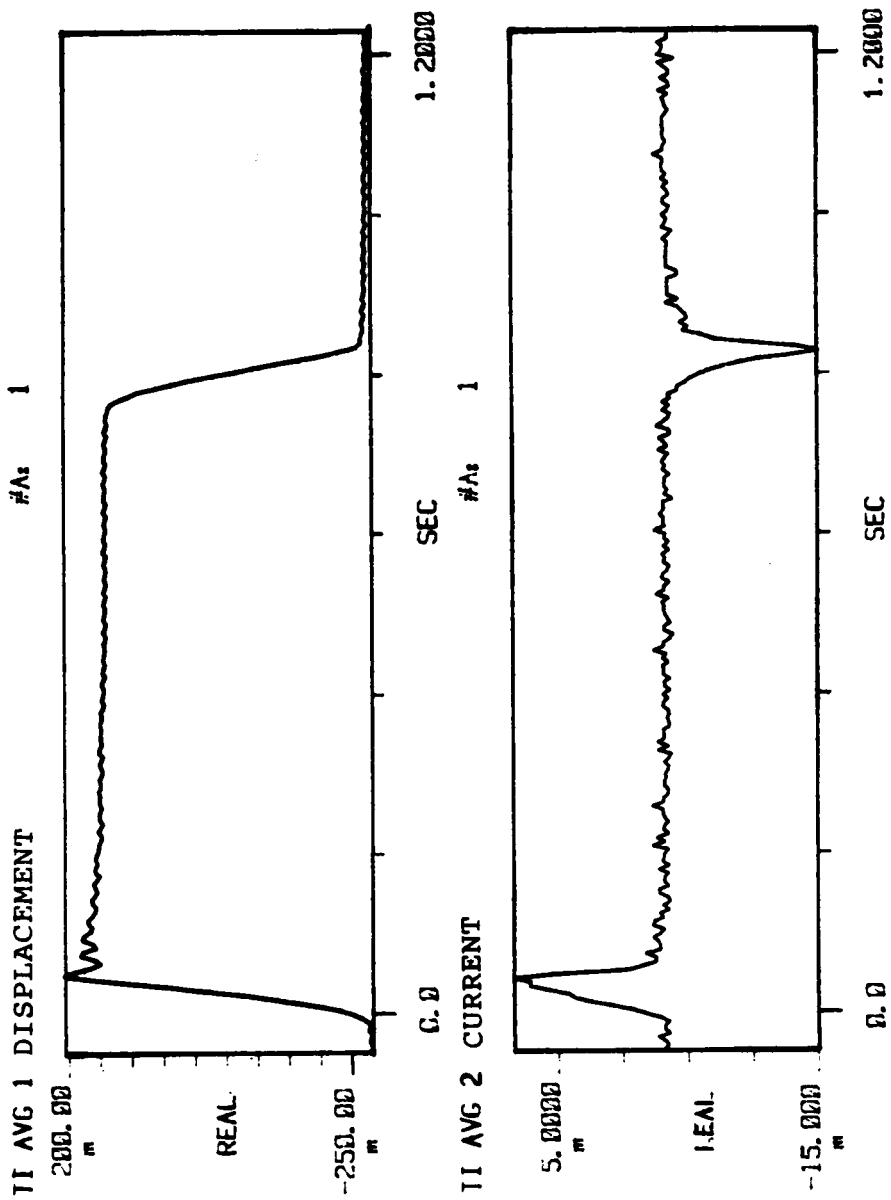


Figure 26. Sample Current Waveform from Bone Step Loading Experiment.

magnitude of those obtained from bone were recorded (Figures 27 and 28). Even with the electrode leads disconnected from the bone, similar signals were obtained. These artefact signals resembled closely the expected signals from the bone which cast doubt on the validity of signals from bone.

XI. SEPARATION OF SIGNAL FROM ARTEFACT

Signal processing did separate the noise signals from the signals from the bone. The voltage response from ten impact tests were collected and averaged (Figure 29). The electrode leads were then reversed and the results from ten more tests were collected and averaged (Figure 30). These signals were then processed by adding or subtracting them (see Appendix C) and the signals from the bone were separated from the vibration induced noise signals (Figures 31 and 32). The same procedure was repeated for the current response (Figures 33, 34, 35, and 36). As can be seen, from Figures 31 and 35, the noise response of the cable is very similar to the response of the bone. The signals obtained from the bone are typical of previously reported results both in terms of the waveshape and the magnitude of the response.

To test for artefact signals, the same procedure was done using the plexiglas "bones." The final results can be seen in Figures 37 and 38. These signals are not as "clean" as the signals obtained from bone and they are much smaller.

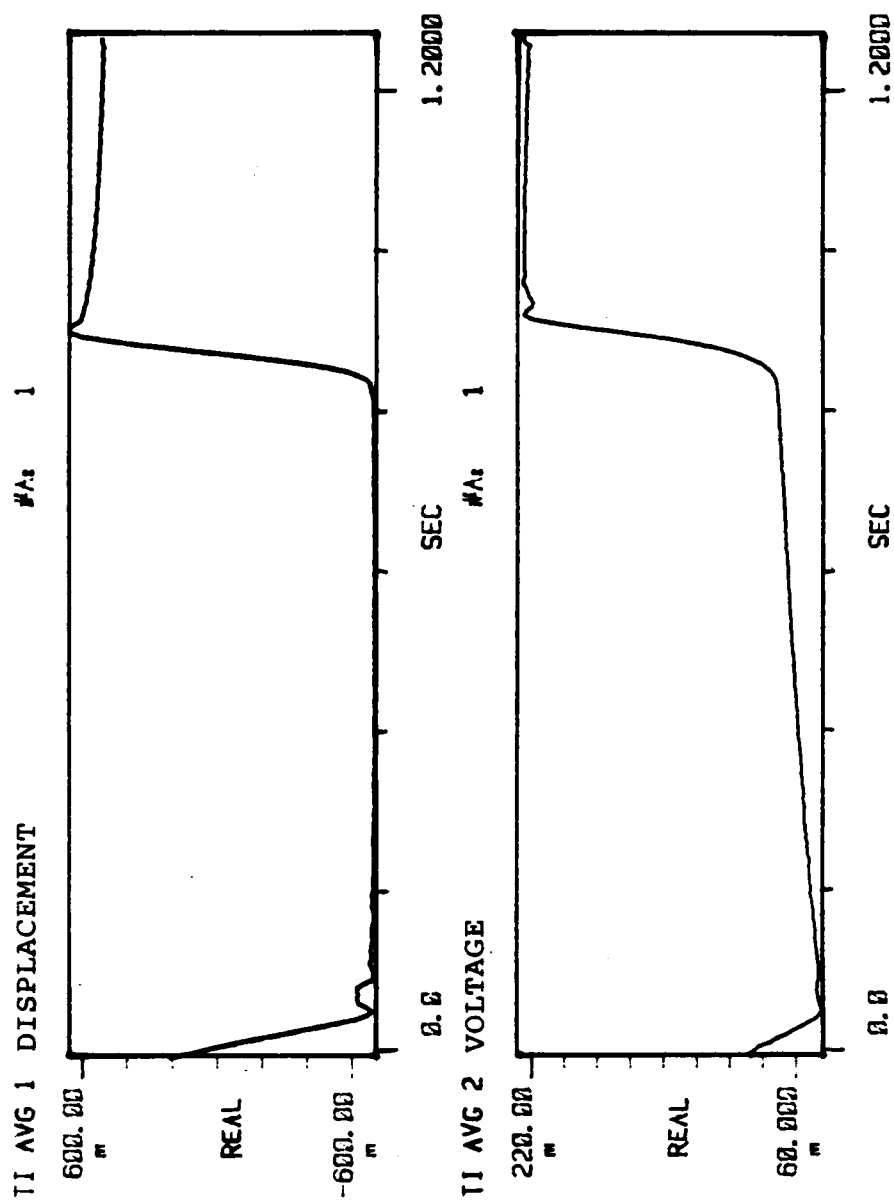


Figure 27. Sample Voltage Waveform from Plexiglas Step Loading Experiment.

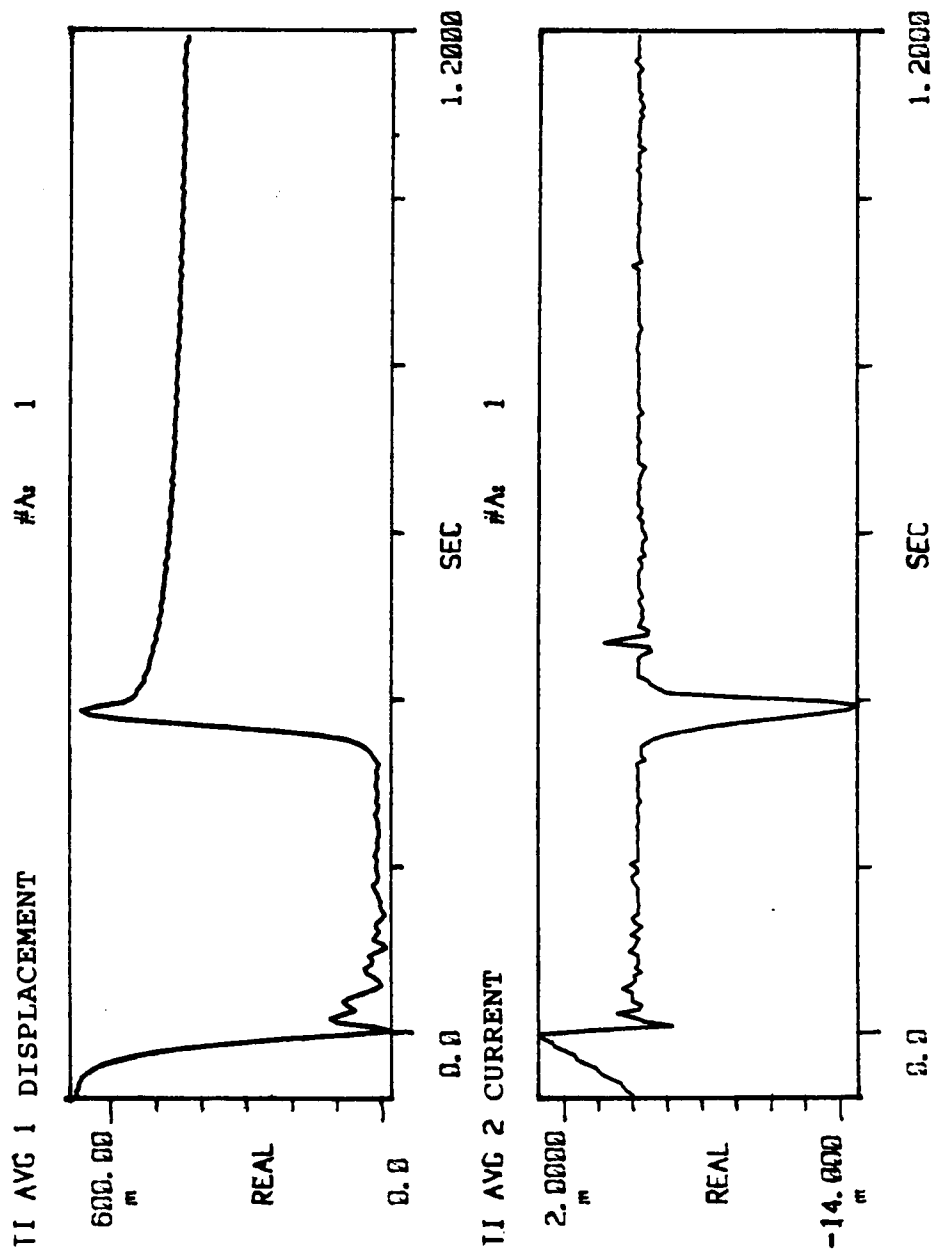


Figure 28. Sample Current Waveform from Plexiglas Step Loading Experiment.

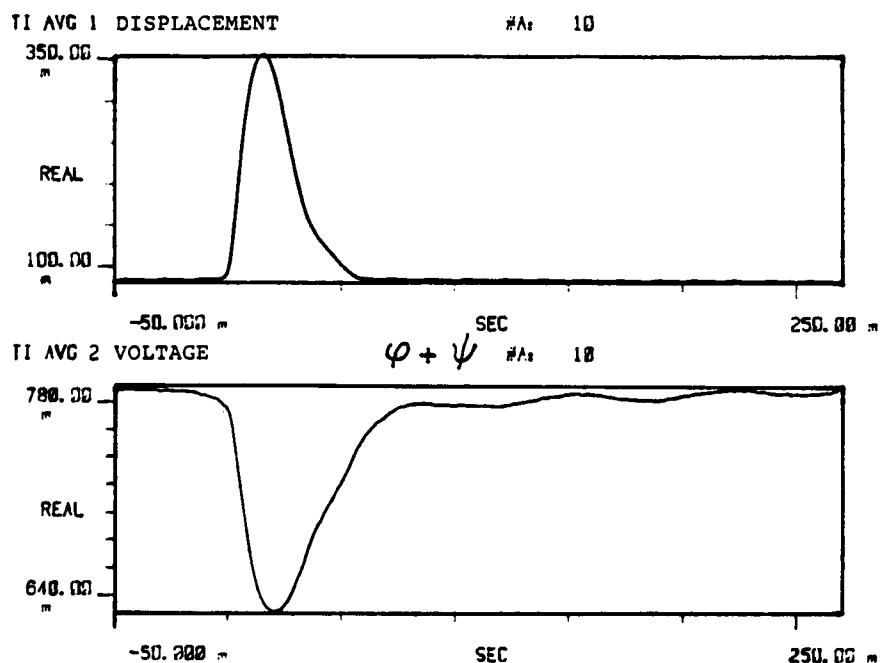


Figure 29. Average of 10 Impact Test of Bone Voltage Response with Normal Polarity at Electrode Leads.

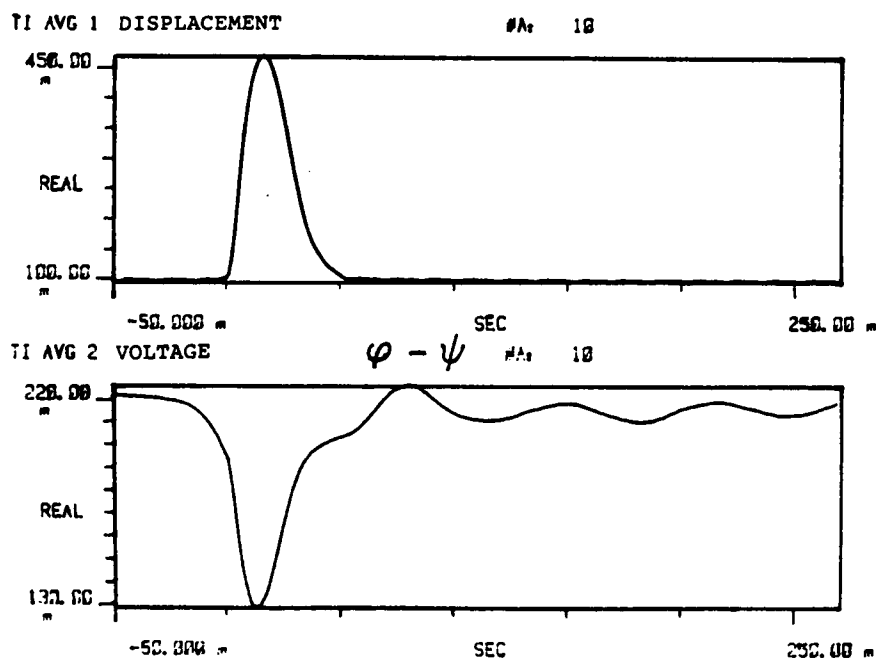


Figure 30. Average of 10 Impact Test of Bone Voltage Response with Reverse Polarity at Electrode Leads.

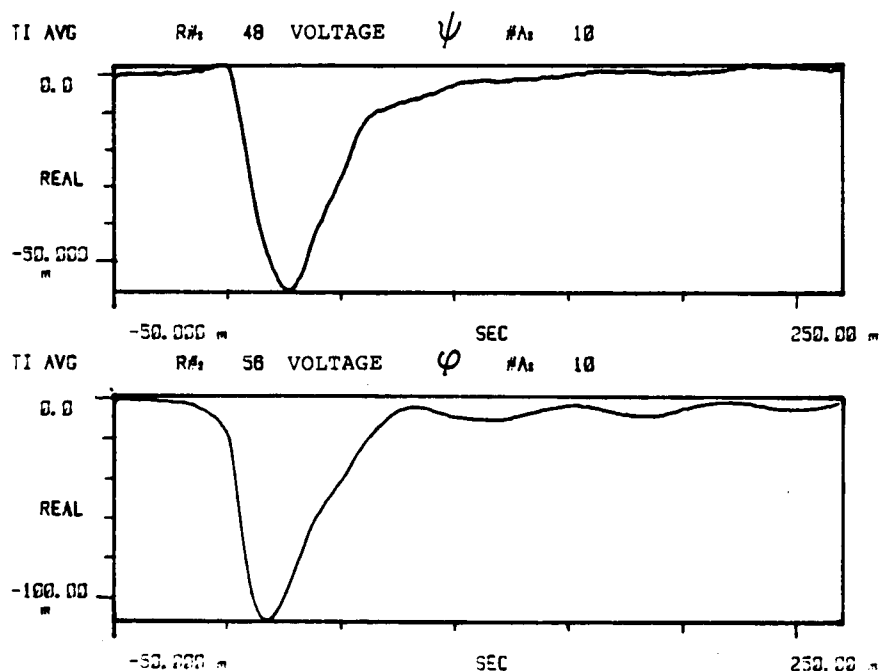


Figure 31. Results of Signal Processing to Separate Voltage Response of Bone (ψ) from Cable Response (ϕ).

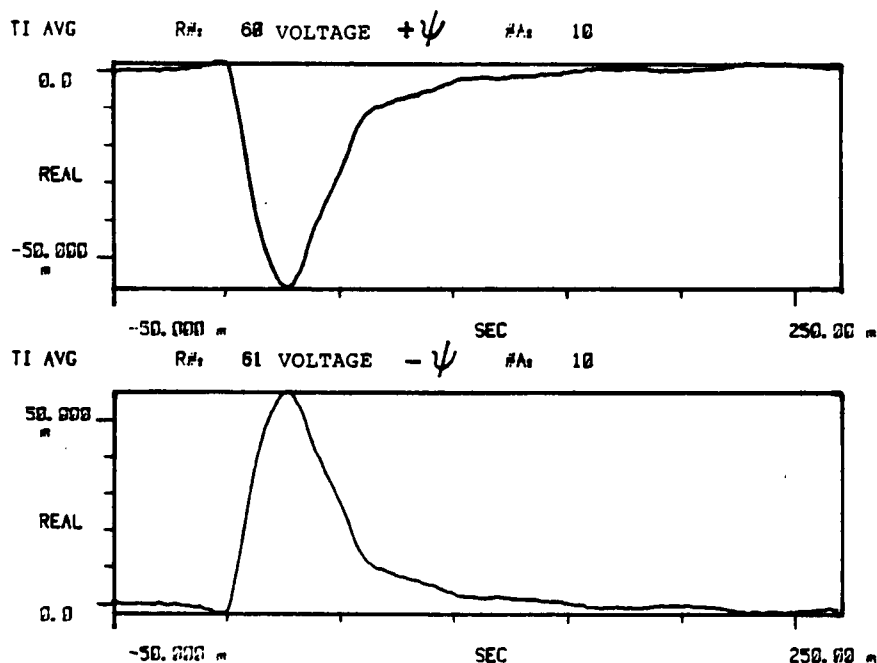


Figure 32. Results of Signal Processing to Separate Normal Polarity of Bone Voltage Response ($+\psi$) from Reverse Polarity of Bone Voltage Response ($-\psi$).

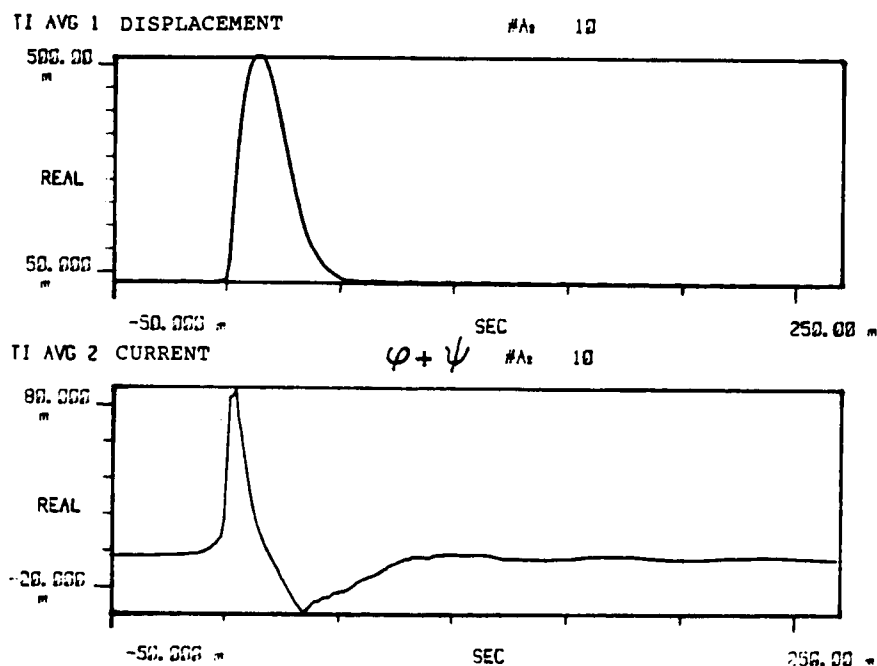


Figure 33. Average of 10 Impact Test of Bone Current Response with Normal Polarity at Electrode Leads.

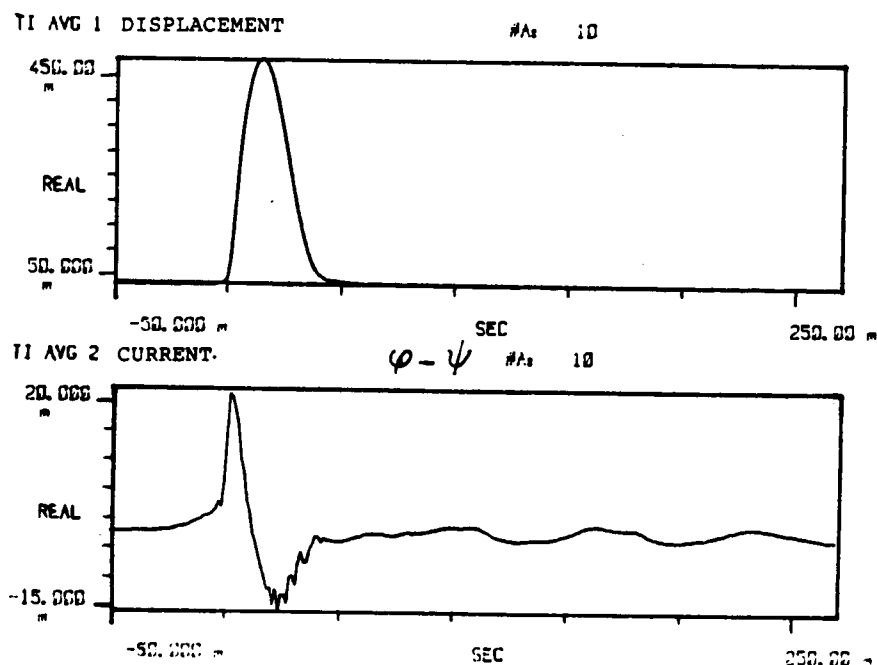


Figure 34. Average of 10 Impact Test of Bone Current Response with Reverse Polarity at Electrode Leads.

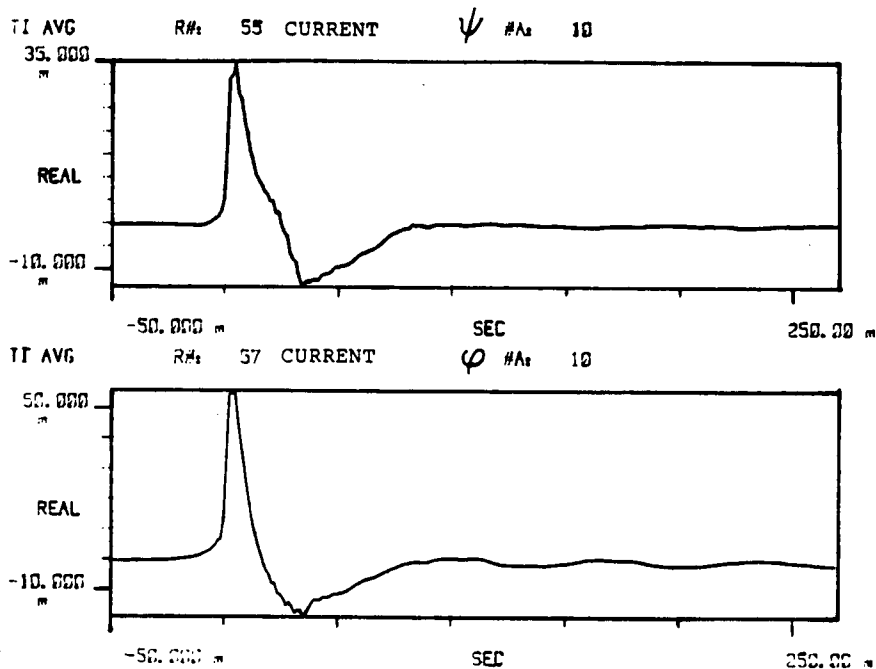


Figure 35. Results of Signal Processing to Separate Current Response of Bone (ψ) from Cable Response (ϕ).

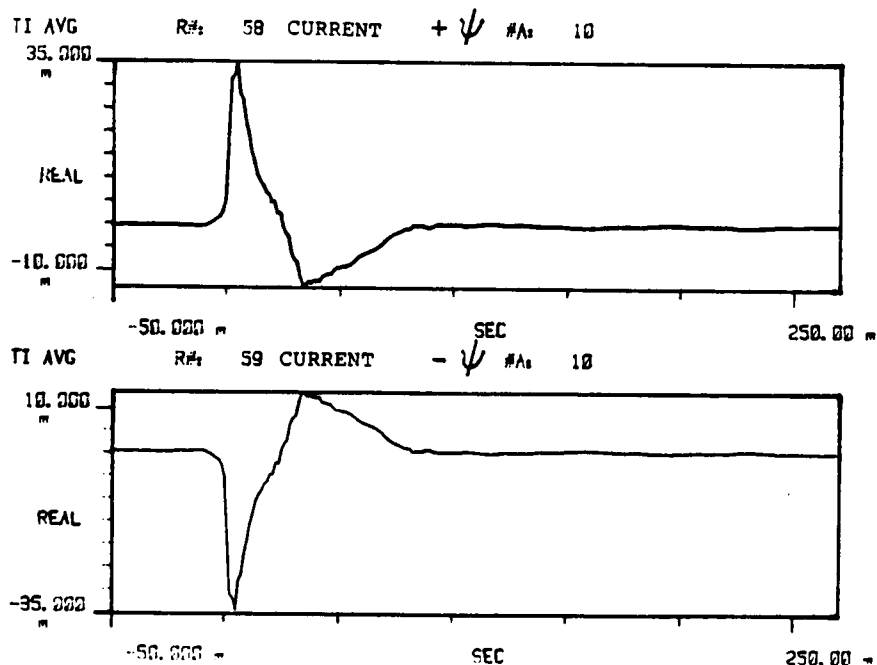


Figure 36. Results of Signal Processing to Separate Normal Polarity of Bone Current Response ($+\psi$) from Reverse Polarity of Bone Current Response ($-\psi$).

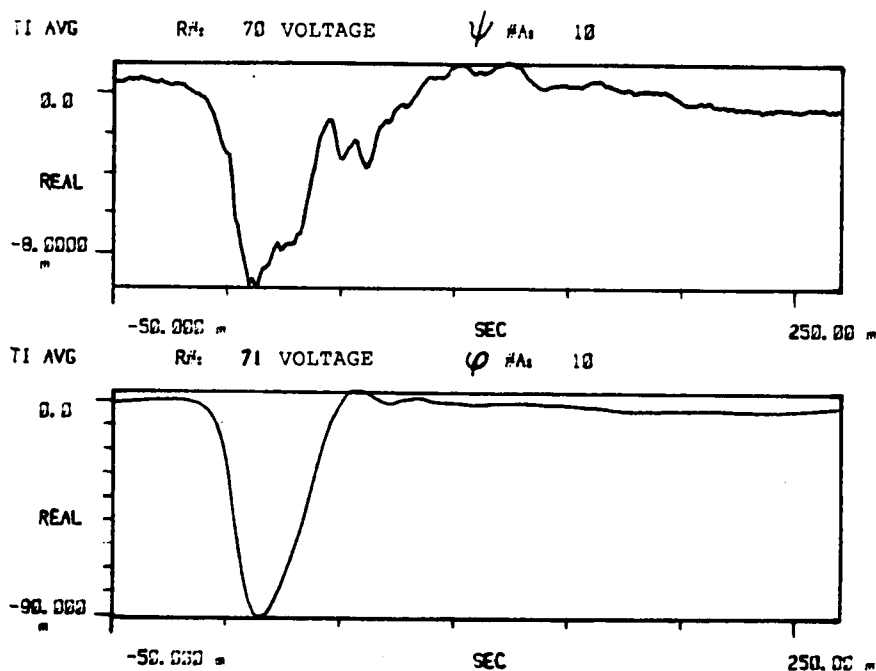


Figure 37. Results of Signal Processing to Separate Voltage Response of Plexiglas (ψ) from Cable Response (ϕ).

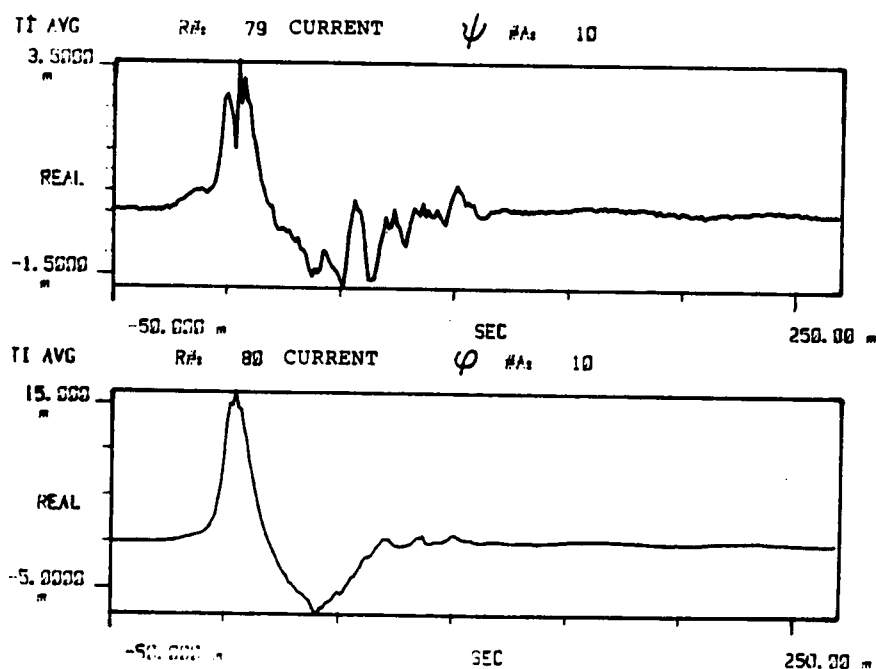


Figure 38. Results of Signal Processing to Separate Current Response of Plexiglas (ψ) from Cable Response (ϕ).

For the voltage response of bone, the piezoelectric signal was $1/2$ as large as the artefact signal and had a magnitude of 4.5 mV whereas for the plexiglas, the piezoelectric signal was $1/15$ as large as the artefact signal and had a magnitude of only 0.5 mV. For the current response of bone, the piezoelectric signal was $3/4$ as large as the noise signal and has a magnitude of 45 pA whereas for the plexiglas, the piezoelectric signal was $1/4$ as large as the noise signal and had a magnitude of only 5 pA. While there does appear to be some signal from the plexiglas, it is very small and of poor quality and may be due to processing error, such as not having enough data to obtain a statistically significant average or not having signal levels properly matched.

CHAPTER VI

DISCUSSION

This chapter contains general discussion based on the results obtained from the current and voltage waveforms experiments. Limitations of the measurement system due to static electricity and vibration of connecting cables are also discussed. Finally, some recommendations for further research are presented.

I. OVERVIEW OF EXPERIMENTAL RESULTS

The data obtained during the tests involving the plunger clearly indicate the high sensitivity of the system to any sort of external interference, in particular static electricity and vibration. Signals were obtained from the testing of bone that were typical of signals obtained in previous work by other researchers. However, signals were also obtained from plexiglas and coax cable alone which looked like typical signals that other researchers have obtained for bone. In the past, if other researchers have tested their systems for artefact signals, they have not reported the results.

The purpose of this research was to develop a highly sensitive measurement system for the measurement of small signals in high impedance environments and then to test for artefact signals. Such signals were found. The magnitudes

of the signals varied from levels reported previously for bone samples, to levels which were 10 to 20 times higher than expected. The signals produced came from three main sources,

- 1) static electricity,
- 2) mechanical vibration of the cabling, and
- 3) electrical response of mechanically stressed bone.

While not measured, the moisture content in the air and to a lesser extent the level of moisture and chemical contamination on the testing jig surfaces will significantly alter the effects of static electricity. Since other test conditions were held constant, static electricity seemed to account for most of the variability in the observed signals.

Since static electricity could not be effectively controlled using plungers for the testing, the most physiologically meaningful results were obtained from drop testing of weights to strings attached to the ends of the bones. This method eliminates the static problem but does not eliminate the vibration problem. Again, since the signals produced by the vibration of the cables is similar to the signals produced by mechanically stressed bone, it is difficult to determine the source of the signals. The electrode leads were reversed during some of the testing and combined with data collected during normal testing to isolate the signals produced by mechanically stressed bone from the signals produced by the mechanical vibration of the cabling.

These results show that there is an electrical response from mechanically stressed bone and that this response is not due to artefact signals. However, the results also indicate artefact signals can be produced which appear to be response due to mechanically stressed bone. This potential problem has not been discussed in the literature and should be further investigated in future research.

II. STATIC ELECTRICITY

Static electricity for this test configuration, was the most difficult noise source to eliminate. This was compounded by the high sensitivity of the measurement system. Several fixture modifications were done to reduce or eliminate this problem. The clamp holding the plexiglas bone and the guide for the plunger were made of plexiglas. Plexiglas was chosen because of its excellent insulating properties ($\rho \approx 1 \times 10^{15}$ ohm-cm), but these same properties also make it a good source of static electricity. To reduce the possibility of stored charge on the surface of the plexiglas parts, the surfaces were coated with silver paint in critical areas and grounded, but this procedure did not produce significant change. The major source of static was found to be the movement of the plunger within the plexiglas guide. Grounding the area around the hole in the guide, through which the plunger slid, and grounding the plunger were tried but this was also only

partially successful. The use of an antistatic solution for plastic was found to be ineffective. Because of the close proximity of the plunger to the positive electrode lead (1.5 mm), a small difference in potential between the plunger and the positive electrode lead was detected as a signal. For this case, two different effects can occur. The plunger can be grounded, which causes the positive electrode lead to bleed to ground through the plunger, or the plunger can be insulated and as it slides through the plexiglas guide, it will pick up charge which will bleed to the positive electrode lead. In either case, these effects will significantly alter the measured response. In low impedance circuits this would be much less of a problem, but since the 1.5 millimeter distance between the plunger and the electrode probably represents the lowest impedance path in the circuit, these differences in potential become major components of the signal.

An insulated micarta plunger was used in some trials in an attempt to overcome the conduction effects of the metal plunger. This was not a solution to the problem either, in fact it compounds the problem. Like plexiglas, the micarta plunger is an excellent source of static electricity. Since it cannot be easily grounded, it will store a large charge. Micarta has a high surface resistivity (3×10^9 to 2×10^{16} ohm-cm) but this resistivity is very sensitive to changes in humidity and surface contamination. The behavior of

static electricity on surfaces is not well understood and therefore, makes the problem of controlling static even more difficult if a micarta plunger is used.

When either plunger was used, large amounts of static charge was produced even with small amounts of movement of the plungers. In fact large signals could be observed if the plungers were raised up off of the bone samples only a couple of millimeters and then placed back on the end of the bone sample. This motion was done very slowly and at the instant of contact, the large signals were observed. Based on this finding, the plunger should always be in contact with the end of the bone and the forcing function should be applied to the plunger. By being in constant contact with the bone, there would not be sudden changes in potential between the plunger and the bone which would contaminate the signals. The plunger should be insulated to avoid bleed off of charge and should be guided by some sort of "frictionless" guide. The ideal material for a plunger would be an insulator that would not generate a static charge on the surface due to frictional motion and would not store or accept charge from other surfaces. The use of weights dropped from a string avoids the problem of static electricity, but does not allow for consistent or controlled input functions.

Possibly the best loading method would be the use of a hydraulic shaker attached to the plunger.

III. VIBRATION OF THE CABLING

The second important effect discovered was the sensitivity of the cabling between the electrode leads and the circuit board to mechanical vibration. Coax provides excellent shielding properties, but has problems with vibration, leakage, and capacitance. The bone piezoelectric signals being measured are small signals measured by high impedance circuits which makes the data susceptible to artefact signals which can significantly mask the true signals. This problem became evident during testing of the amplifiers with only the connection coax attached. The signals obtained from the coax were very similar to the signals expected from a piezoelectric response of bone.

By testing in a metal box, the need to shield the signal is reduced, and the use of other types of interconnecting cables can be considered. However, the twisted pair cables used produced noise signals almost identical with the signals obtained from the coax indicating that the type of cabling was not a major factor in reducing the noise signals. Since the noise signals produced corresponded to the impact signals being applied to the bone, the noise signals obtained from the cabling are a response to mechanical vibration of the cable. Since changing to less vibration-sensitive cables and the use of damping materials had no significant effect on reducing the

sensitivity of the cabling to vibration, the best approach to reducing vibration induced noise in the cables is to reduce the length of the cables. The manufacturer of the AD 515 amplifier suggests designing circuits in which the input is connected directly to the source on the signal being measured.

Even though the steel box and holding jigs were fairly sturdy, they were none-the-less still capable of transmitting vibration caused by impacts to the bone. When either the plunger or weight was used to load the bone, the impact could be felt on the outside of the steel box. The holding jigs and circuit board should be mounted in a much more rigid, vibration free manner to reduce motion in the connection cables and the resultant noise signals.

IV. GUIDELINES FOR ADDITIONAL RESEARCH

In designing the next generation system, special care should be given to eliminating interference due to

- 1) grounding problems,
- 2) static electricity problems, and
- 3) mechanical vibration problems.

The solutions to grounding problems are similar to those reported by others (Becker and Brown, 1965; Bur, 1976,; Cochran et al., 1968). Noise due to grounding should be eliminated before additional testing is begun. Improper grounding will result primarily in a relatively large 60 Hz

signal which can be observed at the output of the amplifiers.

Artefact signals due to mechanical vibration can be reduced by rigidly mounting the sample holding jig, associated electronics and cabling. The length of the cabling should be as short as possible. One possibility for eliminating the problem of cable motion would be to design a "hybrid" circuit that could be mounted directly on the sampling holding jig, eliminating the need for lengthy connecting cable between the bone and the circuit board. Since vibration induced signals look like piezoelectric response signals, care must be taken to isolate and eliminate these signals. The plexiglas "bone" should be used for testing not only to identify vibration induced artefact, but also to verify the response from bone. Reversing electrode leads from the bone is also instrumental in identifying artefact, especially artefact from the cabling, because signals derived from bone will have a reversed polarity whereas artefact signals from cabling will have a constant polarity.

Problems with static electricity may be difficult to distinguish from mechanically-induced signals. The most distinguishing characteristic of static electricity signals is their wide magnitude variation, which sometimes exceeds the magnitudes of normal signals by an order of magnitude. The variation can occur over several days and is probably

due to changes in humidity of ambient air. Static signals also differ from mechanically-induced signals by being independent of the magnitude of the force applied to the bone. Over a short period of time (about an hour), the magnitude of the static response will remain fairly constant under the following conditions:

- 1) very slow movement of the plunger,
- 2) very fast movement of the plunger,
- 3) bringing the plunger in contact with the bone (or plexiglas),
- 4) increasing (or decreasing) the force (weight, velocity) of the plunger.

For mechanically induced artefact signals, 1) and 3) will produce almost no signal, and 2) and 4) will produce a response that is proportional to the motion or force of the plunger.

Static charge is generated as the plunger slides through the plunger guide and is then transferred to the bone as the plunger makes contact with the bone. Since both an insulated (micarta) and uninsulated (metal) plunger were used, other approaches should be tried for applying the load. To avoid static charges produced near the bone, loading should be applied from a source external to the testing apparatus. A hydraulic shaker with a plunger that maintains constant contact with the bone is one possibility.

Any future testing should involve the use of the plexiglas "bone" to test for potential noise sources, especially since noise sources may be indistinguishable from the desired signals to be measured.

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APPENDICES

APPENDIX A

AMPLIFIER THEORY AND DESIGN

Mechanically stressed bone was modeled by Simpson (1980) as a signal source with a complex impedance represented by a resistor in parallel with a capacitor. To simplify analysis, bone was represented as a Norton current source.

Figure 39 shows the schematic of the general, non-ideal, linear amplifier used for analysis. R_1 and C_1 are the input resistance and input capacitance, R_0 , the output resistance, V_1 , the differential input voltage, and $A_V(s)$, the open loop gain. For a single pole system

$$A_V = \frac{A_0}{1 + s/\omega_0} \quad , \quad (1)$$

where A_0 is the magnitude of the open loop gain, and ω_0 is the -3db frequency associated with the pole.

Voltage Amplifier

The model used for deriving the transfer function (Figure 40) showed R_g and C_g as the source resistance and the source capacitance, R_B the bias resistance, R_2 and R_3 the feedback resistances, and R_L the load resistance. R_y and C_y were combined into Z_g , and R_1 and C_1 were combined into Z_1 , to produce the following nodal equations:

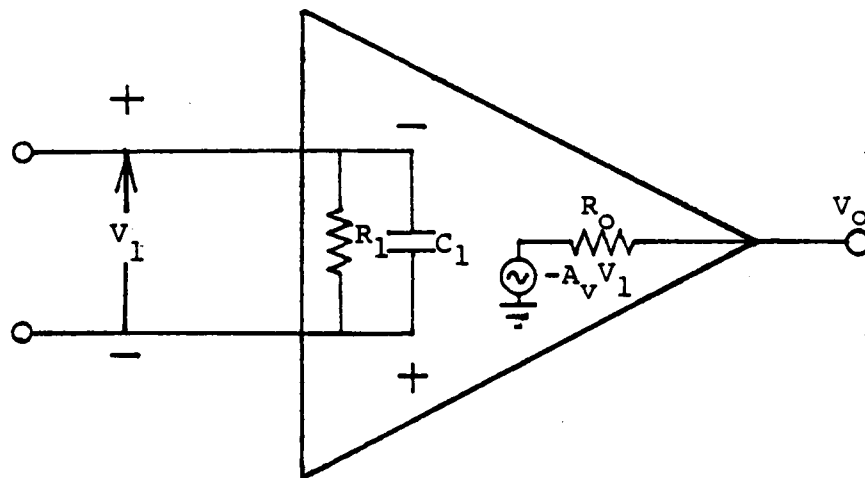


Figure 39. General Op Amp Model.

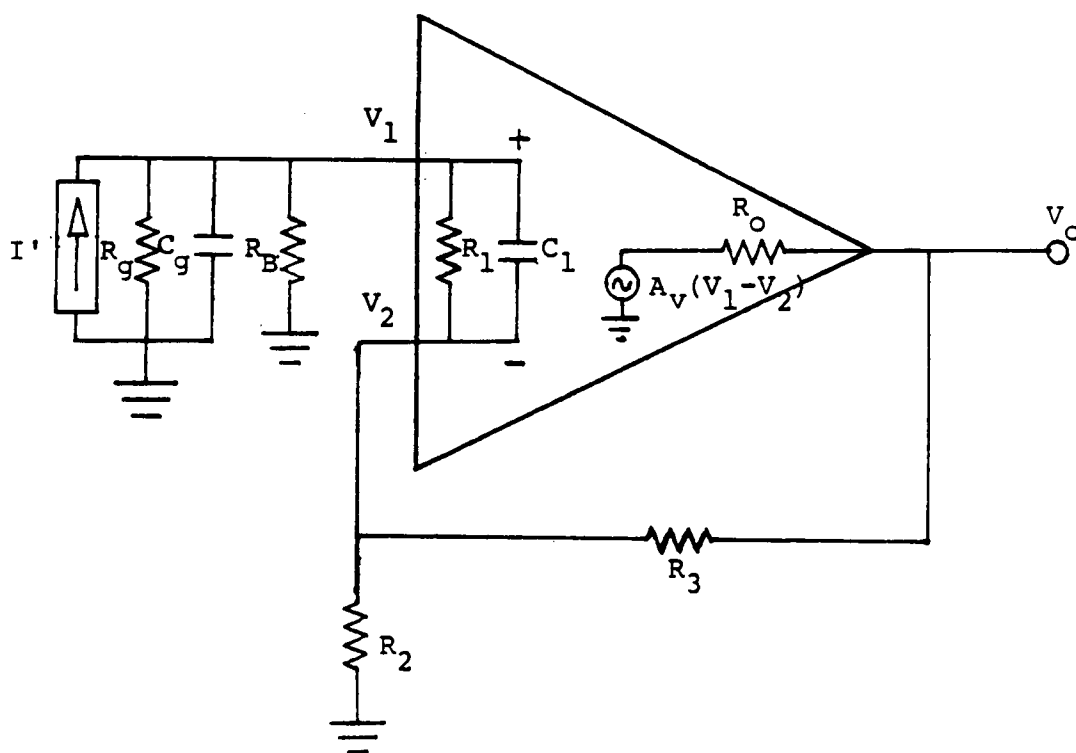


Figure 40. General Model of Voltage Amplifier.

$$\begin{aligned}
I_1 &= V_1 \left(\frac{1}{Z_g} + \frac{1}{R_b} + \frac{1}{Z_1} \right) + V_2 \left(-\frac{1}{Z_1} \right) \\
0 &= V_1 \left(-\frac{1}{Z_1} \right) + V_2 \left(\frac{1}{Z_1} + \frac{1}{R_2} + \frac{1}{R_3} \right) + V_0 \left(-\frac{1}{R_3} \right) \\
0 &= V_1 \left(-\frac{A_V}{R_0} \right) + V_2 \left(\frac{A_V}{R_0} - \frac{1}{R_3} \right) + V_0 \left(\frac{1}{R_0} + \frac{1}{R_L} + \frac{1}{R_3} \right) \quad (2)
\end{aligned}$$

This set of three simultaneous equations was solved for the three unknowns; V_1 , V_2 , and V_3 . The transfer function was reduced even further by substitutions based on known operating conditions. The final form of the transfer function became

$$\begin{aligned}
\frac{V_0}{I_1} &= \frac{\frac{R_2 + R_3}{R_2} \left(\frac{R_g R_B}{R_g + R_B} \right)}{s^3 \left[\frac{R_3 C_g C_1 R_g R_B}{A_0 \omega_0 (R_g + R_B)} \right] + s^2 \left[\frac{C_g R_3 R_B R_g}{A_0 \omega_0 (R_g + R_B)} \left(\frac{1}{R_2} + \frac{1}{R_3} + \frac{1}{R_1} + C_1 \omega_0 \right) \right]} \\
&\quad + s \left[\left(\frac{1}{R_2} + \frac{1}{R_3} + \frac{1}{R_1} \right) \frac{R_3}{A_0 \omega_0} + \frac{R_3 R_g R_B \left(\frac{1}{R_2} + \frac{1}{R_3} \right)}{R_1 A_1 \omega_0 (R_g + R_B)} + \frac{C_g R_g R_B}{R_g + R_B} \right] + 1 \quad (3)
\end{aligned}$$

Current Amplifier

An analysis was also done for the current amplifier, (Figure 41). The nodal equations were written by inspection from Figure 41.

$$\begin{aligned}
I_1 &= V_1 \left(\frac{1}{Z_g} + \frac{1}{R_f} + \frac{1}{Z_1} \right) + V_0 \left(-\frac{1}{R_f} \right) \\
0 &= V_1 \left(\frac{A_V}{R_0} - \frac{1}{R_f} \right) + V_0 \left(\frac{1}{R_0} + \frac{1}{R_f} + \frac{1}{R_1} \right) \quad (4)
\end{aligned}$$

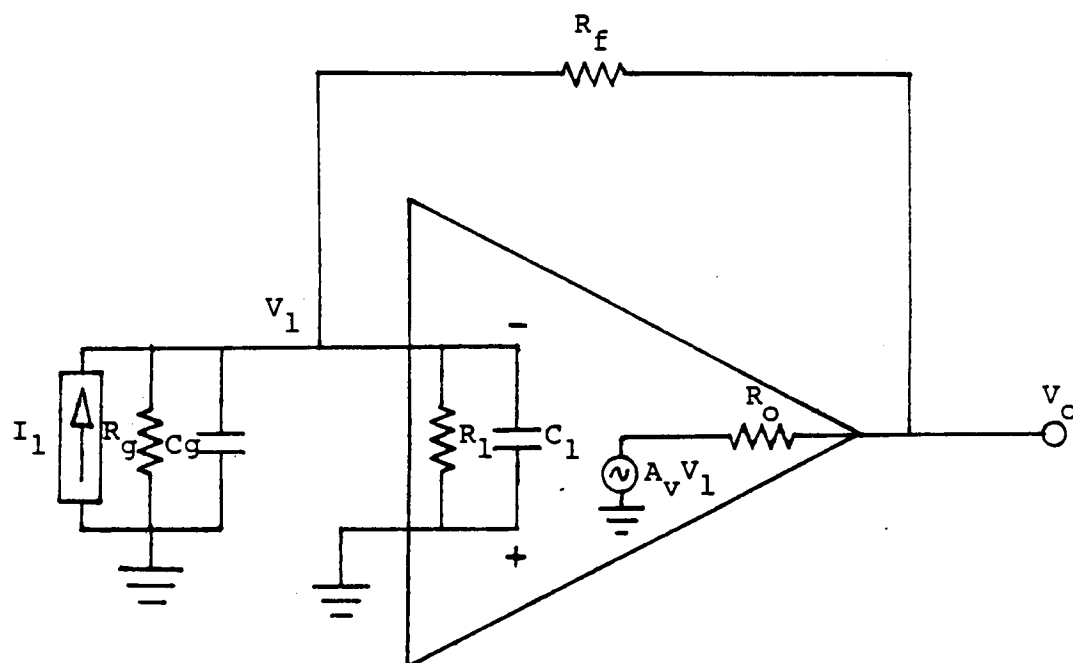


Figure 41. General Model of Current Amplifier.

The same assumptions used for the voltage amplifier were used for the current amplifier, so that the reduced form of the transfer function becomes

$$\frac{V_0}{I_1} = \frac{\frac{R_0}{C_g R_f} \left(s - \frac{A_0 \omega_0 R_f}{R_0} \right)}{s^2 + s \left(\omega_0 + \frac{1}{R_f C_g} + \frac{1}{R_1 C_g} \right) + \frac{A_0 \omega_0}{R_f C_g}} \quad (5)$$

Rise Time

Rise time (a measure of how fast an amplifier responds to an input signal), may be approximated from the equation

$$\tau_r = \frac{0.35}{f_{hi}} \quad , \quad (6)$$

where f_{hi} is the -3dB frequency point of the amplifier.

By converting the transfer function of the voltage amplifier, given by Equation (3), to the frequency domain, Equation (7) was obtained:

$$\frac{V_0}{I_1} = \frac{R_g \left(\frac{R_2 + R_3}{R_2} \right)}{1 - f^2 / \frac{A_0 \omega_0}{R_g} \left(\frac{R_2}{R_2 + R_3} \right)} \cdot \frac{1}{+ j \left[f / A_0 \omega_0 \left(\frac{R_2}{R_2 + R_3} \right) + f / \frac{1}{2\pi R_g C_g} \right]} \quad (7)$$

Substituting known values for resistance and capacitance, Equation (7) became

$$\frac{V_0}{I_1} = \frac{R_g (11)}{1 - \frac{f^2}{\frac{1}{R_g}(1.59 \times 10^{16})} + j \left[\frac{f}{32 \times 10^3} + \frac{f}{\frac{1}{R_g}(7.96 \times 10^{10})} \right]} \quad (8)$$

A Plot of the rise time verses R_g for the voltage amplifier is presented in Figure 42.

For the current amplifier, Equation (5) showed that the transfer function was independent of R_g . This assumption was valid for values of $R_g A_0 \gg R_f$, which was the case for bone (which typically has a range for R_g from 10^9 to 10^{12} ohms). Equation (5) was transformed to the frequency domain to get

$$\frac{V_0}{I_1} = \frac{\left(j f \frac{A_0 \omega_0}{R_0} - R_f \right)}{1 - f^2 \frac{A_0 \omega_0}{R_f C_g} + j \left(f \frac{A_0}{2\pi R_g C_g} + f \frac{1}{A_0 \omega_0} \right) + 1} \quad (9)$$

Substituting values of R_f equal to 10^9 ohms and C_g equal to 2 picofarads, the rise time of the current amplifier became 0.0171 milliseconds (much faster than that of the voltage amplifier).

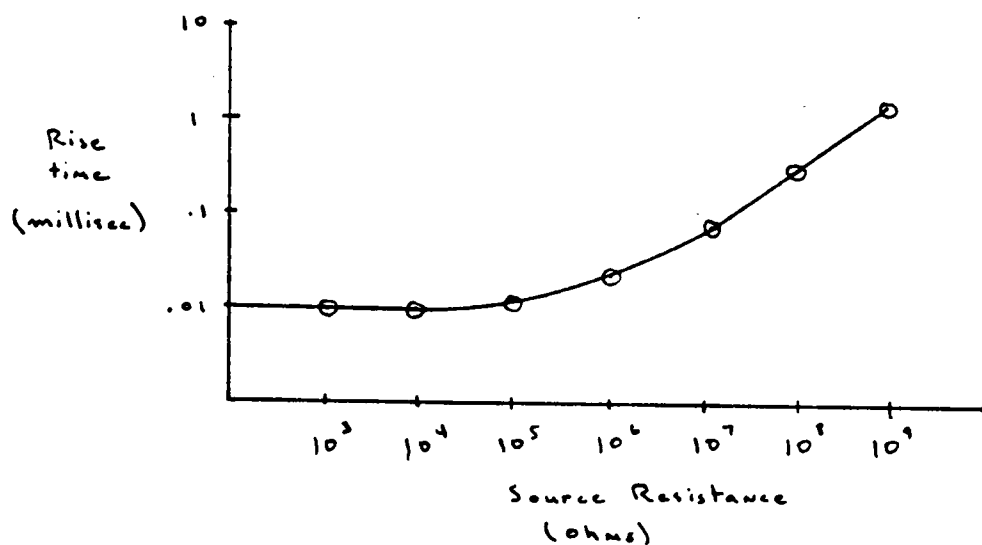


Figure 42. Rise Time Versus Source Resistance for a Voltage Amplifier with 2 Picofarads of Source Capacitance.

Source: M. L. Simpson, "Evaluation and Development of the Instrumentation Used in the Measurement of Bone Piezoelectricity," M.S. Thesis, The University of Tennessee, Knoxville, 1980, p. 59.

Noise Analysis

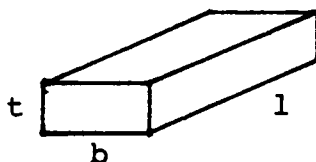
Using typical values of the average magnitude for current (10 picoamps) and voltage (15 millivolts), and using the results from the noise analysis, the signal to noise ratio was calculated for each system assuming R_g equal to 10^9 ohms.

For the voltage amplifier, the signal to noise ratio was calculated to be approximately 20 dB. From the current amplifier, the signal to noise ratio was calculated to be approximately 29 dB which demonstrated that the current amplifier had the best signal to noise ratio.

APPENDIX B

NATURAL FREQUENCY CALCULATIONS OF BONE
SUBJECT TO CANTILEVER BENDING

The natural frequency of a bone chip was calculated by using the formulation presented by Thomson (1981) for a beam that is clamped on one end and free on the other end. The governing equations are given below:



$$\omega_n = (B_n l)^2 EI / ml^4, \quad (10)$$

where

m = mass/unit length

$$(B_n l)^2 = 3.5160$$

$$I = (1/12)bt^3$$

given:

$$t = 2\text{mm}, \quad b = 3\text{mm}, \quad l = 30\text{mm}$$

$$\rho_{\text{bone}} = 1.9 \times 10^3 \text{ Kg/m}^3$$

$$E_{\text{bone}} = 13.79 \text{ GPa}$$

Solving for this equation, the undamped natural frequency was found to be 2.2 kHz.

APPENDIX C

SIGNAL PROCESSING OF THE AVERAGED VOLTAGE
AND CURRENT SIGNALS

In this Appendix, the signal processing theory described in Chapter V is presented. The signal processing was performed on a Hewlett-Packard 5420B Digital Signal Analyzer using data stored on tape. It was theorized that the observed voltage and current signals (in the time domain) could consist of two functions: 1) a noise function ($\phi(t)$) due mostly to motion in the coax, and 2) a piezoelectric signal function ($\psi(t)$) from the motion of the bone samples. The resultant measured signal would then be expressed as the combination ($\phi(t)+\psi(t)$). By reversing the electrode leads, the resultant signal would become the difference ($\phi(t)-\psi(t)$). If this is true, the measured signals could be processed to produce the separate component signal functions.

$$\frac{(\phi + \psi) - (\phi - \psi)}{2} = \quad \text{(total bone response)} \quad (11)$$

$$\frac{(\phi + \psi) + (\phi - \psi)}{2} = \quad \text{(coax cable response)} \quad (12)$$

$$(\phi + \psi) - \phi = + \psi \quad \text{(bone response from } \phi + \psi \text{)} \quad (13)$$

$$(\phi - \psi) - \phi = - \psi \quad \text{(bone response from } \phi - \psi \text{)} \quad (14)$$

The results from the signal processing are discussed in Chapter V.

VITA

Gary Lynn Chamberlain was born April 10, 1956 in Maryville, Tennessee. He attended Union Elementary School and Graduated from Everett High School in Maryville where he received his high school diploma in 1974. In September of 1974 he entered The University of Tennessee at Knoxville where he began working on a double degree in Agriculture and Engineering. He was awarded a Bachelor of Science degree in Agriculture with a major in Animal Science in June 1980 and then a Bachelor of Science degree in Engineering Science with an area of concentration in Biomedical Engineering in December 1981. In September of 1981 he was admitted to a Master's program in Engineering Science and is expected to receive the Master of Science Degree in March 1986.

He was a member of the Block and Bridle Club and the Biomedical Engineering Society. He has been a member of the Graduate Student Council and participated in intramural athletics. He has been active in church and community activities including 4-H. His hobbies include horseback riding, hiking, photography, and electronics.