A Surface Probe for *In Situ* Detection of Cartilage Degradation via Electromechanical Spectroscopy

by

David L. Bombard

Submitted to the Department of Mechanical Engineering in partial fulfillment of the requirements for the degree of

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Abstract

A small, cylindrical (0.4 inch diameter), electromechanical surface probe to measure electrokinetic properties of cartilage in situ was designed, constructed and tested. Changes in these electromechanical properties are correlated with early tissue degradation (due to joint disorders like osteoarthritis). The probe was calibrated and subsequently used to measure electrokinetic properties of explanted cartilage discs and an intact articular cartilage-bone joint surface. The results validated this device as an instrument for measuring electrokinetic properties of cartilage in situ. The measurements made agreed well with trends found in previous experiments and predicted by theory. With some modifications and further testing, this probe design could be utilized to make in vivo measurements during arthroscopic surgery.

Thesis Supervisor: Alan J. Grodzinsky
Title: Professor of Electrical, Mechanical, and Bioengineering
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Chapter 1

Introduction

Osteoarthritis is a slowly progressing disease of synovial joints characterized by the degradation and loss of articular cartilage. Diagnosis of osteoarthritis is currently based on radiographic evidence and clinical evaluation[26]. Both of these methods convey qualitative information and can generally detect only the pronounced loss of cartilage indicative of late stage, currently irreversible osteoarthritis. The need for early detection of cartilage degradation will increase as methods of treatment improve, since the success of most treatments relies on prompt diagnosis. One of the initial changes to osteoarthritic cartilage is biochemical and not easily detectable with standard radiological or imaging methods. A nondestructive surface probe technology has been developed that can detect mechanical and biochemical changes by measuring electromechanical properties of cartilage in vitro [29, 10, 2]. This research incorporates electromechanical spectroscopic measurement techniques into a device to be used on intact joints in situ.

The first chapter of this thesis describes relevant articular cartilage composition and structure, the tissue’s electromechanical transduction properties, the degradative changes to which the probe is most sensitive, and the previous research motivating this intact-surface probe. Chapter 2 documents the design and construction of this probe. Chapter 3 describes experimental methods for probe calibration and measurement of electromechanical properties. Experimental results are contained in Chapter 4 along
1.1 Articular Cartilage

Articular cartilage covers the ends of load-bearing articulating bones and acts as a low-friction, shock-absorbing, wear-resistant surface. Adult cartilage has neither blood vessels nor nerve endings and appears as a homogeneous, whitish tissue under gross and light microscope observation. This appearance hides articular cartilage’s highly ordered complex structure which provides bearing properties unmatched by any man-made substitute. Cells (chondrocytes), occupying a small portion of the total volume of adult human articular cartilage, maintain an extracellular matrix which is largely responsible for this tissue’s extraordinary properties. This matrix also gives rise to the electromechanical behavior that is the basis for measurements made with our surface probe.

1.1.1 The Extracellular Matrix

The extracellular matrix is composed of a highly hydrated framework of structural macromolecules (Figure 1-1). Water comprises from 60-80% of the tissues total weight [21]. The main structural macromolecules include (as % dry weight):

**Collagen** (50%): 90 to 95% in the form of type II banded fibrils.

Type IX and XI are also present.

**Proteoglycan** (30-35%): 90% in the form of aggrecan.

**Noncollagenous proteins and glycoproteins** (15-20%).

The collagen fibrils resist tensile and shear forces, while the proteoglycans account for much of the compressive strength of cartilage. The other constituents contribute to the formation and stability of the framework and are also believed to be involved in cell-matrix interactions.
Figure 1-1: Schematic illustration of the structure of articular cartilage, showing cells (chondrocytes), collagen fibrils, and proteoglycan (PG) aggregates with their constituent glycosaminoglycan (GAG) chains.
The large, aggregating proteoglycans of cartilage (aggrecan) are composed of a protein with one or more attached glycosaminoglycan (GAG) chains. The GAG chains are linear polymers of chondroitin sulfate and keratan sulfate disaccharides. These chains are covalently bound to the protein core [15]. In normal cartilage, these PG molecules are non-covalently associated with hyaluronan (long chain hyaluronic acid) through a link protein molecule to form large aggregates (up to $200 \times 10^6$ Da)(Figure 1-2). Each aggrecan molecule might have 100 to 150 associated GAG chains [16, 15]. At physiologic pH, the high density of ionized fixed charge groups (COO\(^-\), SO\(_3^-\)) on these GAG chains leads to an electrostatic swelling pressure. The chains electrostatically repel each other and the molecule forms a hydrated “bottle-brush” structure. This swelling pressure, in conjunction with the collagen fibril network, helps cartilage to resist equilibrium compressive loads [21]. The high negative fixed charge density and resultant excess of positive, free counter-ions in the fluid impart electromechanical properties to the tissue.
1.1.2 Cartilage Electromechanics

The structure of cartilage gives rise to electromechanical transduction effects. Investigators have found that when cartilage is compressed, an electrical potential can be measured [1, 19, 14]. The mechanism behind this phenomena was found to be an electrokinetic effect, a “streaming potential”: A mechanical deformation causes a flow of fluid (and positive ions) relative to the fixed negative charge groups on the matrix. This separation of oppositely charged groups gives rise to a voltage gradient in the direction of fluid flow [13, 8]. Conversely, when an electric current is applied to cartilage, a measurable stress is developed [8]. In this case, the current exerts a force (electrophoretic) on the fixed negatively charged matrix towards the positive electrode, while the positive counter-ions in the fluid are drawn towards the negative electrode and induce a fluid flow (electroosmotic) in that direction (Figure 1-3). The net result is a measurable “current-generated stress”.

Figure 1-3: Mechanism of current-generated stress (electrical-to-mechanical transduction) in cartilage in a uniaxial geometry. Current injection produces motion of the solid matrix and opposite fluid flow, resulting in a measurable mechanical stress.
This phenomena has been successfully modeled by combining linear electrokinetic coupling laws [6] with the biphasic theory [23] that relates stress, strain, and fluid flow in cartilage. This mathematical model predicts the magnitude and phase of streaming potential and current-generated stress [9]. Studies confirmed that at physiologic pH, this transduction phenomena is due to the charged proteoglycan molecules [18, 8]. A loss of these molecules such as in osteoarthritis (described in the next section) significantly changes the properties of the tissue. A measurement of these tissue properties could provide a quantitative measure of the mechanical and biochemical condition of the tissue.

1.2 Osteoarthritis

Osteoarthritis is a common joint disorder that accounts for more disability than almost all known ailments [17, 7]. The disease is widespread in all populations, particularly afflicting the elderly [7]. The disease is characterized by articular cartilage degradation, particularly from weight-bearing synovial joints, and a resultant loss of joint function [5]. Gross pathological changes include surface fibrillation, fissures, erosion, and exposure of underlying (subchondral) bone, but appear at a relatively late, currently irreversible stage of the disease. One of the earliest changes to osteoarthritic cartilage is a nonuniform decrease in proteoglycan content [4, 20, 22]. The magnitude of the proteoglycan loss has been shown to correlate with the severity of the disease. Concurrent with the loss of proteoglycan molecules is a decrease in the tissue’s fixed charge density. This decrease results in significant changes to physical properties that affect the tissue’s ability to function, and leads to further degradation. A measurement of current-generated stress would be sensitive to these early changes. The magnitude of current-generated stress is proportional to the fixed charge density of the tissue, while also depending on other physical parameters such as the hydraulic permeability and equilibrium modulus.
Figure 1-4: Device for uniaxial measurement of streaming potential and current-generated stress in cartilage samples. This apparatus required that a cartilage disc be excised and placed between two electrodes, precluding its use on intact joints.

1.3 Previous Research: Surface Spectroscopy

The earliest measurements of current-generated stress utilized a uniaxial configuration which required access to opposite sides of a cartilage sample (Figure 1-4) [8, 9]. Theoretical analysis [28] suggested that current applied to a single surface of an electromechanically coupled material like cartilage would produce a measurable stress at the same surface. The normal stress at the surface is predicted to have the same spatial periodicity as the applied current, but be temporally out of phase (for details, see [28]). Such measurements have been successfully made and compared with theory [29, 3]. The device for making these surface measurements was a chamber-mounted structure that required the cartilage to be in the form of plane parallel discs or plugs (Figure 1-5). A multi-electrode probe with variable spatial wavelength has also been constructed and tested for use in this chamber-mounted probe system. The success of measurements made with this device motivated the design and construction
Figure 1-5: Chamber-mounted probe system for making current-generated surface stress measurements on excised cartilage discs.
of a similar device that could make measurements on an *intact* joint surface. This intact surface probe would utilize the same sensing technology as the chamber-mounted model, but be packaged into a smaller, more compact unit. A prototype probe has been designed and constructed that can measure current-generated stress on intact articular cartilage joint surfaces. One potential application is to use this device at the time of arthroscopy. Arthroscopic procedures are commonly performed during routine diagnosis of joint disorders. An intact-surface probe could potentially gather *quantitative* information and detect early degradation during arthroscopic procedures, before visible manifestations of osteoarthritis appear.
Chapter 2

Design and Construction

This section describes the design and construction of an intact-surface probe system for measuring current-generated stress on intact articular cartilage-bone surfaces. The design is a two part, modular structure consisting of a laminated electrode/transducer structure (similar to earlier probes) mounted in a cylindrical body. The body consists of a four part design incorporating an inner core which makes electrical connections to the electrode/transducer structure, covered with a plastic sheath and inserted into a threaded stainless steel outer tube and screw (see Figure 2-2). The sheath and threaded tube with screw combine to allow a portion of the electrode/transducer structure to be in contact with a cartilage surface while protecting, shielding, and sealing the various connections from an aqueous working environment.

2.1 Probe Design

2.1.1 Electrode/Transducer Structure Design

The Electrode/Transducer Structure (ETS) is the active element of the probe system which applies current to the cartilage surface and measures the resultant stress. It is a flexible, 180 μm thick, 3 layer laminated structure. The layers consist of a pair of silver-silver chloride (Ag/AgCl) electrodes for applying current and a corresponding pair of piezoelectric stress sensors, separated by a sheet of metal coated mylar
ground plane (metallization on one side) for electromagnetic shielding and insulating. Figure 2-1 displays these various elements of the ETS in relation to each other.

A pair of silver excitation electrodes etched from a sheet of silver foil and deposited with a layer of silver chloride is used for applying sinusoidally varying current densities to the cartilage surface (Figure 2-1c). These silver-silver chloride electrodes decrease the low frequency impedance between the electrode and cartilage while stabilizing the electrode potential [12].

Stress sensors are fabricated from a single sheet of polyvinylidene fluoride (PVDF) with thin sputtered nickel-copper metallization on both sides (Figure 2-1a). This piezoelectric material acts as a transducer, converting mechanical stress to a measurable voltage signal. Metallization on one surface is formed into two electrodes masked in register with the two silver electrodes. The other side of the PVDF is connected to the ground plane (mylar metallization).

A metallized mylar layer separates the silver electrodes and stress sensors while also serving as a ground plane. The metallization is a ground plane that helps shield the sensitive stress sensors from electromagnetic interference (principally electric fields emanating from the excitation electrodes) when the ETS is placed in the probe body.

A piezoelectric material, such as the PVDF film used as the stress sensor in this system, is a piezoelectric transducer which develops an electrical surface charge proportional to an applied mechanical stress. The PVDF is manufactured by AMP Incorporated (‘Piezo Film Sensor’, Norristown, PA), and was chosen as the stress sensor in this application for several reasons. Its high sensitivity to mechanical stress allows the film to behave like a compact strain gauge with no external power source, and generates signals greater than those from typical strain gauges after amplification [25]. The flexible sheets are relatively inexpensive and can be cut into arbitrary patterns. The sheets come in various thicknesses (9-110 μm) and metallizations. A 52 μm thick sheet with sputtered NiCu metallization was chosen for this design. This particular sputtered metallization is susceptible to microcracking when severely flexed, and the lead attachment options were limited. Such constraints had to be considered during probe body design, and might be related to certain problems that were encountered.
Figure 2-1: Top (a), bottom (c), and exploded cross-sectional (b) views of the Electrode/Transducer Structure (ETS) with silver electrodes (c) for applying sinusoidally varying current densities to cartilage surface, and stress sensor (a) which measures the surface stress. Metallized mylar (b) provides a ground plane for shielding, and a non-conducting mounting surface for the silver electrodes.
Stress Sensor Characterization

Characterizing the sensor as a stress gauge, a force applied normal to the surface of the film develops an electric surface charge on the metallization proportional to the mechanical stress. The film also responds to forces along its length and width, but we neglect these forces in this analysis since the measured current-generated stress is principally in a direction normal to the film surface. The charge $Q$ [coulombs] developed by a stress $\sigma$ [N/m$^2$] over an area $A$ [m$^2$] can be described by

$$ Q = d_t \sigma A, \quad (2.1) $$

where $d_t$ is an empirically determined piezoelectric strain constant. The open circuit voltage between the metallization on either side of the sheet is this charge divided by the capacitance, $\epsilon A/\delta$, where $\epsilon$ is the dielectric constant of the film and $\delta$ is the film thickness. If part of the total metallized area is not being loaded, it adds to the capacitance without generating any charge, thus decreasing the measured voltage.

Representing the total area by $A'$ and the active area being loaded by $A$, the equation for the open circuit voltage signal becomes:

$$ V = \frac{Q}{C_{total}} = \frac{d_t \sigma A}{\epsilon A'/\delta} = \frac{d_t \sigma \delta A}{\epsilon A'} \quad (2.2) $$

As a stress sensor, it is desirable to maximize the measured voltage signal, $V$, for a given stress, $\sigma$. In this design, $A/A' = 0.63$ due to an inactive area of the electrode metallization used for making connections to the external circuitry. This is an improvement over the previous area ratio of 0.20 associated with the chamber mounted ETS [3]. Equation 2.2 also indicates that $V$ is directly proportional to the film thickness. However, a thinner film can better discriminate between closely spaced stresses. The electrodes in this design are spaced 1 mm apart, and a thickness of 52 $\mu$m was found to be satisfactory. As more electrodes are added, the electrode spacing will become smaller and a thinner film might be needed to gain sufficient spatial resolution. Reference [25] contains detailed information on the properties of this piezoelectric...
2.1.2 Probe Body Design

The ETS is placed in a body consisting of an inner core, plastic sheath, and stainless steel tube with screw. The core serves as a smooth, flat mount for the active area of the ETS and makes electrical connection to the excitation and stress sensor electrodes. The sheath is placed over the core and ETS. This subassembly is clamped into the stainless steel tube with the screw (Figure 2-2). These components combine to allow the active area of the ETS to be in firm contact with an intact cartilage surface while protecting and sealing the electrical connections from the surrounding electrolyte present in a synovial joint or test chamber.

Electrical connections to the ETS are made through pressure contacts on the core. The connection to each stress sensor electrode is made through a brass contact on the beveled edge of the core (Figure 2-3). Connections to the silver excitation electrodes are made through a spring clip mounted to the core (Figure 2-4).

The ETS is sealed around its circumference by the force of a screw which drives the core and sheath (with the ETS sandwiched between them) into the chamfered end of a stainless steel tube (Figure 2-2). This clamping action also assures that the brass contacts are firmly seated against the sensor electrode pads. The sleeve helps to maintain a water tight seal and also insulates the excitation electrodes from the tube and each other.

Electromagnetic shielding of the piezoelectric sensor is accomplished by a machined aluminum shell surrounding the core and connected to the coaxial cable shielding (Figure 2-3). The ground plane of the ETS is electrically connected to this aluminum sleeve through the back of the excitation electrode leads.
Figure 2-2: Cross-section assembly drawing of the probe system. Cross-section shows electrical connections to stress sensor. Connections to the silver electrodes are made in a plane rotated $90^\circ$ to this cross-section.
Figure 2-3: Top and cross-section assembly view of probe core and ETS showing stress sensor connections.
Figure 2-4: Top, Cross-sectional, bottom, and auxiliary side view of probe core (cross section plane rotated 90° with respect to the previous figure) showing the excitation electrode connections.
2.2 Construction

2.2.1 ETS Fabrication

It is necessary to replace the ETS periodically. For example, the excitation electrodes might delaminate and the shielding can develop gaps as the ETS is repeatedly calibrated and used in experiments. In the final device concept, the ETS would be considered a “disposable” item to be replaced after each diagnostic procedure. This section describes the process of fabricating this laminated structure.

ETS Assembly

Sheets of material are cut to the appropriate size and then laminated together. The excitation electrodes are made from a 25.4 μm thick silver foil. The shielding layer is made from 25.4 μm thick metallized (on one surface) Mylar polyester film (MADICO, Woburn MA). The stress sensor is 52 μm thick PVDF piezo film.

The following pieces are cut using a sharp scalpel: 35 x 15mm piece of silver foil, 45 x 20mm piece of metallized Mylar, and a 20 x 15mm piece of piezo film. The piezo film is cut so that its long axis is parallel to the longitudinal axis of the film (as determined by striations in the film metallization). The sheets are dipped in a mild detergent and rinsed with deionized water to remove dust and grease. Handling is done with disposable latex gloves to keep all materials clean.

These materials are next bonded together to form a laminated structure. To prepare the silver for photofabrication, both sides are gently abraded with fine (600 grit wet/dry) sandpaper and then dipped in a 15% v/v nitric acid solution. The silver is bonded to the \textit{non-metallized} side of the Mylar (centered, with their long axes aligned) using urethane epoxy (Tycel 7000/7200, Lord Corp., Erie, PA) thinned with methyl ethyl ketone. The piezo film is bonded to the \textit{metallized} side of the Mylar (centered, striations perpendicular to the long axis of the Mylar) with silver conducting epoxy (TRA-DUCT 2902, TRA-CON Inc., Medford MA). This ETS is pressed together for a few minutes to assure good bonding and is allowed to cure overnight.


**Photofabrication**

Both the silver excitation electrodes and the piezoelectric stress sensor electrodes are fabricated with a standard photofabrication technique. Basically, the surface is coated with a light-sensitive organic polymer called *photoresist* that becomes resistant to etching chemicals when cross-linked by ultraviolet (UV) light. An arbitrary pattern of metallization can be created by shining the UV light through a negative of the desired pattern, thus leaving a protective stencil on the surface that will resist the etching chemicals.

The ETS is dehydration-baked at 80°C for 10 minutes in a convection oven to remove residual moisture, and both sides are coated with a photoresist compound (KPR Photoresist, KTI Chemicals, Sunnyvale, CA), dried for 30 minutes in a darkroom, and then baked between paper and glass plates (to keep them flat) at 80°C for another 10 minutes. Electrode patterns created with simple drawing software (XFIG 2.1) were converted to negative images, or masks, on two photographic transparencies (Fotobeam, Waltham, MA). A dry, coated ETS is placed between the two masks, aligned so that the positions of the excitation and sensor electrodes coincide, and glass plates, then exposed to ultraviolet light for 15 minutes. Following exposure, the ETS is bathed in a xylene-based developer solution (KPR Developer, KTI Chemicals, Sunnyvale CA) for 30 seconds, transferred to another bath of developer for 30 seconds, and then rinsed under warm tap water and blotted dry. The developer removes the photoresist that has not been cross-linked by UV light, leaving hardened resist behind in the shape of the electrodes.

Etching occurs while the ETS is mounted in a specially designed two-part poly-(methyl methacrylate) (PMMA) holder incorporating a rubber O-ring gasket to contain the etchant on a defined part of the surface. The etchant used is a 55% w/v solution of ferric nitrate heated to 45°C. The ETS is mounted in this holder, silver side exposed, and fresh etchant is added to the holder every 2 minutes (removing old etchant with a pipet). When etching is complete, the ETS is removed and washed off with deionized water. The very thin metallization on the piezo film is etched by
carefully placing a few drops of etchant on the surface, waiting only 5-10 seconds, and quickly rinsing with deionized water. The photoresist is finally removed from both sides with a cotton swab dipped in xylenes (Mallinckrodt Inc.), and the ETS is rinsed with deionized water.

2.2.2 Probe Body Construction

The body consists of four separate parts: a core for making electrical contacts, a sealing/non-conducting sheath, an outer threaded stainless steel tube, and a screw (Figure 2-2). The stainless steel tube and screw were fabricated by J & H Precision (Marlboro, MA). The core components were fabricated by the Center for Space Research Machine Shop (MIT). Other machining was performed by the author at the RLE machine shop (MIT).

A solid cylinder of polyamide-imid (PAI) was fitted with two cylindrical brass contacts and an aluminum shell (Figure 2-5). The shell and PAI have matching holes to accommodate two coaxial cables (low noise, 0.09 in diameter cables). The central conductors in each cable was bonded to one of the brass contacts with silver conducting epoxy (TRA-DUCT 2902). The coaxial shielding was bonded to the aluminum shell with the same conducting epoxy. The stress sensor shielding is complete when an ETS is mounted in the probe and the ground plane of metallized mylar is over the top of the core.

This upper section of the core was bonded to a slotted lower section which makes connections to the silver excitation electrodes. The bottom section is made of PAI and fitted with two conducting spring clips fastened by two small brass screws (0.06 in diameter, 80 threads/in) (Figure 2-6). When the two sections are bonded together, the spring clips extend from the lower section and press against the aluminum shell of the upper section. A thin wire (30 gauge, single strand) is wrapped around each fastening screw and exits to the slot in the center of the lower section. This slot also accommodates the coaxial cables and allows both wires to exit through a corresponding slot in the steel tube.

The three remaining parts of the probe system are a plastic sheath and threaded
Figure 2-5: Upper core cross-section showing brass sensor contacts, aluminum shell for shielding, and coaxial cables.
Figure 2-6: Cross-section, side, and bottom views of the core showing the lower section with spring clips bonded to the upper section. Silver electrode leads are held between the spring clips and aluminum shell.
stainless steel tube with screw (Figure 2-7). The tube and screw are both machined from 303 series stainless steel. The sheath material is an acetal resin (DELRIN, Dupont, Wilmington DE). The core, sheath, and tube each have 45° chamfered ends. The screw has a flat end for pressing against the core and a spherically rounded end which can articulate in a spherical holder.

2.3 Probe Assembly

The ETS is formed into its final three dimensional shape with a die. First, it is necessary to make relief cuts into the ETS so that the die stamping operation doesn’t introduce folds or wrinkles. The final active area of the ETS should ideally be a flat, wrinkle-free surface against the top surface of the core. A small fold or wrinkle introduces large local stress concentrations that are sensed by the piezo film and distort the measured signal significantly. The ETS is pressed with the die held in a lathe to maintain alignment and apply an even pressure.

The formed ETS is then mounted onto the core (Figures 2-2, 2-3 and 2-4). The silver leads are slipped under the spring clips, and the plastic sleeve is placed over this subassembly. This is slid into the stainless steel tube, allowing the wires to exit through the slot. The screw end is inserted behind and tightened until there is sufficient clamping pressure to hold and seal the ETS in place. To achieve a water tight seal, it is sometimes necessary to apply a thin bead of RTV silicone adhesive sealant to the inside edge of the plastic sleeve before sliding the plug in. The active surface of the ETS extends slightly beyond the tube end and makes unobstructed contact with the cartilage surface.

2.3.1 Electrode Chloridation

The final step is to deposit a layer of silver chloride onto the silver excitation electrodes. The active surface of the fully assembled probe is suspended in a bath of unbuffered 0.1M NaCl, titrated to pH 4.0 with 1 N HCl. The positive terminal of a variable DC power supply (Hewlet Packard) is connected to one of the silver elec-
Figure 2-7: Cross-section of tube, screw and sheath components.
trode wires, in series with an ammeter and a 47 kΩ resistor. The negative terminal is connected to a platinum strip electrode also suspended in the bath. A current of 120 μA is run for 10 minutes, corresponding to a total chloride deposit of 1000 (mA-seconds)/cm², which is within the acceptable range for bioelectric applications [12]. This protocol is repeated for the other electrode. The final assembled probe is pictured in Figure 2-8.
Figure 2-8: Fully assembled probe system. (Note: The excitation electrodes pictured here are from an earlier pattern and have since been changed to the half-moon pattern displayed in earlier figures.)
Chapter 3

Experimental Methods

3.1 Peripheral Electronics

The outputs from the stress sensor and the inputs to the silver excitation electrodes are attached to various electronic devices that process or generate signals.

The coaxial cable outputs from each stress sensor were connected to separate channels of a high input impedance electrometer with an adjustable ramp signal offset to compensate for electrode drift. The two outputs from the electrometer (channels labeled "left" and "right") were passed through a 4-pole low-pass filter (Model 1022F, Rockland Systems, West Nyack NY) with a cutoff frequency of 15.7 Hz to eliminate 60 Hz noise. During calibration, the signal from each channel was recorded on a Gould Brush 2200 chart recorder and digitally sampled by microcomputer. For experiments, the electrometer outputs were connected to a high input impedance differential amplifier (model 11-4113-01, Gould Electronics, Cleveland OH) and the differential signal ("left" minus "right") was recorded by the chart recorder and sampled by the computer. This differential amplifier increased the signal amplitude while suppressing common-mode noise (enhancing the signal-to-noise ratio) because the outputs of the two sensors are expected to be 180° out of phase.

The excitation electrode inputs were connected to a bipolar operational amplifier (Kepco, Flushing NY), configured as a current source and driven by a programmable
frequency synthesizer (Rockland Systems, West Nyack NY). The current was monitored by a digital ammeter (Keithley Instruments, Cleveland OH) in series with the load and recorded by the computer (digitally) and chart recorder (hard copy). The current was determined by scaling the output voltage by an empirically determined constant. Current densities of 1.0 and 0.5 mA/cm² were applied to the cartilage surface at frequencies of 0.025, 0.05, 0.1, 0.25, 0.5, and 1.0 Hz. These current densities correspond to total current amplitudes of 73.2 µA and 36.6 µA (electrode area = 0.0732 cm²).

A microcomputer was used to control the frequency synthesizer and digitally sample various signals during calibration and experiment. The frequency synthesizer was used during calibration as the dynamic input to a servo-controlled material testing machine (DynaStat, IMASS, Hingham MA) to generate known sinusoidal loads, and to drive the current source during experiments. An analog-to-digital converter onboard the computer was attached to outputs from the sensor, excitation circuits, and DynaStat. On-line Fourier analysis helped determine the amplitude, and phase relative to one of the inputs, of the fundamental component and four higher harmonics of each input signal. The higher harmonics were used to compute a total harmonic distortion,

\[
THD = \sqrt{\sum_{n=2}^{5} x_n^2},
\]

where \(x_n\) is the fractional content of the \(n\)th harmonic. The response was deemed to be essentially linear if the THD was less than 10%.

### 3.2 Experimental Setup

To simulate a joint surface in a controlled environment, a chamber to hold cartilage samples was mounted to the DynaStat actuator collet, while the probe was mounted to the load cell collet (Figure 3-1). The chamber is a machined PMMA container that fits into the actuator collet of the DynaStat and can hold specimens while bathing them in a recirculating bath. Inlet and outlet ports were attached to silicone tubing that
Figure 3-1: Probe and chamber experimental setup schematic. PMMA chamber and probe are mounted in the DynaStat mechanical testing machine during calibration. This setup is also used during experiments to maintain a static offset.
runs through a reciprocating pump. The temperature of the bath was controlled by a heater connected to a proportional controller (Digi-Sense, Cole-Parmer Instrument Co., Chicago II). The bath solution was run through aluminum tubing submerged in water heated by the controller. Temperature was sensed and looped back to the controller by a thermistor placed in the bath. The same chamber was also used during calibration, with rubber replacing the cartilage samples.

A ball and socket type joint was used to connect the probe to the DynaStat load cell (Figure 3-1). A holder with a spherical seat was fashioned to accept the rounded end of the probe. The holder fits into the load cell collet and allows the probe tip to be positioned anywhere within an angle of 10⁰ revolved about its axis.

3.3 Transducer Calibration

3.3.1 The Transfer Function

To use the piezoelectric film as a stress sensor, it is necessary to determine the relationship between a force applied to the sensor and the voltage measured. It was assumed that the response of the piezoelectric film system is linear, so that an imposed sinusoidal stress, \( \sigma \) with frequency \( \omega \), produces a sinusoidal voltage, \( V \), at the same frequency. The signals in this system can be represented with complex notation as:

\[
\sigma = \Re \{ \hat{\sigma} e^{j\omega t} \} \quad (3.2)
\]

and

\[
V = \Re \{ \hat{V} e^{j\omega t} \} . \quad (3.3)
\]

To convert a signal from the sensor into a stress, a transfer function can be found,

\[
H_{V/\sigma} = \frac{\hat{V}}{\hat{\sigma}} = r(\omega)e^{j\theta(\omega)} \quad (3.4)
\]

where \( r(\omega) \) is equal to the ratio of the real amplitudes, while \( \theta(\omega) \) is the phase difference between \( V \) and \( \sigma \). The transfer function \( H_{V/\sigma} \) can be determined by applying
loads of known amplitude and frequency and measuring the sensor output. This calibration transfer function can then be used to determine the stress measured by the sensor during an experiment.

3.3.2 Calibration Protocol

To determine the transfer function at various frequencies, the probe and chamber were mounted into the DynaStat jaws as described above. The frequency generator was connected to the DynaStat dynamic input and the DynaStat load output connected to the microcomputer. With the DynaStat applying a 30 kPa dynamic load at 1.0 Hz, the probe position and static load offset was adjusted until the sensor outputs were approximately of equal amplitude and in phase. Superimposed on the static offset, dynamic loads corresponding to stress amplitudes of 10, 5, 2.5, and 1.25 kPa were applied at frequencies of 0.025, 0.05, 0.1, 0.25, 0.5, and 1.0 Hz. The static stress typically ranged from 50 to 75 kPa. Each stress sensor (left and right) is recorded and sampled separately. The amplitude and phase of each channel was taken to be the amplitude and phase of the first harmonic of the Fourier decomposition. The transfer function for each channel was then computed. For each frequency, the magnitude of the transfer function is the slope of the line relating output voltage to applied stress (slopes calculated using a least-squares linear regression). The phase of the transfer function was assumed to be independent of stress amplitude and calculated by averaging the phase obtained for each amplitude. The magnitudes and phases of the left and right channels were averaged to obtain a single transfer function at each frequency. The probe was calibrated directly before each experiment, because the transfer function can vary over time and is different for each ETS.

3.4 Preparation of Cartilage Samples

Intact knees (including femur and tibia sections) of calves were obtained from a specialized tissue transporter (Research 87, Boston MA) within one day of slaughter. The femur was transected distal to the knee joint with a hacksaw and mounted in
a vise connected to a ball and socket joint. The tibia, surrounding soft tissue, and patella were removed exposing the surface of the femoropatellar groove. At this point two different procedures were used, one that extracts cartilage discs, and another for obtaining intact cartilage/bone sections.

### 3.4.1 Cartilage Discs

To extract cartilage plugs, the vise was then moved to a drill press with a 3/8 inch coring bit attached, and pivoted so that part of the cartilage surface was approximately perpendicular to the the drilling axis. Keeping the surface irrigated with 0.15 \( M \) NaCl to lubricate and cool the cartilage, the drill bit was lowered to cut a 3/8 inch diameter, 1-2 cm long plug. This procedure was repeated four times each on the medial and lateral facets, for a total of eight plugs. The plugs were stored at \(-20^\circ C\) in sealed containers.

On the day of experiment, plugs were thawed at room temperature in a bath of 0.15 M NaCl plus 2 mM [ethylene_dinitrilo]tetraacetic acid (EDTA) protease inhibitor, then mounted in a sledge microtome and adjusted until the articular surface was approximately parallel to the blade. The upper surface was removed until level (upper 80-160 \( \mu m \)), and then as many 1 mm slices were cut from the plug before hitting subchondral bone. These slices were stored in the NaCl-EDTA bath during experiments (less than 2 hours).

### 3.4.2 Intact Articular Surfaces

For intact surface experiments, a large box-like section of cartilage was removed from the femoropatellar groove. Using a hack saw, longitudinal cuts were made medial and lateral to the cartilage surface, with a final cut through the subchondral bone to release the cartilage/bone section. From this large section, smaller pieces were cut to fit into the testing chamber (approximately 2 × 2 × 1 cm), and then stored at 9°C until tested. Testing of the intact cartilage/bone sections was performed within two to three hours of dissection. The sections were subsequently stored at \(-20^\circ C\).
3.5 Current-Generated Stress in Cartilage

3.5.1 Measurements on cartilage discs

Following probe calibration, a cartilage disc was placed in the chamber with bathing solution, the probe lowered into it with the same static stress as in calibration, and allowed to equilibrate for at least 15 minutes. The bath solution was 0.15 M NaCl at pH 7.0 plus protease inhibitors 2 mM EDTA, 5 mM benzamidine-HCl, and 1 mM phenylmethylsulfonyl fluoride (PMSF) to minimize matrix degradation [24]. The bath was circulated and maintained at 25°C during calibration and experiment. Current densities of 1.0 and 0.5 mA/cm² were applied at frequencies of 0.025, 0.05, 0.1, 0.25, 0.5, and 1.0 Hz to the cartilage surface. The magnitude and phase of the stress sensor output voltage was recorded and converted to stresses using the transfer function determined from calibration. The disks were stored at −20°C for later biochemical analysis. A control experiment utilizing this same protocol was performed by replacing the cartilage disk with a porous polyethylene platen.

3.5.2 Measurements on Intact Cartilage/Bone Sections

After calibration, sections of articular cartilage surface were placed in the chamber, the probe lowered onto the surface with the static stress of calibration, and allowed to equilibrate for at least 15 minutes in a bath of 0.15 M NaCl/2 mM EDTA. The probe was positioned approximately normal to cartilage surface. Current was applied at the same densities and frequencies as in the disc experiment Measurements were made at different locations on the same surface. Sections were subsequently stored at −20°C.
Chapter 4

Results and Discussion

4.1 Calibration

The response of the stress sensor to applied mechanical loads is typified by the data of Figure 4-1. This particular calibration curve was obtained during the cartilage disc experiments, but demonstrates some typical sensor characteristics. The magnitude of probe output was directly proportional to the applied stress ($r > 0.995$), and diminished slightly with increasing frequency. Figure 4-2 is an enlarged section of the left channel calibration curves which shows this subtle frequency dependence more clearly. The phase angle relative to the applied load signal became more negative with increasing frequency (Figure 4-1).

The magnitudes of the transfer functions (calibration coefficients $\approx 60$ mV/kPa) for the probe of Figure 4-1 were 3-6 times higher than those typically observed for the chamber-mounted probe [3]. During preliminary design testing, the magnitude ranged from 10-60 mV/kPa. This wide range could be due to differences in ETS construction and assembly. Still, the general characteristics of these calibrations agree with those for the chamber-mounted probe. Figure 4-3 shows a calibration curve from a two electrode chamber-mounted probe. The transfer function magnitude varied more strongly with frequency in the older model.

The measured calibration coefficients obtained for the probe of Figure 4-1 were
Left Sensor Channel Calibration

Figure 4-1: Left channel calibration of the piezoelectric film as a stress sensor. Channel output (mV and phase angle relative to applied load) was recorded as dynamic stresses were applied at various amplitudes and frequencies. At each frequency, the output voltage was proportional to the applied load ($r > 0.995$), while the phase was approximately independent of load. A transfer function for each frequency was obtained from this channel and averaged with the right channel to obtain the transfer function used for experiments.
Figure 4-2: An enlarged portion of the magnitude calibration curves for the left channel calibration of the stress sensor.
Figure 4-3: Typical calibration result for the chamber-mounted probe model.
as much as 18 times greater than predicted, based on the properties of the piezo-electric film. In response to a 1 kPa stress evenly distributed over the ETS active area, Equation 2.1 predicts a developed charge, $Q$, of $161 \times 10^{-15}$ Coulombs. The total capacitance, $C_{\text{total}}$, of the probe system, including the cable and electrometer capacitances, but neglecting the capacitance of the upper probe core (valid since $\delta$ is relatively large here),

$$C_{\text{total}} = C_{\text{PVDF}} + C_{\text{cable}} + C_{\text{elect}} \approx 97 \text{ pF}. \quad (4.1)$$

Including the gain of 2 from the electrometer, the predicted measured voltage ($V = Q/C$) is 3.33 mV. This corresponds to a calibration coefficient of 3.33 mV/kPa, a value much smaller than those measured. This discrepancy could be due to incomplete characterization of the PVDF material at very low frequencies. It is also possible that the local stress in the area of the stress sensor electrode is much higher than the applied load normalized to the active area of the ETS. A stress concentration could be introduced by the raised 1mm thick silver electrodes opposite from the stress sensor or through uneven application of epoxy during ETS construction. Nevertheless, once the calibration transfer functions are measured, the signals from the sensor during current-generated stress measurements can be unambiguously quantified in terms of a stress measurement.

The output of the sensor was found to be sensitive to the offset load and ETS irregularities. As static stress was increased, the sensor output (mV/kPa) diminished. Because the ETS must be bent to fit into the probe body, folds and creases are sometimes introduced which can severely affect the output. On some occasions, the output from the separate channels was extremely uneven and distorted. The channels would have unequal amplitudes and be out of phase with each other during calibration. An increase in static stress would sometimes remedy the situation, at the cost of decreasing sensor output. A possible cause for this phenomena might be folds or creases that get pressed out at a high enough static stress. At low static stresses, these bumps and folds could cause local stress concentrations that introduce significant
artifacts into the measured sensor signal.

### 4.2 Current-Generated Stress in Cartilage Discs

Small periodic current densities (0.5–1.0 mA/cm²) at frequencies of 0.025 to 1.0 Hz applied to the surface of calf cartilage discs induced mechanical stresses at the same frequencies. Figure 4-4 shows the dependence of the current-generated stress on applied current density and frequency. The data are mean ± SD for 6 discs taken from the same calf knee joint. Two plugs were used, each generating three slices for a total of 6 discs. For each disc, the magnitude of the stress was proportional to the applied current density and inversely proportional to the frequency.

A control experiment was performed with the cartilage replaced by a porous polyethylene disc. Current densities of 0.5, 1.0, and 2.0 mA/cm² were applied and the output from the stress sensor was measured and converted to stress (Figure 4-5). Periodic stresses at the same frequency as applied current were measured at magnitudes 2–20% of the response with cartilage. Similar to cartilage responses, the stress amplitude was inversely proportional to frequency. The phase angle was also independent of current density, but phase angles were slightly larger (clustered near 120°). The measured stress (at a particular applied current frequency) as a percent of the corresponding stress with cartilage was proportional to frequency and ranged from 2% at 0.025 Hz to 20% at 1.0 Hz.

The current-generated stress measurements on cartilage discs agree with theory and previous measurements. The dependence of current-generated stress on the frequency and amplitude of an applied current is consistent with the mathematical model [27, 28]. The model predicts a lower stress for higher excitation frequencies because the fluid flow contributing to the stress has less time to develop. These predictions are consistent with the probe measurements. Also, these measurements are consistent with those taken with the chamber-mounted model. Figure 4-6 compares the values of current-generated stress verses frequency from the two probes. The measured magnitudes from the current probe are 5 to 7 times higher than values measured
Figure 4-4: Amplitude and phase of the fundamental component of measured stress (converted from sensor voltage with the calibration transfer function) as a function of applied current density and frequency. Data are mean ± SD for 6 discs.
Figure 4-5: Results of a control study which replaced cartilage with a porous polyethylene disc. Amplitude and phase of the fundamental component of measured stress as a function of applied current density and excitation frequency.
with the chamber-mounted model, while the phase angles are in good agreement. This higher stress magnitude could be due to a differences in the physical properties of the samples measured. The GAG content (fixed charge density) and hydration of the calf cartilage samples tested here is compared to the adult samples tested with the chamber-mounted probe in Figure 4-7. The higher GAG content of the calf discs could partially account for the higher current-generated stress measured with the intact-surface probe. Other material properties, including the hydraulic permeability and equilibrium modulus (both related to hydration and GAG content) would affect the magnitude of current-generated stress.

The imposed spatial wavelength, \( \lambda \), equal to twice the center-to-center spacing of the excitation electrodes affects the depth that the applied current will penetrate. According to the mathematical model, the depth of penetration is from \( \lambda/5 \) to \( \lambda/3 \). For this design, \( \lambda = 6 \) mm which corresponds to penetration depths from 1.2–2 mm. For the 1 mm thick cartilage discs tested (\( \delta = 1 \) mm), the current penetrated through the full thickness and resulted in electroosmotic fluid flow throughout the disc. The spatial wavelength of the chamber-mounted probe was 5 mm which corresponded to penetration depths from 1–1.67 mm. The current in these experiments also penetrated the full thickness of the discs tested (\( \delta = .8 \) mm). Yet, there could be differences in the distribution of current densities and resultant current-generated stress because of differences in \( \lambda/\delta \) between the two probes.

The control experiment performed with the porous polyethylene disc resulted in measured stress values 2-20% of those typically measured with cartilage. At the lower frequencies (0.025 to 0.1 Hz), this artifact was small compared to stress measurements with cartilage (2-9%). At higher frequencies (0.25 to 1.0 Hz), the artifact signal was a larger percentage of the cartilage measurement (12-20%). The artifact is directly proportional to the current density, and becomes more significant at larger currents. For example, at 1 Hz a current density of 0.5 mA/cm\(^2\) resulted in an artifact stress 10% of the average stress measured from the cartilage disks, but rose to 20% at 1.0 mA/cm\(^2\). This artifact could be due, in part, to capacitive coupling between the applied current and stress sensing electrodes, and, in addition, to a bending of the
Figure 4-6: Current-generated stress measurements vs. frequency comparing the current design and chamber-mounted model. Chamber-mounted results are 800 μm thick adult bovine discs (N=9), while the other data is for calf cartilage discs (N=6). Data are mean±SD.
Figure 4-7: Comparison of GAG content and hydration of the calf discs tested in this experiment with the adult cartilage discs tested with the chamber-mounted probe. Data are mean±SD, N=6 for the calf discs and N=5 for the adult cartilage discs.
excitation electrodes as current is passed through them (an \textit{electrocapillary} effect, for details see [11]).

\section*{4.3 Current-Generated Stress in Intact Articular Cartilage-Bone Joint Surfaces}

To approach the conditions in an intact joint, initial measurements were made using sections of intact femoropatellar groove sections from bovine joints. Figure 4-8 shows the results of an experiment on such an intact cartilage-bone joint surface. The current-generated stress was proportional to current density and inversely proportional to frequency with values similar to those obtained in the disc experiment (Figure 4-4). It is important to note that the cartilage surface is intact. This is in contrast to previous electrokinetic measurements of cartilage (including chamber-mounted model) which involved cutting off the top 80 \textmu m of cartilage tissue in order to obtain plane parallel slices. This surface layer is significantly different from the bulk of cartilage tissue, yet including it does not appear to dramatically change the current-generated stress measurement. This suggests that meaningful nondestructive measurements can be made on cartilage joint surfaces \textit{in situ} and eventually on articular cartilage \textit{in vivo}.

\section*{4.4 Summary and Future Work}

The results from these experiments validate the intact surface probe as an instrument for measuring current-generated stress on intact articular cartilage surfaces. The magnitude of current-generated stresses are higher than previously reported, but display the trends associated with previous measurements and theory. Also, the phase of current-generated stress measurements with the intact-surface probe on both the extracted discs and intact surfaces agree very well with theory and earlier experiments.

The design presented here might be improved with certain modifications and can be scaled down to the size of a typical arthroscopic cannula (3 - 5 mm diameter) to
Figure 4-8: Amplitude and phase of the fundamental component of measured stress on an intact articular surface as a function of applied current density and frequency.
make \textit{in vivo} measurements. Problematic bending of the piezoelectric stress sensor electrodes could be avoided by making the contact to these electrodes on the flat surface of the core, rather than on the beveled edge. Although this would not be a clamped contact, pressure from the static offset stress might be sufficient for electrical connection. This would involve fabricating a new upper-core section with the brass contacts moved to the flat surface (see Figure 2-5) and would actually simplify its construction. \textit{In vivo} measurements would require a way of maintaining a fairly constant offset load and orienting the probe flat against the articular surface. An advantage of using sinusoidally varying excitation currents is that signal processing can filter out static offset drift, or other artifacts that occur at frequencies other than that of the excitation current (and current-generated stress).

It might be possible to use current microfabrication techniques to incorporate excitation electrodes and a piezoelectric stress sensor on a silicon chip. This would offer many distinct advantages including on-board signal amplification and processing that could increase sensitivity and signal-to-noise ratio. A probe incorporating these techniques would also decrease the variance associated with fabricating and mounting each ETS by hand. Microfabrication, along with micromachine technology, is currently being investigated for possible future incorporation into the intact-surface probe system.
Appendix A

Design Iterations

This appendix contains a design report on the probe, including previous designs and problems encountered, written in October 1994. The final design described here has since been modified slightly to arrive at the design described in Chapter 2 of this thesis.
1. Problem Statement:

To allow nondestructive measurements of current-generated stress in intact synovial joints.

2. Design Specifications:

- Size: Probe must be small enough to fit into a joint space. The insertion of the probe into a joint space will be through an arthoscopic cannula. Typical arthoscopic cannulas are between 3 and 5 millimeters in diameter.

- Sealing: The probe must be able to operate in aqueous solution (submerged in the synovial joint fluid).

- Electromagnetic shielding: The piezoelectric film electrodes used to detect the current generated mechanical stress must be shielded from the electric fields produced by the cartilage stimulation electrodes, and from any other stray electromagnetic fields.

- Maneuverable: The probe must be able to orient itself onto the curved articular cartilage surface.

- Consistent, reliable performance is required.

3. Probe Design

We are currently at version 3 of the intact-surface probe system (see page 4). Below are the two previous designs as well as the current design that we have developed. The third version is in the initial testing phase and results should be available within the next few weeks.
A.1 Probe I

The original prototype incorporates a transducer structure similar to that used in our *in vitro* chamber-mounted prototype (Berkenblit et al., J. Biomechanical Engineering, 1994, in press), but placed on a cylindrical plug. Standard photofabrication procedures with slight modifications are used to create the electrode patterns. The transducer consists of a three layer laminated design (Figure 1b): silver electrodes bonded to the mylar side of a metal coated mylar sheet, and the Kynar mounted with conducting epoxy to the metal coated side of the same sheet of mylar. The Kynar stress sensor is cut into a circular shape before being mounted on the metal coating of the mylar (Figure 1a). The silver electrodes are equipped with leads (Figure 1c) that fold down over a Plexiglas plug (Figure 2b) to allow for connections to excitation wires. Connections to the Kynar electrodes are made through the Plexiglas plug that the transducer structure is mounted on. These electrical connections supply the mechanical force that holds the transducer structure to the plug.

A.1.1 Design Description

Electrical connections to transducer:

- Silver electrode connection: silver conducting epoxy connects excitation wire to leads on the silver electrode pattern (Figure 2a).

- Kynar connection: silver conducting epoxy joining coaxial conductor to electrode pattern on Kynar (Figure 2b).

Shielding:

- Aluminum tube acts as the shielding for Kynar (Figure 2a).

- Coaxial cable shielding connects to a metal plug which contacts the aluminum tube (Figure 2b).

- Conducting silver epoxy applied around end of tube completes the Kynar shielding (Figure 2a).
Sealing:

- RTV silicone rubber adhesive sealant applied around the circumference of the end of the aluminum tube (Figure 2a).

**A.1.2 Problems Encountered**

A. Electrical connection to Kynar not reliable.

- Aligning electrodes with holes in Plexiglas plug filled with conducting epoxy was difficult and inconsistent.

- Smearing of conducting epoxy can short the Kynar circuit.

B. Uneven application of conducting epoxy (due to “mounds” of epoxy at electrodes) creates an uneven surface and thus a non-uniform pressure distribution across the transducer.

C. Shielding unreliable due to inconsistent, manual application of conducting epoxy around the circumference of tube end (between plug and tube).

D. Seal around the circumference of the tube end unreliable because of inevitable gaps in the silicon rubber adhesive material applied.

**A.1.3 Conclusion**

Design inadequate. Mechanical calibration data for the right and left channels was obtained for a Probe I model. Mechanical stresses from 1-10 kPa at frequencies ranging from 0.025 to 1 Hz were applied and the amplitude and phase of the piezo film voltage output was recorded. These measurements showed that the channels behaved non-uniformly as described above. Current-generated stress in cartilage was also measured. Although the expected trends were observed, the response was far from ideal, and some investigation revealed that a part of this response was due to coupling from the excitation electrode electric field as a result of inadequate shielding. The probe design was modified to address problems A and B in particular (see Probe II).
a) Piezo electrode pattern

Piezo sensor electrodes

b) Transducer Layers (laminated together)

Piezo sensor (Kynar: PVDF with thin sputtered Ni Co)
Metal coated mylar (ground plane - shielding)
Silver excitation electrodes (.001" thick silver sheet)

c) Excitation electrode pattern (contacts cartilage surface)

Silver excitation electrodes
Leads

2 cm

Figure 1: Probe I transducer structure
(cut along dotted lines before mounting)
Figure 2: Probe I system
A.2 Probe II

The major design change in this probe version is the movement of all electrical connections to the inside of the tube. To accomplish this, the Kynar electrodes were equipped with leads (Figure 3a), and the silver electrode leads were rotated 90° with respect to the Kynar electrodes (Figure 3c). Having the electrical connections moved from the end of the plug allowed the transducer to be bonded to the plug surface with a thin, even layer of epoxy (Figure 4b).

A.2.1 Design Description

Electrical connections to transducer

- Silver electrode connection: pressure contact of the silver leads against an outer tube of polyamide equipped with strips of copper tape soldered to wires (Figure 4a).

- Kynar connection: pressure contact of Kynar leads against contacts on the inner Plexiglas plug (connected to coaxial cables) via an inner tube of metal coated mylar around the leads and plug (Figure 4b).

Shielding

- Accomplished by the metal coating on the inner mylar tube in contact with the Plexiglas plug metal coating. This tube contacts the side of the Kynar leads which is the transducer structure ground plane (Figure 3b and 4b).

Sealing

- Application of various adhesive sealing/potting agents around the end of the aluminum tube, keeping the transducer surface extended beyond all other components to allow unobstructed contact with the cartilage surface (Figure 4b).
A.2.2 Problems Encountered

A. Inadequate shielding. Although an improvement over Probe I, the shielding does allow a gap between the end of the metal coated tube and the transducer ground plane (Figure 4b). This was evident from strong coupling of excitation outputs and Kynar outputs during initial experiments.

B. Inadequate sealing. None of the sealants applied to the multi-tube, open end design would consistently seal the tube from the bath solutions, resulting in low resistance pathways for both Kynar and excitation circuits (electrical short).

A.2.3 Conclusion

Design inadequate. Problems A and B addressed in Probe III.
a) Piezo electrode pattern

b) Transducer Layers (laminated together)

(c) Excitation electrode pattern (contacts cartilage surface)

Figure 3: Probe II transducer structure
(cut along dotted lines before mounting)
Figure 4: Probe II System
A.3 Probe III

This design utilizes a mechanical seal and clamped contacts to assure sealing and electrical contact. The transducer design is similar to probe II, with shorter Kynar leads (Figure 5a) and longer silver leads (Figure 5c). The total electrode active area is smaller (Figures 5a and 5c). The mechanical seal and clamped contacts to the Kynar are accomplished by applying force through a screw in the body of the stainless steel tube to a beveled edge at the end of the tube (Figure 6a). Silver electrode clamped contact is made through a spring clip (Figures 4c and 4d).

A.3.1 Design Description

Electrical connections to transducer

- Transducer is clamped around its circumference against the beveled end of a stainless steel tube (Figure 6a).

- Kynar connection: a clamped contact between brass contacts on the inner polyamide-imid (PAI) plug and the beveled edge of a PAI sleeve (Figure 6b).

- Silver electrode connection: silver lead is slipped between a spring clip and the aluminum sleeve which surrounds the PAI plug (Figure 6c). The spring clip is attached to the second plug (Figure 6d).

Shielding

- Accomplished by a machined aluminum sleeve press fit onto the PAI plug and cemented to the coaxial cable shielding with conductive epoxy (Figure 6b). The connection from the ground plane of the transducer to this shielding is through the back of the silver lead which is pressed against the aluminum tube by the spring clip (Figure 6c).

Sealing

- A PAI sleeve with the same beveled end as the PAI plug and steel tube maintains a seal through the clamping action of the screw (Figure 6b). The transducer fits
between the plug and the sleeve. The clamping force is transmitted to the PAI plug from the screw at the end of the tube (Figure 6d). This plastic sleeve also keeps the silver leads and the steel tube out of electrical contact.

A.3.2 Conclusions

All the components for this design have been fabricated and assembled. Calibration and current-generated stress measurements are now in progress.
Figure 5: Probe III transducer structure
(cut along dotted lines before mounting)
a) Top and cross-section (Outer tube)

b) Top and cross-section

Beveled end
Stainless steel tube
Screw

Kynar electrodes
Brass contacts
Plastic sleeve (PAI)
Plastic plug (PAI)
Coaxial shielding
Coaxial cables

Silver electrodes

Aluminum plug shielding
Spring clip
Excitation wire

c) Top and cross-section (rotated 90)

d) Second plug: side and bottom

Figure 6: Probe III system
Bibliography


