

# Large-Area Flexible-Array Piezoelectric Ceramic/Polymer Composite Transducer for Bone Healing Acceleration

I.A. Cornejo<sup>1</sup>, B. Jadidian<sup>1</sup>, A.A. Winder<sup>2</sup>, and A. Safari<sup>1</sup>

<sup>1</sup>Department of Ceramic and Materials Engineering, Rutgers University  
607 Taylor Rd., Piscataway, NJ 08854, USA

<sup>2</sup>Exogen Inc.

10 Constitution Ave., Piscataway, NJ 08855, USA

*Abstract* - The processing, electromechanical and acoustical properties of large-area flexible-array composite transducers that generate pulsed and low intensity ultrasonic vibrations for accelerating the healing of bone fractures is presented. The device is composed of a PZT/polymer piezoelectric array, matching layer, and housing. The array is made of several single composite elements arranged in a matrix form. Each element has a minimum flexibility of 2mm deflection. The flexibility of the entire device permits it to conform to the natural contours of the human body. The piezoelectric elements were processed by two different routes: Dice & Fill and Woven Fibers. A ceramic volume fraction and resonance frequency of 30% and 1MHz was maintained.

## INTRODUCTION

The discovery of the piezoelectric effect by Pierre and Jacques Curie in 1880 and the observation by Wolff in 1892 that bone tissue is biologically active and would respond to micromechanical stress produced by acoustic pressure waves [1] can be considered as the beginning of diagnostic and therapeutic ultrasound. For the past fifty years, most notably since 1970, ultrasound has been demonstrated to stimulate the repair of soft tissue wounds and bone fractures, and to relieve accompanying pain. Some of the significant findings in this field can be attributed to Dyson [2-4], Duarte [5], Pilla [6], Heckman [7], and Kristiansen [8]. An exciting discovery was that non-therapeutic intensity levels, i.e., several orders of magnitude lower than 1 watt/cm<sup>2</sup>, were extremely effective in bone healing and general tissue wound repair. The mechanisms behind these results are well beyond the scope of this article (the exact cellular processes for biological tissue repair are still unknown).

Piezoelectricity is defined as the ability of certain materials to develop an electrical polarization proportional to the applied mechanical stress [9]. This is the *direct* piezoelectric effect, where in tensor notation can be expressed as follows:

$$P_i = d_{ijk} \sigma_{jk}, \quad (1)$$

where  $\sigma_{ij}$  is the applied stress,  $P_i$  is the developed net polarization, and  $d_{ijk}$  is the piezoelectric coefficient. In addition, there is a *converse* piezoelectric effect, where an applied electric field  $E_k$  develops a strain  $\varepsilon_{ij}$  in the material. In tensor notation,

$$\varepsilon_{ij} = d_{ijk} E_k, \quad (2)$$

where  $d_{ijk}$  is the piezoelectric coefficient. In both of these effects, the  $d_{ijk}$  coefficients are the same. High  $d$ -

coefficients are necessary for materials intended to develop motion or vibration such as acoustic projectors.

Another important piezoelectric constant is the piezoelectric voltage coefficient  $g_{ijk}$ , which gives the electrical field produced by an applied stress.

$$E_i = g_{ijk} \sigma_{jk}. \quad (3)$$

This coefficient can also be defined as:

$$g_{ijk} = \frac{d_{ijk}}{k_{ii} \varepsilon_0}, \quad (4)$$

where  $k_{ii}$  is the dielectric constant of the material in the direction of the electric field, and  $\varepsilon_0$  is the permittivity of free space ( $8.854 \times 10^{-12}$  F/m). High  $g$ -constants are desirable for receiving devices such as phonograph pickup and hydrophones.

When utilizing both the direct and converse effects, the transducer can be used for applications such as medical diagnostic imaging, therapeutic treatments, and non-destructive evaluation. In these applications, the transducer generally performs in a pulse-echo mode, which sends out an acoustic pulse and detects the echoes produced by reflections coming from the tested object. The echoes are caused by differences in acoustic impedance between materials. In these applications, high  $d_{33}$  and  $g_{33}$  coefficients are required.

Lead zirconate titanate (PZT) ceramic suffers from several disadvantages when used as a hydrophone or pulse-echo transducer for medical applications. First, its hydrostatic piezoelectric coefficient  $d_h$  ( $=d_{33}+2d_{31}$ ) is low due to the opposite signs of the piezoelectric coefficients  $d_{33}$  and  $d_{31}$ . Second, the voltage coefficient  $g_{33}$  and  $g_h$  ( $=d_h/\varepsilon_0 k$ ) are small due to the large dielectric constant of PZT. Furthermore, its characteristic acoustic impedance  $Z$  ( $=\rho v$ , where  $\rho$  is the density and  $v$  the longitudinal sound velocity) is high ( $\sim 30$ MRayls) compared to that of the body and water ( $\sim 1.5$ MRayls), resulting in an acoustic impedance mismatch. Then, the acoustic transmission coefficient at the transducer-load interface is low. To optimize the magnitude of the acoustic and electromechanical properties, a number of PZT-polymer composites have been studied [10]. Piezocomposites have shown superior properties when compared to single phase materials. They combine high coupling thickness coefficient, low acoustic impedance and dielectric constant, and intermediate dielectric losses. In addition, their flexibility can be easily tailored and are moderately priced [11-12].

## EXPERIMENTAL PROCEDURE

The development of large-area flexible-array PZT-composite transducers involve several processing steps:

- *Polymer Selection and Mechanical Testing*

In this task, Spurr epoxy (Ernest F. Fullam Inc., Schenectady, NY) and Eccogel epoxy resins (series 1365-65) (Emerson and Cumming, Deway and Almay Chemical division, Canton, MA) were used. Spurr epoxy was modified to improve its flexibility by increasing the weight fraction of one of its components. Eccogel epoxy is naturally soft and flexible.

Based primarily on polymer flexibility, two polymers were used to make the piezocomposites for mechanical testing. The four point bending method was performed on composites by using an Instron testing device. Composite samples were prepared in two different geometry configurations. First, spiral and 2-2 diced PZT composites were formed as single elements. Second, the ceramics were arranged in a  $1 \times 3$  array with 1 mm spacing between them. Then, two sets of elements were filled with modified Spurr and Eccogel polymers and cured. After polishing the structures to 1 mm thickness, bars of  $0.5 \times 4.0 \times 0.1 \text{ cm}^3$  (for the single-element composites) and  $1.0 \times 8.0 \times 0.1 \text{ cm}^3$  (for the 3 element arrays, where each element length was 25.4 mm), were cut. Three sample bars were prepared for each type of polymer. Electromechanical properties of all samples were tested before and after the mechanical test.

- *Flexible Matching Layers*

Solid E-glass spheres (Potter Industries Inc.) with mean particle sizes of  $2 \mu\text{m}$  were selected as the ceramic filler. This filler was added to Eccogel epoxy. Matching layers, containing 0 to 30 volume percent filler, were prepared for acoustic testing in increments of 5 volume percent filler. The proper amounts of Eccogel epoxy and spherical glass were weighed and hand-mixed. The mixture was then further mixed with a high shear mixer for 20 minutes. Then, the mixture was poured into a hexagonal plastic dish of 13 cm diagonal, and placed in a vacuum chamber for degassing. The samples were held at  $-760 \text{ mmHg}$  for 1 hour with several flushing in between to prevent excessive foaming in the mixture. However, for the samples with more than 20 volume percent filler, it was necessary to increment the holding time under vacuum to 4 hours to eliminate all trapped air bubbles. Next, the samples were put in a dryer and cured at  $72 \text{ }^\circ\text{C}$  for 12 hours.

Large samples in diameter ensured that their dimensions were at least ten times the wavelength at which the acoustic measurements were made minimizing diffraction effects. These samples were polished by a diamond saw (Kulicke & Soffa Industries Inc., Model 775 wafer saw, Horsham, PA) to ensure parallel surfaces. The acoustic impedance and attenuation measurements were performed by using a modified Selfridge method [13]. The sample was placed in a sample holder which provided x, z and  $\theta$  degrees of freedom. This allowed the surface of the sample to be parallel to the surface of the transducer, which was partially submerged in DDD-water. A non-focused, 1MHz center frequency transducer (Panametrics V314-SU) was used in the acoustic tests. The dimensions of the acoustic test tank were 32.4cm wide, 76.2cm long and 47.6cm deep, more than 100

wavelengths in either direction at the test frequency, establishing free-field acoustic propagation.

- *Transducer from PZT fibers*

Advanced Cerametrics Inc. (Lambertville, NJ) has developed a cost-effective method for producing large quantity of continuous PZT ceramic fibers, using rayon technology [15]. This technique, called Viscous Suspension Spinning Process (VSSP), uses cellulose as the fugitive carrier for the production of ceramic fibers.

Frequently, ceramic additive [16] have been used to modify the properties of rayon tows. To generate ceramic fibers, 75-90 wt% ceramic is added to the viscose and the resulting spin mix produces green fibers with 50 to 76 vol.% ceramic-to-carrier loading. In this work, PZT-5H powder (Morgan Matroc Inc., Bedford, OH) was applied to viscose to form a spinnable suspension.

Green PZT fibers required sizing to increase their strength for handling [17]. Polyvinyl alcohol (PVA) and polyvinyl acetate (PVAc) are two low-cost water-soluble polymers with good film-forming ability. In this study, PVA 107 and PVA 205 (Air Products and Chemicals, Inc., Allentown, PA) and polyethylene glycol (PEG) (Carbowax series 200-1450, Union Carbide Chemicals and Plastics Company Inc., Danbury, CT) were used as major and minor sizing components, respectively. Ethylene glycol (EG) (Fisher Scientific Co., Chemical Manufacturing Division, Fair Lawn, NJ), a plasticizer for both PVA and PEG, was added to the sizing solution in order to increase the flexibility of sized fibers.

PZT fibers were soaked in the binder and passed through a steel sizing die to compact and align the fiber bundle. The sizing die (Hoosier Ajax, Fort Wayne, IN) had  $750 \mu\text{m}$  diameter to size 800 individual filaments. The product was a collimated multifilament tow (a bundle of hundreds of fibers), dried at room temperature. Then, green continuous sized tows containing 800 fibers were woven into plain fabrics. The fabric was cut into  $4 \times 150 \text{ cm}^2$  strips, hand-rolled tightly, and wrapped with bundles of PZT fibers to support the structure and to keep shape during firing. This structure is called spiral.

For heat treatment, the spiral structures were placed on platinum foils sitting on an alumina crucible. The entire arrangement was placed into a furnace for binder burn out (BBO). In the first step of heat treatment, the spirals were heated at  $550 \text{ }^\circ\text{C}$  for 4 hours with a  $1.5^\circ\text{C}/\text{min}$  heating. Then, a pre-firing was performed at  $780^\circ\text{C}$  for 1 hour at  $3.5^\circ\text{C}/\text{min}$ , to strengthen the spiral structure for handling. In the second step of heat treatment, the spirals were sintered at  $1285^\circ\text{C}$  for 40 minutes with a  $3.5^\circ\text{C}/\text{min}$  ramp in a sealed alumina crucible under lead-rich atmosphere. After sintering, the spirals were placed in plastic dishes and filled with Eccogel, de-aired at  $-760 \text{ mmHg}$  for 30 minutes in a vacuum chamber. Then, the dishes were placed in an oven at  $72^\circ\text{C}$  for 12 hours for curing. The chamber was slowly cooled to room temperature to prevent warping and cracking of samples.

The cured spiral composites were sliced into 2.5 mm thick samples and then polished to a thickness of 2 mm with 240, 400, and 600 grit SiC sand papers. One-inch diameter discs (2.54 cm) were cut out of each sample.

The composites were electroded with silver paint on one side and poled via the corona method [14]. In this technique, the unelectroded top surface of the sample is exposed to a shower of positive ions while the bottom surface is grounded to a metallic plate placed on a hot plate. A potential of 26 kV was applied to the corona needles tips, which was 45 mm away from the sample. Poling was done for 20 minutes and the temperature was dropped from 60 to 45°C during the poling process.

After poling, the other sides of spiral composites were electroded with air-dried silver paint and dried at room temperature. Samples were aged for 24 hours, before electromechanical evaluation. The capacitance ( $C_p$ ) and dissipation factor ( $\tan\delta$ ) were measured at 1 kHz with a 1689M precision RLC Digibridge (GenRad Inc., Boston, MA). The dielectric constant  $K$  of a sample of thickness  $t$  was calculated by:

$$k = \frac{C_p t}{\epsilon_0 A} \quad (5)$$

where  $A$  is the electroded area. The piezoelectric charge coefficient  $d_{33}$  of the composites was measured at 100 Hz using a Berlincourt Piezo  $d_{33}$ -Meter (CPDT-3300-Chanel Product, Inc., Cleveland, OH). Two flat probes were used to test the composite and two round probes were used for the ceramic samples. The thickness and planar coupling coefficients were calculated by measuring the resonance and antiresonance frequencies with an Impedance/Gain-Phase Analyzer (4194A Hewlett-Packard Inc., Palo Alto, CA).

Three twelve-element (77.42 cm<sup>2</sup>) transducer arrays were fabricated using these elements. The elements were glued in a plastic dish (with same polarity up or down) in the specified 4 x 3 arrangement (with an inter-element spacing of 1mm) to prevent any displacement during the curing stage. The arrays were back-filled with Eccogel, de-aired, and cured as discussed before. The cured array composites were polished to desired thickness corresponding to 1 MHz resonance frequency. Polishing was performed using a 2 mm thick 340-mesh diamond blade (Diacut Inc., Palmer Lake CO). The blade swept the surface of array composite with 1-millimeter increment in x-y plane and 100 μm depth in z-direction.

The arrays were re-poled by the same technique mentioned before. Re-poling was necessary due to partial thermal and mechanical depolarization, caused by the processing steps. Thermal depolarization may occur upon curing the polymer and also during the gold sputtering process. Mechanical depolarization is attributed to the applied stress during polishing, in which PZT fibers were under compression. However, this step could also contribute to thermal depolarization resulting from the friction between sample and polishing blade. The negative sides of arrays were dc-gold sputtered for 4 min (Polaron Instruments, SE Coating Unit E5100, Doylestown, PA) to form common electrodes. Individual elements on the top surface of the arrays (the positive sides) were gold sputtered utilizing proper masks to cover the spacing between transducer elements.

- *Transducers from PZT Diced Ceramic Discs*

Several processing steps were similar to the spiral process, such as BBO, poling, electroding, and polishing,

that will not be repeated here, unless necessary. All the samples were processed from Morgan Matroc, Inc. PZT-5H powders mixed with 6 wt% of PVA binder solution (PVA/ water ratio was 20/80) in a mortar and pestle and then sieved through a 70 mesh sieve.

Pellets of diameter 4.45 cm were pressed uniaxially at 152 MPa (22 ksi). The average green density was computed to be 4.58±0.02 g/cc which is 58 per cent of the theoretical density of PZT. These pressed samples were subjected to a slow BBO cycle. Sintering was carried out at 1285°C for 1 hour in a closed and sealed Al<sub>2</sub>O<sub>3</sub> container. A mixture of PbO+ZrO<sub>2</sub> was used as a lead source to compensate for lead loss from the samples. In this step, the samples rested on a bed of PZT-5H coarse powder (0.5-1.0 mm). An average lead loss in the samples of 0.30 ± 0.06 wt.% was found after sintering. The mean sintering density, by measuring the mass and dimensions of the sample, was found to be 7.45±0.01 g/cc (~94.7±0.2% of the theoretical value).

The sintered samples were blade-polished. The polished surfaces were smoothed with a 1200 SiC grit paper. Silver paint was applied on one side of the pellets and poled by corona method at ~25kV for 15 minutes at 70°C. Then, the silver was removed with acetone and both faces of the samples were gold sputtered. The samples were left to age for 24 hours at room temperature with the gold electrodes short circuited, before electromechanical characterization.

Dielectric constant,  $\tan \delta$ , and  $d_{33}$  piezoelectric coefficients were measured. Coupling coefficient  $k_t$  and  $k_p$  were obtained from the resonance and antiresonance methods. The average dielectric constant, dissipation factor,  $d_{33}$ ,  $k_t$  and  $k_p$  coefficients of the PZT-5H pellets were 3020±36, 0.0187±0.0005, 608±5 pC/N, 56%, and 73% respectively.

The poled samples were cut into a 6.452 cm<sup>2</sup> (1in<sup>2</sup>) plate and diced into three different connectivity patterns: 2-2, 2-(1-3)-2, and 1-3. Due to poor adhesion between PZT diced ceramic and Eccogel, especially in elements with 2-2 connectivity; a low viscosity hard Spurr polymer was used to refill the diced ceramic elements. The polymer was then removed from the ceramic, leaving a thin Spurr film behind. Curing of this film at 70 °C for 12 hours was done before back filling the arranged 4x3 array with Eccogel. This processing procedure significantly improved the mechanical stability of the whole array. After air-gas removal, the whole array was cured, polished, and gold electroded.

## RESULTS AND DISCUSSION

Polymer selection was one of the most important considerations to ensure device integrity at the end of the fabrication process, and even further, during service. In addition, the polymer selected must be flexible, biocompatible, form a strong bond with the ceramic phase to prevent failure at the ceramic-polymer interface due to repetitive bending, and have low to moderate viscosity to fill very fine pores and spaces present in the ceramic phase. All the samples tested supported a 2 mm bend without failure, except the 2-2 Horizontal single

elements, as expected. This was the maximum displacement measurable by the Instron device. Furthermore, no significant changes in the electromechanical properties were observed in the composites after mechanical testing, even for the broken 2-2 horizontal single elements. These results, in part, resulted in Eccogel to be chosen as the most suitable polymer for the desired application. This selection was based on more general criteria, which is beyond the scope of this article.

Flexible matching layers were prepared as mentioned in the Experimental Procedure Section. The design achieved biocompatibility, low acoustic attenuation, and a strong bond at the interface with the array. The acoustic impedance of the matching layer for optimum acoustic transmission was calculated by the relation [18]:

$$Z_m = (2 Z_b^2 Z_t)^{1/3} \quad (6)$$

where  $Z_m$ ,  $Z_b$ ,  $Z_t$  are acoustic impedance of matching layer, human body ( $\approx 1.5$  MRayls), and transducer element, respectively<sup>1</sup>. An average acoustic impedance of 8 to 10 MRayls was measured for all the elements. Employing Equation (6), the calculated matching-layer impedance must fall between 3.3 and 3.6 MRayls to ensure proper matching with the human body.

Most polymers used for piezocomposites have densities lower than  $2 \text{ g/cm}^3$ , and thus, their acoustic impedance is less than 2.6 MRayls. To increase density as well as the acoustic impedance of the polymer, it is necessary to add a suitable filler. M. Grewe [19] found that the dispersion of filler of various volume fractions in epoxy yielded matching layers with 0-3 connectivity. Grewe also showed that silica glass dispersed in an epoxy had lower acoustic impedance than tungsten, tungsten carbide, PZT and alumina. In addition, the more uniformly dispersed the filler particles are in the epoxy, the lower the acoustic attenuation for the 0-3 composites. Figure 1 shows the acoustical properties of different matching layers (Eccogel and composites) measured at 1 MHz. By varying the filler's volume fraction ( $V_f$ ) of the Eccogel composite (0-30 %), acoustical impedance can be tailored from 2.889 to 3.475 MRayls while keeping the attenuation relatively low.

Figure 2 shows SEM photographs of single filaments before and after sintering. This firing step produces well sintered fibers with diameters of  $\approx 30\mu\text{m}$  ( $\approx 20\%$  shrinkage in diameter) and grain sizes below  $10\mu\text{m}$ .

Figure 3 shows a spiral array before polishing to the specified thickness. Each element in this array presented an average density,  $\tan \delta$ , dielectric constant,  $d_{33}$ ,  $k_t$ ,  $\nu$ , and resonance frequency of 3.3 g/cc, 0.04, 340, 400 pC/N, 64%, 2670 m/s, and 1.1 MHz, respectively.

For the arrays fabricated from the dicing&fill technique (see Figure 4), three different connectivity patterns were investigated: 2-2, modified 2-2 (or 2-(1-3)-2), and 1-3. The thickness of the ceramic wall ( $p$ ), the spacing ( $e$ ) between them and the connectivity were

<sup>1</sup> This relationship was derived requiring the serial and parallel mechanical Q factors of the equivalent Mason circuit of the transducer to be equal.

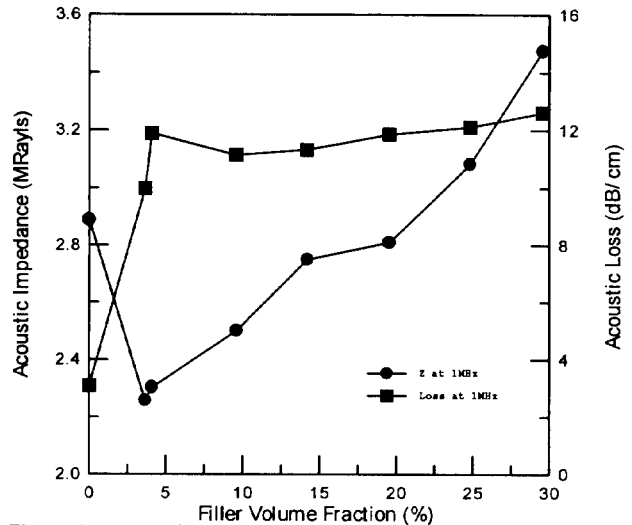


Figure 1. Acoustic Impedance and Loss at 1 MHz measured as function of filler volume fraction

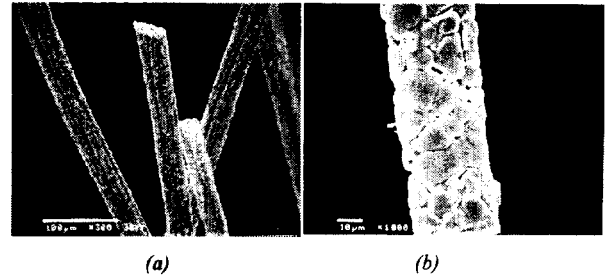


Figure 2. SEM pictures of single filaments (a) before and (b) after firing at 1285°C under lead controlled atmosphere.

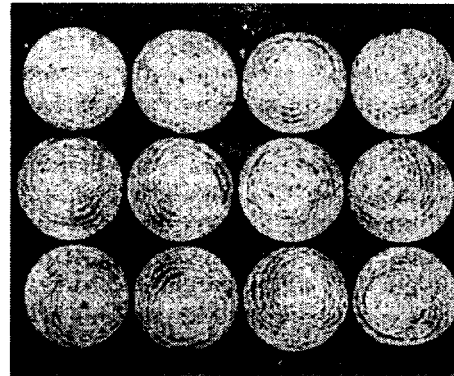


Figure 3. Photograph of a 4 x 3 flexible array transducer before polishing to the desired thickness.

chosen to achieve the desired direction of flexibility and the volume fraction of the final element.

The volume fraction of the piezoelectric diced ceramic ( $V'_c$ ) is related to  $p$  and  $e$  by:

$$V'_c = \frac{p}{p+e} \times \frac{p'}{p'+e} \quad (7)$$

The primed terms indicate wall and spacing dimensions, cut in different perpendicular directions. Notice that  $p$  and  $p'$ , as well as  $e$  and  $e'$ , may or may not be equal. A unique feature of diced composites is that volume fraction can be tailored a priori by the proper selection of these parameters.

Usually, re-poling was not necessary in these diced composite arrays. Typical electromechanical properties of a PZT 2-(1-3)-2 array were:  $K=425$ ,  $\tan \delta=0.02$ ,  $d_{33}=400 \text{ pC/N}$ ,  $k_t=63\%$ ,  $\nu=3280 \text{ m/s}$ , and  $Z=9 \text{ MRayls}$ .

Figure 4 shows an actual photograph of a modified 2-2 composite array before polishing to final thickness. The cuts ( $p$ ) perpendicular to the majority diced cuts are intended to increase flexibility in both x and y directions as well as strength of the entire structure, allowing the Eccogel polymer to be connected in three dimensions. These modified 2-2 structures can be denominated as 2-(1-3)-2 piezocomposites.

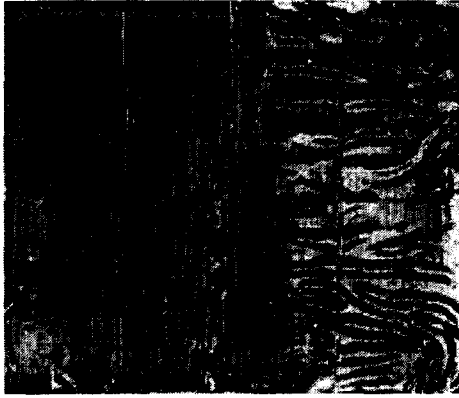


Figure 4. Photograph of a 2-(1-3)-2 flexible array transducer before polishing to the desired thickness.

#### SUMMARY AND CONCLUSION

The processing, fabrication and the electromechanical and acoustical properties of large-area, flexible piezocomposite transducers were presented. These arrays have application in medical diagnostic and therapeutic ultrasound. Flexibility of the array and its individual elements can be obtained by the proper selection of the polymer, such as Eccogel-65, and the specific element design configuration. The latter can be realized with a 2-(1-3)-2 diced structure or a spiral design of the fiber-fabric.

#### ACKNOWLEDGEMENTS

This research was supported by ONR, Exogen Inc., and ACI. The authors would also like to thank the undergraduate students in the Electroceramic Group at Rutgers University; K. Hill, S. Lee, A. Schwager, and M. Wotton.

#### REFERENCES

1. J. Wolff, The Law of Bone Remodeling, 1892. Eng. Translation, Springer Verlag, Berlin, 1986.
2. M. Dyson, J.B. Pond, J. Joseph, R. Warwick, "Stimulation of tissue regeneration by pulsed plane-wave ultrasound," *IEEE Trans. Sonics and Ultrasonics*, SU-17 (3), (1970): 133-139.
3. M. Dyson, C. Franks, J. Suckling, "Stimulation of healing of varicose ulcers by ultrasound," *Ultrasonics*, (1976): 232-236.
4. M. Dyson, M. Brookes, "Stimulation of bone repair by ultrasound," *Proc. Third Meeting of the World Federation in Medicine and Biology*, Brighton, England, July 1982. Pergamon Press, (1983): 61-66.
5. L.R. Duarte, "The simulation of bone growth by ultrasound," *Arch. Orthop. Trauma Surg.*, 101, (1983): 153-159.

6. A.A. Pilla, M. Figueiredo, P. Naser, "Non-invasive low intensity ultrasound accelerates bone repair: rabbit fibula model and human colles' and tibial fractures," *IEEE-EMBS Meeting*, November 1989.
7. J.D. Heckman, J.P. Ryaby, J. McCabe, J.J. Frey, R.F. Kilcoyne, "Acceleration of tibial fracture healing by non-invasive low intensity pulsed ultrasound," *J. Bone and Joint Surg.*, 76-A (1), (1994): 26-34.
8. T.K. Kristiansen, J.P. Ryaby, J. McCabe, J.J. Frey, L.R. Roe, "Accelerated healing of distal radius fractures using specific, low-intensity ultrasound," *J. Bone and Joint Surg.*, 79-A (7), (1997): 961-973.
9. B. Jaffe, W.R. Cook, and H. Jaffe, Piezoelectric Ceramics, Academic Press, London, 1971.
10. A. Safari, "Development of piezoelectric composites for transducers," *J. Phys. III*, 4[7], (1994): 1129-49.
11. T.R. Gururaja, "Piezoelectric transducers for medical ultrasonic imaging," *Am. Ceram. Soc. Bull.* 73[5], (1994): 50-55.
12. L.F. Brown, "Ferroelectric polymers: Current and future ultrasonic applications," in *Proc. IEEE Ultrasonics Symposium*, (1992): 539-550, Piscataway, NJ, 1992.
13. A.R. Selfridge, "Approximate material properties in isotropic materials," *IEEE Trans. Sonics & Ultrasonics*, SU-32[3], (1985): 381-394.
14. D. Waller, A. Safari, "Corona Poling of PZT Ceramics and Flexible Piezoelectric Composites," *Ferroelectrics*, 87 189-195 (1990).
15. R. Cass, "Fabrication of Continuous Ceramic Fiber by the Viscous Suspension Spinning Process," *Am. Ceram. Bulletin*, 70 [3], (1991): 424-429.
16. S. Mark Herman, "*Encyclopedia of Polymer Science and technology*," John Wiley & Sons, New York, 1969, pp. 810-844.
17. W. D. Bascom, "Fiber Sizing," pp. 122-124, *Engineering Materials Handbook*, Vol. 1, *Composite*, Edited by Cyril A. Dostal, ASM International Handbook, Metals Park, 1987.
18. J. Souquet, Ph. Defranould, and J. Desbois, "Design of low-loss wide-band ultrasonic transducer for noninvasive medical application," *IEEE Trans. Sonics & Ultrasonics*, SU-26[2], (1979): 75-81.
19. M.G. Grewe, "Acoustic matching and backing layers for medical ultrasonic transducers," M.Sc. Thesis, Pennsylvania State University, 1989.