PIEZOELECTRIC DEVICES

Two kinds of piezoelectric devices will be considered in this article: piezoelectric bimorphs and ultrasonic transducers for medical applications.

PIEZOELECTRIC BIMORPHS

A piezoelectric bimorph is a laminate of two piezoelectric thin, flat elements, that are glued together over their entire large surfaces. Electrodes are provided on both elements so that electric fields can be applied to them. The polarization of the elements and the electrode configuration is such that when a voltage is applied to the electrodes, one element will expand and the other will contract. This will cause the elements to exert forces and moments to each other, which will bend the construction.

They are an invention of C. Baldwin Sawyer (1). In this paper, Sawyer gives a brief historical account of the development of piezoelectric materials for acoustic uses. He describes methods for the inexpensive commercial production of Rochelle salt which was the most important piezoelectric material in those days.

He states "The underlying principle of the special assemblies is that of mutual opposition with resultant magnification of motion." In this respect, piezoelectric bimorphs are completely analogous to bimetallic strips, two sheets of metals with different thermal expansion coefficients, attached to each other with their large surfaces. These strips will bend when the ambient temperature changes because of differences of expansion.

Microphones, pick-up elements, and speakers were the most important devices then. They are now used in a large variety of applications, such as beepers in watches and microwave ovens and as the microphone and speaker elements in telephone handsets.

Bimorphs are made mainly in rectangular or circular shapes, the latter forms are mostly applied as the microphone or speaker elements in computers, but they are also used as pumping elements in medical drug delivery systems. The former are used as display elements and optical choppers, as cooling fans, as the element that deflects the laser beam in certain bar code reading devices in supermarkets or drug stores, and as many other applications where a modest but easily controllable displacement or rotation is required.

Three Kinds of Bimorphs

Three construction methods are used to make piezoelectric bimorphs. Figure 1(a) shows how two thin piezoelectric elements are covered with electrodes and how they are polarized. The upper element has its upper electrode connected to the positive side of the voltage source, whereas its under electrode is connected to the negative side of the voltage source; therefore, the electric field E is pointed downward. The dielectric polarization vector, which points from the negative charge of the elementary dipoles to the positive charge, is downward oriented, indicating that the positive charge of the elementary dipoles is under the negative charge in this figure. The electric field E pushes the positive charge farther down and the negative charge farther up, thereby increasing the distance between all the elemental charges that form the elemental dipoles. This means that this piezoelectric element becomes a little thicker, and to maintain its volume, it has to shrink sideways, in both the length and width directions. The underlying piezoelectric element has the same polarization direction, but as the electric field is reversed here, its action has the opposite effect, making the element thinner, longer, and wider. These joint effects result in bending, as indicated in Fig. 1.

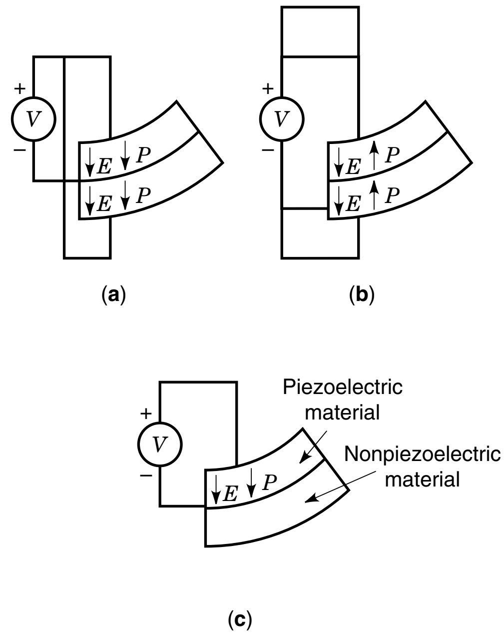

Figure 1. The connections and constructions of three different kinds of piezoelectric bimorphs.

Figure 1(b) shows the second construction; here the center electrode is omitted, and the electric field is present between the top and bottom electrode. The polarization of the lower piezoelectric elements has been reversed so that still one polarization vector is opposing E whereas the other is parallel with it. This construction is simpler because the center electrode does not have to be fabricated, so it also does not have to interfere with the gluing of the bimorphs, but the electric field for the same element thickness and the same voltage from the voltage source is only half that of the bimorph in Fig. 1(a) which results in half the deflection.

The third constuction is shown in Fig. 1(c). Here, a piezoelectric element is glued to a nonpiezoelectric element. The piezoelectric element will still respond to the applied electric field by expanding or contracting according to the voltage, whereas the nonpiezoelectric element will resist expansion or contraction. For this reason, a counteracting stress will build up in the lower element, which works together with the stress in the top element to form a moment that bends the bimorph. This construction is often referred to as a unimorph or a monomorph to distinguish it from the other two constructions in which two piezoelectric elements are used. This construction is often cheaper or more practical, but because there is only one active piezoelectric element, it also has less bending.

Constitutive Equations

Because a bimorph can deflect under the influence of a voltage, it can exert a force. The deflection is associated with a rotation of the tip, which implies that it can also exert a moment. To present the relation between forces, moments, and voltage, we refer to Fig. 2. In which a bimorph is shown of the second type, as in Fig. 1(b). Consider that the elastic modulus of the piezoelectric material is s_{11}^E, whereas the piezoelectric constant of importance here is d_{31}, and the dielectric constant is ϵ_{33}^T. The length of the bimorph is L, the width is w, and the height of each element is h. A voltage V is applied to the electrodes, whereas a force F acts perpendicular to the plane of the bimorph at the tip. A moment M bends the bimorph as indicated. The tip deflection is called δ, and the tip rotation is α. The charge on the electrodes is Q. For completeness a pressure p is considered to act uniformly on the bimorph, which results in a displaced volume ν. Now the following matrix relations holds:

$$\begin{pmatrix} \alpha \\ \delta \\ \nu \\ Q \end{pmatrix} = \begin{pmatrix} \dfrac{3s_{11}^E L}{2wh^3} & \dfrac{3s_{11}^E L^2}{4wh^3} & \dfrac{s_{11}^E L^3}{4h^3} & \dfrac{-3d_{31}L}{4h^2} \\ \dfrac{3s_{11}^E L^2}{4wh^3} & \dfrac{s_{11}^E L^3}{2wh^3} & \dfrac{3s_{11}^E L^4}{16h^3} & \dfrac{-3d_{31}L^2}{8h^2} \\ \dfrac{s_{11}^E L^3}{4h^3} & \dfrac{3s_{11}^E L^4}{16h^3} & \dfrac{3ws_{11}^E L^5}{40h^3} & \dfrac{-d_{31}wL^3}{8h^2} \\ \dfrac{-3d_{31}L}{4h^2} & \dfrac{-3d_{31}L^2}{8h^2} & \dfrac{-d_{31}wL^3}{8h^2} & \dfrac{\epsilon_{33}^T Lw}{2h}(1-k_{31}^2/4) \end{pmatrix} \begin{pmatrix} M \\ F \\ p \\ V \end{pmatrix}$$

The blocking force F_b can now be derived as the force the bimorph will exert when it is restrained from deflection as $F_b = 3d_{31}\mathrm{whV}/4s_{11}^E L$. A full account of the constitutive equations can be found in Ref. 2.

ULTRASONIC TRANSDUCERS FOR MEDICAL APPLICATIONS

Medical uses of piezoelectric transducers include imaging, surgical devices, and hyperthermia applicators; the largest of these applications is imaging. Ultrasonic transducers for medical imaging are used with ultrasonic imaging systems to transmit short pulses of ultrasound into the body and receive the echoes from various anatomical structures. These echoes are processed by the imaging system to produce a picture of a two-dimensional slice in the region of interest. In some cases, multiple slices may be combined to form a three-dimensional representation.

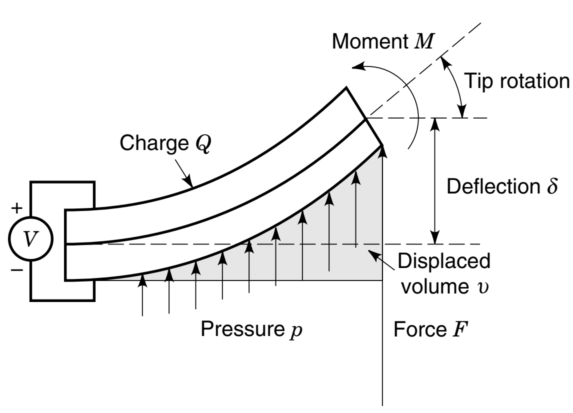

Figure 2. The definitions of the four external parameters M, F, p, and V and the four internal parameters α, δ, ν, and Q.

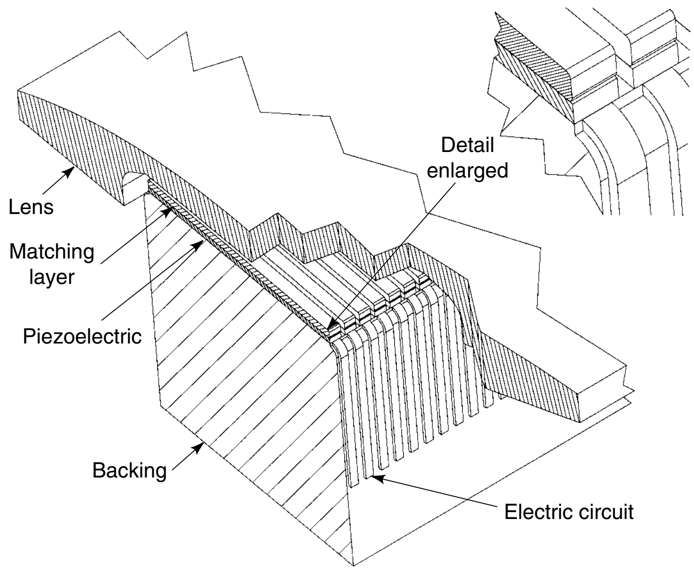

Figure 3. Medical ultrasound transducer.

Most medical ultrasonic imaging is done in the frequency range of 2 MHz to 12 MHz, and specialized transducers are built up to about 50 MHz for imaging of the eyes and skin or for mounting on a catheter to image the interior of blood vessels. Experimental imaging transducers have been built with operating frequencies in excess of 100 MHz. The earliest imaging transducers consisted of a single piezoelectric element that was scanned mechanically to cover the region of interest. These transducers are still common in the low-cost market and for high-frequency applications. The overwhelming majority of medical imaging transducers are electronically scanned arrays, in which the aperture is broken up into numerous small elements, each a miniature independent transducer. Scanning the beam is performed by varying the timing of the signals to the different elements so that the resultant phase front is focused on the point of interest. A cross section of a typical electronically scanned array is shown in the figures. It consists of a long row of individual elements attached to a common backing and covered by a cylindrical acoustic lens. The size of the transducer is determined by the desired beam profile and varies by application and frequency. Large apertures are usually desired because they produce more compact beams while small apertures give a greater range of focus and better anatomical access. A transducer for cardiac imaging might operate at 2.5 MHz and have an active aperture of 12 mm by 20 mm. This transducer would have 64 elements, each 12 mm by 0.28 mm with a 0.04 mm space between elements. Other transducers could have a few hundred elements in a longer row or arranged in multiple rows to provide additional electronic focusing. The elements consist of a layer of piezoelectric material sitting directly on top of the backing, and one or more acoustic matching layers between the piezoelectric layer and the lens. Electrodes are applied to the top and bottom surfaces of the piezoelectric. A common ground connection may be made, but each element requires its own signal lead. The layers are bonded together, typically with a thin layer of adhesive, to ensure good acoustic contact. The elements are separated by saw cuts to provide acoustic isolation. Electric connection to the imaging system is made through a cable, typically 2 m in length, having an individual miniature coaxial cable for each element.

The number of elements in the array is determined by the aperture size, the operating frequency, and the amount of beam steering required. In the same manner as a diffraction grating, these arrays of elements will produce beams in several directions unless the elements are very closely spaced. For large steering angles, the element pitch must be no greater that about half a wavelength of the operating frequency, or first-order grating lobes will cause image artifacts. A coarser pitch may be used at some sacrifice in steerability and image quality, but the pitch usually is not greater than 1.5 wavelengths. These limits also roughly coincide with the small element size dictated by diffraction theory to allow transmission and reception over the desired range of steering angles.

For the best spatial resolution in the image, it is important that the transducer have a short impulse response and, therefore, a wide bandwidth. A 40% bandwidth about the operating frequency is a minimum, and 80% to 90% bandwidth transducers are becoming common. The depth of penetration of the image is limited chiefly by acoustic attenuation in the body, so it is important that the transducer itself have as low a loss as possible. These two requirements are best met with piezoelectric materials having a high electromechanical coupling constant and a high capacitance. This has led to the near universal use of the lead zirconate titanate (PZT) family of ceramics for array transducers either as a solid ceramic or as a composite material consisting of vertical slats or posts of PZT in a polymer matrix.

The piezoelectric layer is polarized in the vertical direction so that, when an electrical signal is applied, the element vibrates in a bulk thickness mode. This is a longitudinal wave plate mode modified to account for the narrow width of the element and the stress-free boundary conditions on the sides of the element. The maximum coupling occurs near the first-order thickness resonance. The width of the element is usually somewhat less than the thickness of the piezoelectric layer so that the lateral resonance frequency will be just above the thickness resonance. Elements that are too wide may cause interference with the imaging spectrum. In such cases, it is necessary to divide the elements into subelements with additional saw cuts to move the lateral resonances away from the imaging band.

The piezoelectric element is schematically represented as having one electric port and two acoustic ports. Achieving low loss and wide bandwidth requires the efficient coupling of energy through both the electric port and the front acoustic port. The rear acoustic port is loaded with a dissipative backing material to help damp the resonance.

The front acoustic port impedance match is accomplished by using acoustic matching layers. These act as acoustic transmission lines and are about a quarter wavelength thick, forming a transmission line transformer completely analogous to the quarter wave transformers used in microwave circuits. The characteristic acoustic impedance of the modified longitudinal mode in PZT is about 30 MRayls, but because of the loads on the other two ports, the impedance seen looking into the front acoustic port of the piezoelectric will vary considerably from this and have a large reactive part. The acoustic impedance of the patient's body is about 1.5 MRayls, giving a large impedance mismatch. This mismatch may be considerably reduced by the use of composite piezoelectric materials as described earlier, but at the expense of much lower element capacitance and consequently higher electric impedance. The materials used for the matching layers must have not only the correct acoustic impedances but also mechanical and chemical characteristics that allow a robust manufacturing process. The most common matching layer materials are plastics and epoxies, which may be loaded with other materials to increase their acoustic impedance.

The rear acoustic port is terminated with a backing material having an impedance less than the piezoelectric. This backing serves to further damp the resonator and acts as a mechanical support for the elements. The impedance of the backing is selected to couple sufficient energy from the piezoelectric to damp the resonance, but not so much as to cause excessive signal loss. Additionally, the backing must have a high acoustic loss so that the energy it absorbs does not reflect off the boundaries and re-enter the piezoelectric causing image artifacts.

The electric impedance of the element is determined by the properties of the piezoelectric material and by the acoustic impedance match. A large electromechanical coupling constant is required for wide bandwidth. For very small elements, a high dielectric constant is desired to keep the electrical impedance low, usually necessitating the use of a solid ceramic. For larger elements, the increased area results in a higher capacitance that may permit the use of a composite material with its desirable acoustic properties. The acoustic match is reflected as a radiation resistance at the electric port, so the better the acoustic match, the less reactive is the electrical impedance. The electric port of each element is connected to a cable, typically about 2 m in length, to allow the operator to position the transducer as required for imaging. The cables are terminated in a connector that allows easy changing of transducers on the ultrasound system. The connector usually also contains tuning networks to provide electric impedance matching. Often a single series inductor is used to tune out the combined capacitance of the transducer element and the cable.

There are many variations on the transducer architecture presented here. The size, shape, and number of elements vary considerably according to application. Different numbers of matching layers may be used or the piezoelectric itself may be a multilayer structure. Electrostrictive materials have been used in place of piezoelectrics to take advantage of their electrically modifiable properties. In place of a lens, the elements themselves may be given a cylindrical curvature. Numerous variations exist in manufacturing, as in the bonding together of the various layers, sawing of the elements, and attachment of electric leads. Mechanically scanned transducers also have a similar structure but consist of just a single large element that is wobbled or rotated mechanically to scan the desired field of view.

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