A Non-Invasive Technique for Detecting Stress Waves in Bone Using the Piezoelectric Effect

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Abstract—A stress wave propagating in a long bone is accompanied by a traveling electrical charge generated as a result of the piezoelectric character of bone. An electromagnetic device has been developed which can detect the magnetic fields associated with this charge and which is capable of monitoring stress waves in bone *in vivo*. The field measurement is independent of the mechanical properties of the soft tissue overlying the bone, so that difficulties previously encountered with vibration and wave propagation tests to determine *in vivo* properties of bone are avoided. Applications to the diagnosis of bone disorders are discussed.

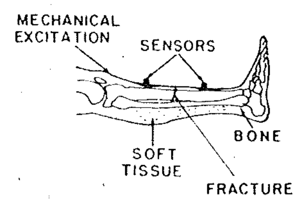

Fig. 1. Generalized wave propagation scheme for assessing fracture healing.

I. Introduction

Elastic wave propagation techniques have been a subject of considerable investigation as experimental methods for the diagnosis of bone disorders. The objective of such experimentation is to non-invasively determine the *in vivo* mechanical properties of bone tissue. The development of a viable technique of this type would provide information currently unavailable from radiography and would be of use in the diagnosis of osteoporosis [1], [2] and of fracture healing. As early as 1958, Anast et al. [3] propagated ultrasonic waves across fracture sites in living subjects; however the lack of statistical significance of this work has been criticized [4]. Experiments in which physical properties of bone have been successfully correlated with wave propagation parameters [5], [6], [7], [8] have generally involved excised specimens and are therefore not directly applicable to clinical situations. More recently, instrumented hammers and accelerometers have been used to determine the propagation speed of stress pulses in bone *in vivo* [9]. However no correlation with disease states was obtained. The delay of an ultrasonic wave across a fracture site has been found to be too small to cause a measurable change in the average propagation velocity, particularly when the delays associated with soft tissue *in vivo* are considered [10]. Amplitude attenuation appears to be a more suitable measure of the degree of union. Other authors have attempted to use impedance tests to determine the *in vivo* mechanical properties of bone [11].

A major drawback inherent in these approaches is the soft tissue through which the pulses or ultrasonic waves must be propagated to excite the bone and to be detected [Fig. 1]. The variations in the quantity and quality of soft tissue from patient to patient constitute a complicating variable which cannot be easily evaluated [12]. The present research is aimed at developing a technique for the detection of stress waves in living bone, which is independent of the mechanical properties of the soft tissue and therefore is not subject to these limitations. The technique is based on a measurement of the magnetic field generated by a stress wave as a result of the piezoelectric effect in bone. The measurement of artificially induced magnetic fields used here is to be distinguished from techniques such as magnetocardiography [13], [14] in which naturally occurring magnetic fields resulting from currents in the heart, are detected.

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II. The Piezoelectric Effect

Piezoelectric response in a material is said to occur if the electric displacement vector D_i depends not only on the electric field E_j but also on the stress tensor σ_{jk}. The constitutive equation for linear piezoelectricity with stress and electric field as independent variables is:

$$D_i = d_{ijk}\,\sigma_{jk} + K_{ij}E_j$$

where K_{ij} is the dielectric permittivity tensor and d_{ijk} is the piezoelectric tensor. In general these coupling coefficients will depend on frequency as a result of relaxation effects, so that K and d must be regarded as complex quantities.

A number of authors have suggested alternative nomenclature to describe the electromechanical behavior of bone, in view of the uncertainty regarding the mechanisms responsible for such behavior, and in view of recently observed deviations from classical piezoelectric response [15]. Nevertheless, in the interest of brevity, in the present work we shall use the term "piezoelectric" to describe stress or strain related polarizations in bone. The existence of piezoelectric behavior in dry bone was reported by Fukada and Yasuda [16] in 1957; more recently the effect has also been observed in wet bone [17], [18], [19].

The magnitude of the d coefficients, which are a measure of the strength of piezoelectric coupling, is known to depend on temperature, frequency, and relative humidity for compact bone. The largest piezoelectric coefficients for bone are those associated with shearing deformations. For compact bone at body temperature and 97% relative humidity, Re $[d_{123}]$ is of

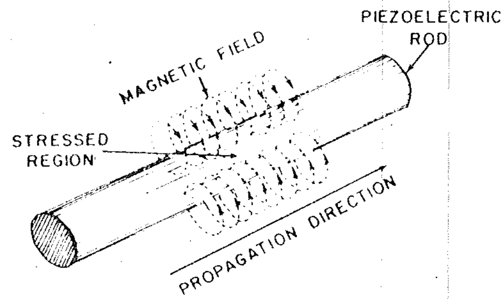

Fig. 2. Magnetic field resulting from a stress pulse in a piezoelectric rod.

the order 10^{-13} C/Nt between 1 and 100 Hz [18] and attains a relative minimum in this domain. It should be pointed out that piezoelectric response in bone is relatively weak; for comparison, d_{223} for quartz is 4.6×10^{-12} C/Nt while for a lead titanate zirconate ceramic, d_{333} is 5.9×10^{-10} C/Nt. To the authors' knowledge, no data are currently available for the piezoelectric d coefficients of bone at frequencies which are relevant to wave propagation studies. The only available datum at high frequency is the value of d_{14} at 5,600 Hz which varied from 5.5×10^{9} ESU for dry bone, down to 3×10^{-9} ESU at 100% relative humidity [19]. But even this frequency is about one tenth of that utilized in this study.

Piezoelectric effects in bone are thought to constitute the mechanisms for remodeling of bone according to Wolff's law and have been shown to vary as a function of at least one disease state (lathyrism) [20]. Further information regarding the phenomenology of and the mechanisms for bone piezoelectricity may be found in a review article by Williams [15].

Elastic waves in piezoelectric media are in general accompanied by both electric and magnetic fields [21] (Fig. 2). This phenomenon forms the basis of the present technique for remotely detecting these waves in bone. The electric field could be detected by means of electrodes placed on the skin. Such a measurement is affected to first order by variations in the complex dielectric coefficients of the surrounding soft tissue. By contrast, magnetic fields, in the frequency range for which the effect of soft tissue conductivity is small, are not perturbed significantly by variations in the soft tissue properties. Therefore the present technique is based on the detection of the magnetic fields associated with a stress wave in bone.

III. Materials and Methods

A time-varying magnetic field may be detected by observing the voltage induced in a coil of wire wrapped about a core of magnetic material (e.g. ferrite) embedded in the field. This type of sensor is limited in useful sensitivity by the following phenomena:

(i) Johnson noise is associated with the resistance of the windings, and this noise masks weak signals.

(ii) Interfering magnetic fields from external sources are detected by the sensor and also will mask weak signals from the bone.

(iii) Distributed capacitance [22] between windings and the so-called leakage inductance form a resonant circuit.

(iv) The inductance of the sensor and the input capacitance of the preamplifier to which it is connected also constitute a resonant circuit. These resonances limit the portion of the frequency domain in which the sensor can be used to detect pulsed waveforms and in effect place an upper bound on the usable sensitivity.

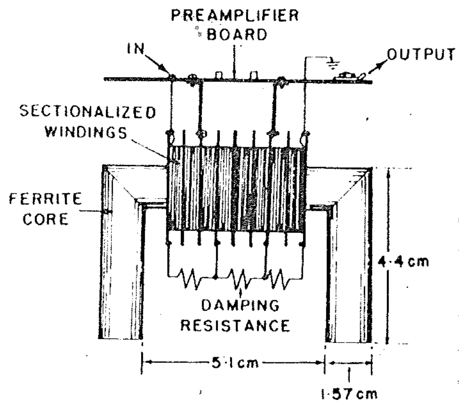

Fig. 3. A magnetic sensor.

Inter-winding capacitance depends upon winding geometry; this capacitance may be reduced by a factor $1/n^2$ by means of sectionalizing the coil into n segments [22]. Now in dealing with (iv) it must be recognized that the sensor inductance can be reduced only by reducing either the number of turns or the core permeability, which would increase the bandwidth only at the expense of reduced sensitivity. The reduction of load capacitance, on the other hand, is a feasible approach since preamplifiers can be obtained with input capacitance as low as 3 or 4 pF. Since typical coaxial cable has a capacitance of approximately 30 pF/ft, the preamplifier must be mounted directly atop the sensor to minimize the total capacitive load presented to the sensor and therefore maximize its resonant frequency.

A magnetic sensor designed with these points in mind is depicted in Fig. 3. Damping resistors R_D are included to suppress the sharp LC resonances which would otherwise introduce excessive ringing into the pulse response of the sensor. The ferrite material used had an initial permeability of about 2200. Windings were divided into nine segments, each containing 500 turns of wire. The self-inductance of the winding was determined by shunting the sensor with a large capacitor and measuring the resonant frequency. Damping resistors were omitted for this procedure. For $C_{\text{shunt}} \geqslant 80$ pF, the inductance determined in this way was independent of capacitance and was equal to 3 H. For small values of shunting capacitance, the sensor did not behave like a pure inductance; the residual capacitance in the system was 10 pF of which 3 pF was associated with the input capacitance of the preamplifier.

Feedback can be employed in the preamplifier circuit to simulate a negative input capacitance so that a portion of the source capacitance of the sensor as well as the preamplifier capacitance is neutralized [Fig. 4a]. This circuit tends to become unstable if the source impedance is large, as it is at high frequencies for the highly inductive magnetic sensor. This instability imposes limits on the maximum neutralizing capacitance C_N which can be used. The modified circuit in Fig. 4b avoids this difficulty by reducing the feedback at high frequencies.

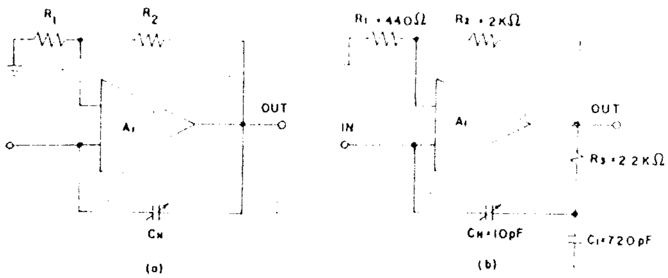

Fig. 4. (a) Negative input capacitance circuit. (b) Capacitance neutralizing preamplifier with modified feedback.

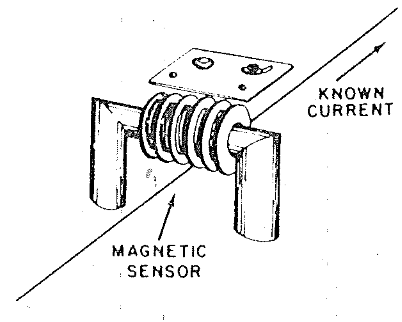

Fig. 5. Test configuration for magnetic sensor.

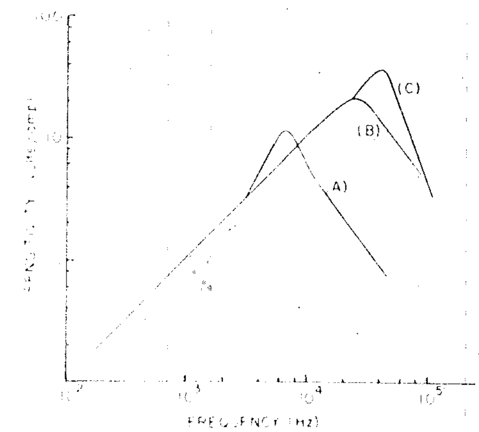

Fig. 6. Magnetic sensor response. *A*. Into oscilloscope, 55 pF//1 MΩ through 124 cm cable. *B*. Into preamplifier mounted above sensor; $C_N = 0$. *C*. Into preamplifier; $C_N \neq 0$.

The sensitivity, i.e. the voltage response to a sinusoidal current input, divided by the current amplitude, of different types of magnetic sensor was determined in a standard geometry using the testing arrangement diagrammed in Fig. 5. For the sensor described above, the measured sensitivity is plotted as a function of frequency in Fig. 6.

To generate the stress pulses in bone, a transducer containing a stack of lead-titanate-zirconate [LTZ] piezoelectric ceramic discs (Transducer Products, Inc.) was manufactured. With suitable excitation, this driver was capable of generating single pulses 10 μs wide in a thin plexiglas rod. Surface strain in the specimens of plexiglas, aluminum, and excised bone was determined directly by means of semiconductor strain gages cemented with a cyanoacrylate adhesive [Eastman 910] and excited with 6.3 V dc. Observations were carried out using the experimental configuration shown in Fig. 7. The oscilloscope used was a Tektronix type 5030 dual beam.

Despite the relatively low sensitivity of the sensor to magnetic fields of low frequency, output resulting from 60 cycle "hum" pickup proved to be excessive [approx. 1 mV p-p]. The use of an active filter with a rejection of 4×10^4 at 60 Hz and constant gain between 5 kHz and 50 kHz was found to be sufficient to eliminate this signal. The pickup of magnetic interference from the pulse generator (Hewlett-Packard, model 212 A) proved to be a more serious problem since these signals were synchronous with the output pulses and overlapped the signals of interest in the frequency domain, so that they could not be eliminated by filtering. This interference was reduced in amplitude by moving the pulse generator across the room, pending the installation of a pulser designed to minimize external fields.

IV. Results

Three specimens of air-dried human embalmed femoral bone were examined using the magnetic sensor apparatus. Typical strain-gage and magnetic sensor outputs are shown in Fig. 8. This photograph contains a superposition of the responses to the stress waves associated with about fifty pulses. A time interval of 2500 μs between pulses was provided so that reflected waves would have time to be damped out before the arrival of the following pulse. Figure 8 clearly shows qualitative agreement between the magnetic sensor and strain gage outputs except that during the initial 80 μs period the strain gage output is approximately zero while the magnetic sensor trace contains a peak. The delay in the appearance of a strain signal following excitation of the driver is due to the finite propagation speed of bending waves in bone while the initial peak in the magnetic sensor output is a noise signal resulting from electrical activity within the pulse generator. Note that the amplitude of the maximum strain at the bone surface caused by the transducer is of the order 0.6×10^{-6}, or less than 1/300 of the strains associated with normal walking in animals [23], [24].

Owing to the detector's sensitivity to extraneous magnetic fields, the following tests for artifacts were performed:

(i) The driver was energized but not pressed to the bone, thus generating no stress waves in the bone. The photograph in Fig. 9 shows that the interference from the pulser during the first 80 μs remains; however neither the strain gage nor the magnetic sensor exhibit any response above noise following this.

(ii) The driver was pressed to the bone, generating stress waves resulting in strain gage output as before (the bottom trace in Fig. 8). When the magnetic sensor was moved 1 m from the bone it responded as in the top trace in Fig. 9, that is, no response above noise.

(iii) Stress waves were generated in strain gaged rods of aluminum and plexiglas using the same procedure as was used for bone. The magnetic sensor again behaved as in the top trace in Fig. 9, showing no response to the presence of stress waves in these non-piezoelectric materials.

(iv) To examine the possibility that the sensor-preamplifier assembly was responding to electric fields, a resistance equivalent to that of the magnetic sensor with damping resistors, was soldered to the preamplifier input. The sensor itself was left in place but disconnected electrically. The experimental arrangement in Fig. 7 was otherwise unchanged. The resulting output exhibited neither the initial "pickup" nor any wave-

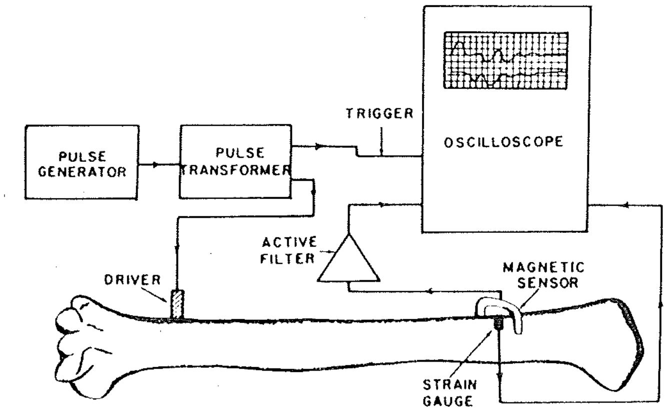

Fig. 7. Apparatus for the magnetic detection of stress waves in bone.

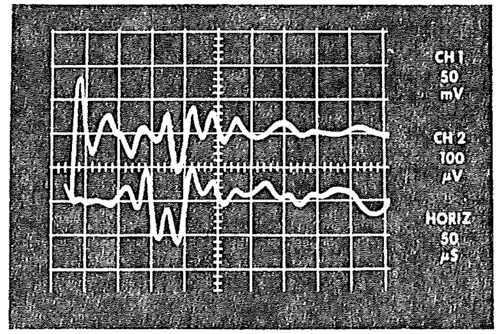

Fig. 8. Waveforms associated with stress waves in bone. *Top:* Magnetic sensor output, vertical scale 500 μV/division, referred to sensor output. *Bottom:* Strain gage output, vertical scale 100 μV/division, or 0.6×10^{-6}/division strain. Horizontal scale 50 μs/division.

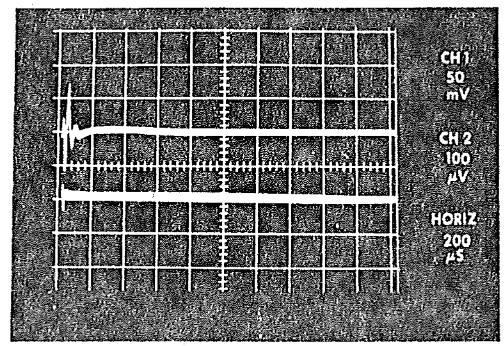

Fig. 9. Waveforms associated with magnetic and electric interference. Driver not pressed to bone. *Top:* Magnetic sensor output, vertical scale 50 μV/division, referred to sensor output. *Bottom:* Strain gage output, vertical scale 100 μV/division, or 0.6×10^{-6}/division strain. Horizontal scale, 200 μs/division.

form associated with strain in the bone above the noise level shown in Fig. 9.

(v) Removal of the dc excitation voltage from the strain gage had no effect on the magnetic sensor output. This shows that currents in the strain gage circuitry are not responsible for the signals seen in the magnetic sensor output.

The results of these tests indicate that the observed magnetic sensor signals result from magnetic fields created by the stress waves via the piezoelectric effect, rather than from any artifacts. The close correspondence between the magnetic sensor and the strain gage outputs indicates that this sensor could be used to detect stress waves in long bones.

Magnetic signals associated with elastic waves were also observed in wet, fresh, excised bone, as well as in the living bones of human volunteers. These signals were weaker than those obtained from dry bone; therefore a signal averager [Princeton Applied Research, model TDH-9] was used to extract them from the noise. Waves of larger amplitude could be used in living subjects to eliminate the need for an averager. These in-vivo tests are being continued and the results will be reported at a later date.

Discussion and Conclusions

An instrument has been constructed which is capable of remotely detecting elastic waves in bone by sensing the piezoelectrically generated magnetic field. It has been observed that pulsed waves in dry bone of a strain amplitude less than 0.3% of the strains encountered in walking produce magnetic fields which are strong enough to detect. A variety of tests have been performed in order to rule out possible artifactual causes for the observed signals.

The objective of the various stress wave propagation techniques (including ultrasonics) for the diagnosis of bone disorders is to obtain objective information concerning the quality and structural integrity of the bone in question which can complement the subjective information presently available using radiography. In the application of such techniques, elastic waves can be readily excited in bones by means of a transducer pressed to the skin over the bone [6, 10]. The presence of soft tissue over the bone, however, introduces errors in measurements of the wave amplitude and velocity when these measurements are performed using mechanical sensors placed on the skin. The present technique does not entail direct mechanical measurements and therefore is not affected by errors related to the mechanical compliance of the soft tissue.

To apply the present technique to the evaluation of fracture healing, an experimental configuration resembling that shown in Figure 1 might be used. Magnetic detectors not in contact with the skin would be used as sensors. A comparison of the outputs of the two sensors would then provide a measure of the transmission coefficient of elastic waves through the fracture site. A preliminary study on simulated healing bones indicates that the transmission coefficient is correlated to the degree of union [25, 26]. Therefore by monitoring the change in transmission coefficient of a healing bone as a function of time, one may then be able to evaluate the rate of fracture healing.

Alternatively, the degree of union could also be estimated by comparing the transmission coefficient of the fractured limb with that of the contralateral limb. However, the variations in the piezoelectric coefficients among the bones of the paired limbs may affect the result of such comparisons. An upper bound in the variations in piezoelectric coefficients of bovine bone samples is of the order of 15%-20% [18]. The present authors are not aware of any measurements of the variability of the piezoelectric coefficients within a single bone or between the bones of paired limbs; therefore an estimate of possible errors due to such variability cannot be made at the present time.

With regard to osteoporosis, the changes in porosity, cortical thickness, and bone mass which occur in this disorder are suggestive of changes in the wave-propagation characteristics of the bone. This observation has led some authors to propose the use of acoustic methods as a diagnostic tool [6]. Unfortunately, the preliminary *in vivo* experimentation correlating various wave-propagation parameters (e.g. wave speed, attenuation, normal mode structure) with the state of osteoporosis of a bone has not yet been done. If significant correlations are found, then the present technique could be used in a configuration similar to that shown in Figure 1 to eliminate the effect of soft-tissue compliance as a confounding variable. Again, variations in piezoelectric moduli from individual to individual are not expected to significantly influence results based on a comparison of the outputs of two sensors; it is only necessary that these moduli remain reasonably constant over the length of the bone.

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