A noncontacting electromagnetic device for the determination of in vivo properties of bone

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Lakes, R. S., and S. Saha. A noncontacting electromagnetic device for the determination of in vivo properties of bone. *Med. Instrum.* 12: 106–109, 1978.—A noncontacting, electromagnetic device to monitor stress waves in bone has been developed. Since bone exhibits piezoelectric-like behavior, a propagating stress wave in bone generates both electric and magnetic fields. The present device detects the magnetic field and thus provides a measure of the stress-wave amplitude in the bone. Excised dry and wet human femora, as well as a model system consisting of a bar of strongly piezoelectric ceramic, were examined using the device. Magnetic signals associated with elastic waves were also recorded in the long bones of human volunteers. Since the observed signals are not perturbed by the mechanical quality of the soft tissue over the bone, the device may be used to quantitatively assess the mechanical properties of bone in vivo.

bone properties; electromagnetic sensor; magnetic signals; stress waves

Introduction

AT PRESENT, ULTRASONIC TECHNIQUES are perhaps the most widely used methods for nondestructive evaluation of materials (24). In such applications, the structures are generally of regular geometry, and transducers can be affixed directly to the specimen. This allows a relatively straightforward measurement of the ultrasonic transmission and reflection coefficients. These wave propagation parameters can then be correlated to the size and nature of defects in the material.

Ultrasonic techniques and vibration tests (as used to evaluate engineering materials) have also been tried as a noninvasive means for the determination of the mechanical integrity of long bones and for the assessment of the rate of fracture healing (1,2,16,21,23). However, in these methods, the response of the bone is measured by a transducer pressed on the soft tissue over the bone, and the results are thus dependent on the thickness of the tissue and on the magnitude of force pressing the transducer to the skin (6,18). In an attempt to develop methods that avoid such difficulties, several scientists have proposed methods based on measurement of the motion of metal objects in contact with the bone. For example, traction pins used in the management of some fractures will vibrate in response to a stress wave induced in the bone (19). This vibration can be measured independently of soft tissue effects; however, this approach has potential utility only for patients treated with traction pins. It is also possible to excite and detect stress waves in a bone by means of needles passed through the skin and embedded in the bone (22). Although such a method can, in principle, avoid errors related to soft tissue, the invasive character of this technique is a significant drawback in the clinical setting.

At present, roentgenographic examination is used clinically to determine the mechanical integrity of bones and to assess the progress of fracture healing. Some bone disorders, however, are difficult to diagnose accurately by radiographic techniques alone. For example, in osteoporosis, approximately 30 percent bone loss must occur before the loss is noticeable on an X-ray film (12,13). Furthermore, it is sometimes difficult in fracture cases to distinguish between malunion and bony union. Instances have been reported in the literature in which a fractured bone appeared completely healed in an X-ray examination, yet it refractured during the patient's normal activity (20). By contrast, in other cases immobilization of fracture patients may be unduly prolonged (22).

It is evident from these considerations that there is a need for additional capabilities for a more accurate determination of the characteristics of an intact bone and of the state of union of a healing bone. This paper describes the development of a new electromagnetic device which can, without

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touching the skin, monitor piezoelectrically generated fields associated with a stress wave in a long bone. It is expected that this device can be used to determine the in vivo mechanical properties, and possibly also the piezoelectric properties, of long bones, and that such information can be used to measure the rate of fracture healing.

Materials and Methods

Instrumentation

The operation of the device described in this paper is based on the piezoelectric effect in bone. This effect has been observed in dry bone (9), bone at different hydrations (4,17), wet bone (5,17), and in bone in vivo (5). Piezoelectric materials acquire an electrical polarization when subjected to mechanical stress. This polarization results in an external electric field; if the stress varies with time, a magnetic field is also generated. The present device is designed to detect the magnetic field associated with stress waves in bone. The magnetic field was chosen on the basis of a preliminary theoretical model, which suggested that magnetic fields outside the skin would be only weakly perturbed by changes in the complex dielectric properties of the soft tissue over the bone.[1] Electric fields, in contrast, would depend more strongly on soft-tissue properties.

Sensors were constructed by winding 4500 turns of fine wire (No. 38) around a ferrite core, the permeability of which was about 2200. To minimize capacitive loading, a specially designed preamplifier was mounted directly above the sensor, as shown in Fig. 1. Sensor output was conditioned using an active filter that removed the strong 60-cycle "hum" component of the signal and restricted the bandwidth to 80 kHz. A driver for the excitation of stress waves was made by constructing a stack of lead-titanate-zirconate piezoelectric elements. A pulse amplifier, shown schematically in Fig. 2, was designed to energize the driver and to match its impedance. The input to the amplifier was provided by a pulse generator (type 505, Tektronix, Inc., Beaverton OR). The pulse amplifier was designed to be compact so that the effect of external electromagnetic interference was minimized; shielding produced further reduction of such interference. In preliminary tests, a large tube-type pulse generator (model 212A, Hewlett-Packard Co., Palo Alto CA) had been used; however, this produced excessive interference and insufficient output for in vivo trials.

The longitudinal component of surface strain in the bone specimens was measured using semiconductor strain gauges cemented to the bone with a cyanoacrylate adhesive (Eastman 910, Eastman Chemical Products Inc., Kingsport TN). Noise in the strain gauge output was reduced by means of an active filter, which restricted the output response to a bandwidth of 100 kHz.

Methods

Pulsed bending waves of small amplitude were generated in both dry and wet embalmed, human femora using the configuration shown in Fig. 3. An identical setup was used in tests for artifacts, using 0.5-in.-diameter rods of nonpiezoelectric aluminum and Plexiglas in place of a bone. Electrical pulses supplied to the driver were 10 μsec in duration and were separated by an interval of 10,000 μsec, corresponding to a repetition rate of 100/sec. This interval was sufficient for reflected waves to be damped out before the arrival of the next pulse.

Whole bone has geometric and structural complexities that cause considerable alteration of incident waveforms as they propagate. Therefore, a setup similar to that shown in Fig. 3 was used to examine a simplified model system consisting of a solid bar of piezoelectric ceramic. Longitudinal stress waves were excited at one end of a bar of strongly piezoelectric lead-titanate-zirconate ceramic, of dimensions 0.95 × 0.95 × 27 cm. The strain gauge in this case was mounted on a patch of silver substrate and was shielded with conductive paint to eliminate interference due to piezoelectrically induced electric fields. Electrodes (0.3 × 0.95 cm) painted on the lateral surfaces of the bar were used to monitor the electrical signal produced as a result of the piezoelectric polarization in the bar.

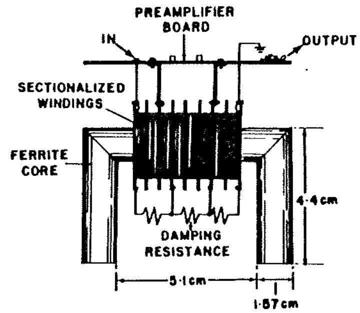

Fig. 1. **Diagram of magnetic sensor.**

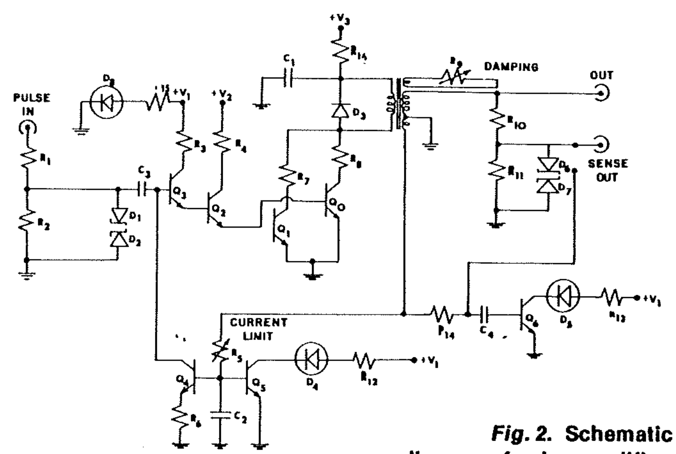

Fig. 2. **Schematic diagram of pulse amplifier.**

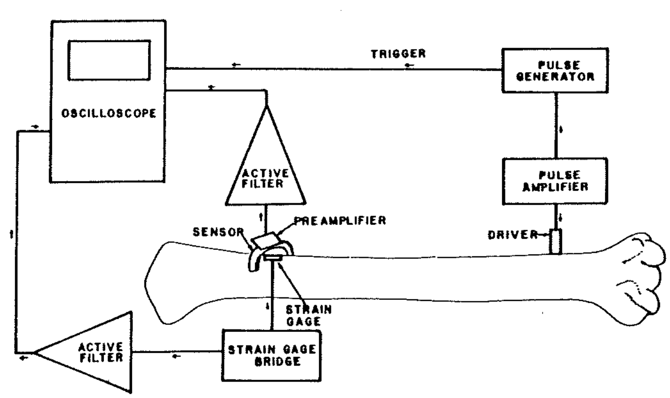

Fig. 3. **Experimental configuration.**

[1]R. S. Lakes and S. Saha. Unpublished data.

In tests upon human volunteers, the driver was pressed, with a force of 23 newtons, to the skin over the anterior portion of the tibia, about 8 cm distal from the tibial plateau. The sensor was placed about 20 cm distal to the driver, 0.5 cm above the skin over the anterior portion of the tibia. Signals obtained using wet bones, either in vitro or in vivo, are significantly weaker than signals from dry bones; therefore, a signal averager (model TDH9, Princeton Applied Research Corp., Princeton NJ) was used to extract the repetitive signals from random noise in the sensor.

Results and Discussion

Typical sensor and strain gauge outputs for dry bone are shown in Fig. 4. Some dissimilarities between the magnetic sensor and strain gauge outputs are to be expected, since the strain is measured at a point on the surface while the sensor responds to strain-generated polarization in a stressed region of the bone in close proximity to the sensor. Nevertheless, the device produces a signal proportional to the strain amplitude in the bone.

In Fig. 4, the maximum strain amplitude in the bone is of the order 1.4×10^{-6}, which is less than 0.5 percent of the strain typically measured in the leg bones of walking animals (15). The first peak in the strain signal occurs 110 μsec after the exciting pulse. Since the driver was 11 cm from the strain gauge, the fastest bending waves generated travel at 1000 m/sec. The speed of plane longitudinal waves in bone, by contrast, is about 3500 m/sec (1).

The speed of bending waves depends on their frequency and in general is less than the speed of longitudinal waves (11). Moreover, the propagation of stress waves in hollow tubes is quite complex. The general problem has been solved theoretically for elastic solids (10). In bone, the bending waves propagate in a complex fashion as a result of the hollow geometry of bone and also the natural dispersion of bending waves; therefore, the incident 10-μsec-wide pulse has resulted in the strain waveform shown in Fig. 4.

A variety of tests were performed to ensure that the sensor output did, in fact, result from piezoelectric polarizations in bone: First, the driver was energized and removed from the bone, which generated no waves causing the strain gauge signal to vanish. Simultaneously the magnetic sensor output also vanished. Second, the driver was pressed to the bone, generating stress waves. When the magnetic sensor was moved to a distance of 1 m from the bone, the signal again vanished. Third, when stress waves were generated in strain-gauged bars of aluminum and Plexiglas, which are not piezoelectric, no magnetic sensor signals were observed. Fourth, the sensitivity of the preamplifier input to electric fields was examined by disconnecting the coil winding and replacing it with an equivalent resistance. The physical arrangement of the components was otherwise unchanged.

Excitation of the preamplifier by piezoelectrically generated electric fields accounted for 16 percent of the total "magnetic" output observed in the ceramic rod. The removal of the dc excitation voltage from the strain gauge had no effect on the magnetic sensor output. Currents in the strain gauge circuitry are therefore not responsible for the magnetic sensor signals.

These tests indicate that the observed magnetic sensor signals result from fields generated by stress waves in piezoelectric materials only, and that the sensor must be near the stressed region for signals to be recorded. Although electric fields did contribute to the output, this contribution was small and could be reduced further (at some cost in sensitivity) by decreasing the impedance of the preamplifier.

Signals resulting from stress waves in a piezoelectric ceramic rod are shown in Fig. 5. The propagation of longitudinal waves in this rod of strongly piezoelectric ceramic could be analyzed more easily because little distortion of the incident waveform took place. In this model system, the waveforms associated with the strain, electric field, and magnetic sensor output exhibited a similar shape (Fig. 5). In addition, several reflections of the wave from the ends of the specimen could be readily observed. Such reflections are not prominent in bone due to the larger attenuation values that bone exhibits. In particular, the trabecular bone at the ends of a long bone has a very high attenuation coefficient.

The waveforms shown in Fig. 5 are simpler than those for bone, since the single, 10-μsec-wide pulses can be identified easily. The wave speed calculated from the experimental results was 3040 m/sec. By comparison, the bar velocity ($v = \sqrt{E/P}$) computed from the manufacturer's specifications of density (7.45 gm/cm^3) and Young's modulus (7.27×10^{10} N/m^2) was 3140 m/sec, a satisfactory agreement in view of the 10 percent tolerance given for these specifications.

Wet bone differs from dry bone in its properties. For example, wet bone is more compliant than dry bone by about 18

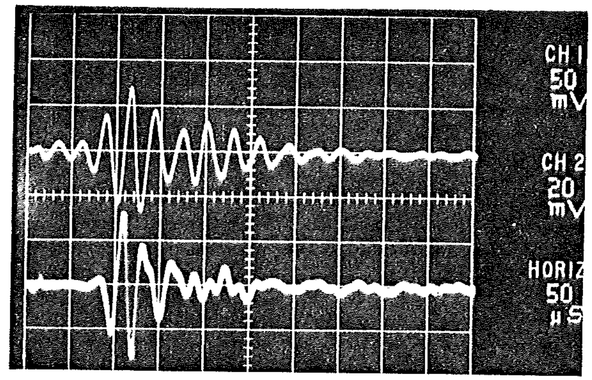

Fig. 4. Signals associated with stress waves in dry bone. *Top:* Magnetic sensor output, amplified. *Bottom:* Strain gauge output, 8.7×10^{-7} per division. Horizontal scale: 50 μsec per division.

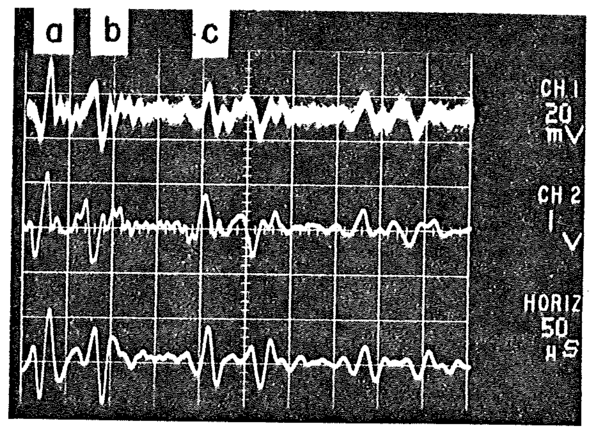

Fig. 5. Signals associated with stress waves in a piezoelectric ceramic. *Top:* Strain gauge output, 0.9×10^{-7} per division. *Middle:* Electrode output, 200 mV per division. *Bottom:* Magnetic sensor output, amplified. Horizontal scale = 50 μsec per division; delay = 40 sec; a = peak for directly transmitted wave; b = first reflection from the end of the rod; c = the wave following reflection from the driven end.

percent (7), resulting in a slower wave speed in wet bone. More significantly for the purpose of the present experiments, the piezoelectric coefficients of bone vary by a factor of from 2 to 10 (4,17), and the dielectric coefficients vary by several orders of magnitude as the bone hydration is varied (14). Wet bones, therefore, were also examined using the present device. For wet bone in vitro, the sensor output was reduced, as shown in Fig. 6.

In vivo studies of human volunteers were also performed since fresh bone in the body might be expected to differ from embalmed bone in its dielectric behavior. Typical in vivo results obtained using signal-averaging techniques are shown in Fig. 7. Although the outputs obtained in these in vivo tests were significantly smaller than those obtained in the in vitro tests, it is clear from Fig. 7 that the signal could be easily recorded and that it was much larger than the coherent noise level, which is shown by the bottom curve in Fig. 7.

Summary

The electromagnetic device described in this paper detects signals associated with stress waves in a bone, without contacting the soft tissue. This overcomes the difficulties encountered in previous wave-propagation and vibration tests, in which the results were affected by the properties of soft tissue and by the magnitude of the preload force applied on the sensing transducer.

The magnetic signal detected by the present device depends on the mechanical and piezoelectric properties, as well as on the histological structure (e.g., porosity) of bone. Therefore, this technique is potentially useful in the diagnosis of bone disorders in which these physical properties are affected. Examples include the diagnosis of osteoporosis and the evaluation of fracture healing.

Another possible area of application of this device is to noninvasively detect the piezoelectric character of bone. The piezoelectric effect in bone has been implicated as a possible mechanism for Wolff's law activity, by which bone is remodeled in response to stress (3). Since the magnetic technique provides a measure of piezoelectrically induced polarizations, this technique may be used to noninvasively determine these polarizations in vivo. If an independent estimate of the stress amplitude can be obtained, the value of the piezoelectric modulus can be calculated. Such a measurement might be useful in investigations of bone remodeling activity.

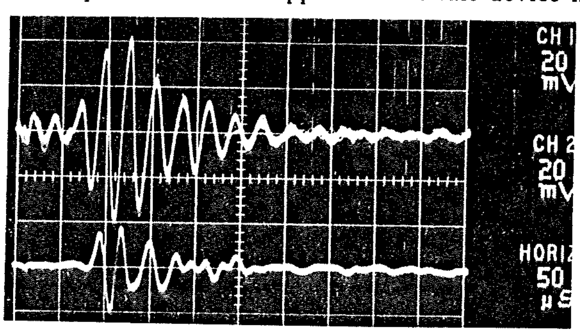

Fig. 6. Signals associated with stress waves in wet bone. *Top:* Magnetic sensor output. *Bottom:* Strain gauge output, 8.7 × 10^{-7} per division. Horizonal scale: 50 μsec per division.

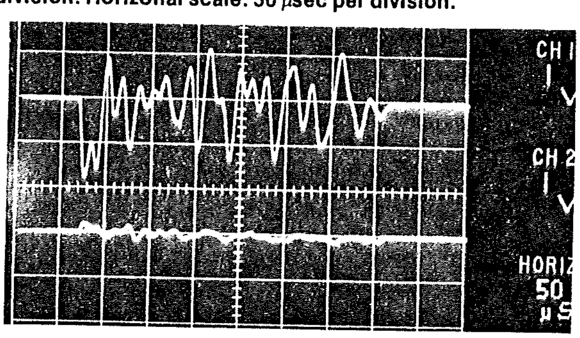

Fig. 7. Signals associated with stress waves in a human tibia in vivo. *Top:* Magnetic sensor output, with leg in place. *Bottom:* Sensor output with leg removed; driver is still energized. With signal averaging. Horizontal scale: 50 μsec per divison.

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