JOURNAL OF MATERIALS SCIENCE: MATERIALS IN MEDICINE 2 (1991) 118–124

The electromechanical properties of fluid-filled bone: A new dimension

G. W. HASTINGS, F. A. MAHMUD
Department of Materials, Queen Mary and Westfield College, University of London, Mile End Road, London, E1 4NS, UK

The current understanding of electromechanical effects in fully hydrated bone is that they are electrokinetic (streaming potential) in nature. The presence of a second competing mechanism, piezoelectricity, which dominates in dry conditions, has been denied in conditions of full hydration based on the absence of a piezoelectric response from wet collagen. Since the mineralized collagenous matrix of bone can only absorb 26 wt% water (Relative Humidity (RH) = 60%), there seems no reason to dismiss a piezoelectric component entirely and experimental evidence was sought using a new measuring technique. A comprehensive analysis has been developed to relate both mechanisms to bone structure at different levels of hydration. Our results indicate the presence of both mechanisms at full hydration, with the piezoelectric effect leading streaming potential in the time domain. The immediate implication of this finding is that it is the piezoelectric effect which determines the characteristics of the generated electrical signal and may subsequently influence bone generation and remodelling.

1. Introduction

The electromechanical transduction in bone has been thoroughly investigated over the past three decades. A review of the current state of knowledge [1–3] reveals that two different mechanisms are responsible for this transduction in dry and fully hydrated bone. Piezoelectricity is the generating mechanism in dry bone, and streaming potential is thought to be dominating in conditions of full hydration. In either case, the generation of the electrical response takes place within the mineralized collagenous matrix of bone.

The electrical/electromechanical phenomenon in bone matrix is derived from the structural components of its collagen. At molecular level, the location of the charged residues along the molecular axis can be identified, by using X-ray diffraction (XRD) techniques, to a resolution of approximately 0.6 nm [4]. From an electrostatic point of view, there are approximately 250 positively charged (basic) and 230 negatively charged (acidic) residues per collagen molecule under physiological conditions. Most of these residues are neutralized locally through the process of stabilization of the molecule or the fibril and the remaining "unbonded" charged residues have an excess of positive charges, thus giving the molecule a nett positive charge.

Katz *et al.* [4] have shown the charge distribution of the residues both in the helical parts of the molecule and in the side chains not involved in salt-link formation. They indicated that there are about ten regions along the 67 nm segment that can be considered to be very large dipoles. These dipoles are concentrations of positive charges separated spatially from concentrations of negative charges by distances ranging from 2.5 to 7 nm. The above authors also indicated that many of these dipoles are paired with adjacent dipoles, having an opposite polarization, and that they are somewhat periodically disposed along the fibril axis. The ordered structure of collagen would therefore contain an aggregation of orientated dipoles in the form of dipole domains.

The piezoelectric potential developed across a dry bone specimen, according to a model proposed earlier by the present authors [5], results from the displacement of the dipoles in the mechanically deformed collagenous matrix. The model also demonstrates that the characteristics of the piezopotential signal are largely influenced by the hydration condition of the specimen. Unfortunately, data from partially hydrated specimens were not obtainable until now due to the unavailability of a reliable electrode system. Such data would contribute to our understanding of the actual mechanism(s) in partial and full hydration.

In full hydration, the generated potential is short-lived due to charge leakage taking place within the specimen. The inability to capture this short-lived potential by conventional measuring techniques, and the fact that collagen acquires higher symmetry levels at full hydration, has lead to the dismissal of any piezoelectric effect in full hydration, and consequently streaming potential was proposed as the dominating mechanism for the observed potential. Streaming potential arises from the existence of a relative velocity between an ionic fluid and a charged surface, in the case of bone, between tissue fluid and the mineralized collagen matrix. In the present study, attention was focused on the concept of the charged surface and the mechanism(s) by which it becomes charged were investigated. Our aim in this publication is to draw attention to the fact that what is presumed to be static

0022–2461/91 $03.00 + .12

(charged surface) becomes electrically active when deformed mechanically, and therefore exerts a controlling function on the electromechanical properties of fluid-filled bone.

2. Theory

Piezoelectric studies are usually performed on dry bone. In considering wet bone or collagen the effect of water in structural modification is important. The collagen helix is saturated at 26 wt % water (RH = 60%). At lower levels the interfibrillar spaces become progressively unfilled as water preferentially adsorbs to the collagen molecule. The actual influence of adsorbed water at hydration levels below 26 wt %, as Maeda and Fukada [6] speculated, is that water forms interchain hydrogen bonds inside the triple helices of collagen (Fig. 1) increasing the crystallinity of collagen with a corresponding increase in the piezoelectric coefficient (d). However, water molecules, at hydration levels above 26 wt %, are adsorbed between the triple helices and microfibrils, thus expanding the distance between them. Such structural changes, according to Maeda and Fukada [6], decrease the effective density of piezoelectrically active dipoles, thus leading to decrease of (d) in decalcified bone.

The mineral hydroxyapatite in bone, limits the efficiency of adsorbed water for expansion of the distance between the triple helices and microfibrils, but comparison of the adsorption isotherms of bone and decalcified bone [6], shows that bone mineral decreases the number of water adsorption sites and in effect displaces water from and/or restricts its access to the collagen molecules, thus partially dehydrating the collagen. Therefore the reduction in piezoelectric response may not be as marked as would be expected. This view is supported by the works of Anderson and Erikson [7], Reinish and Nowick [3] and Marino and Becker [9].

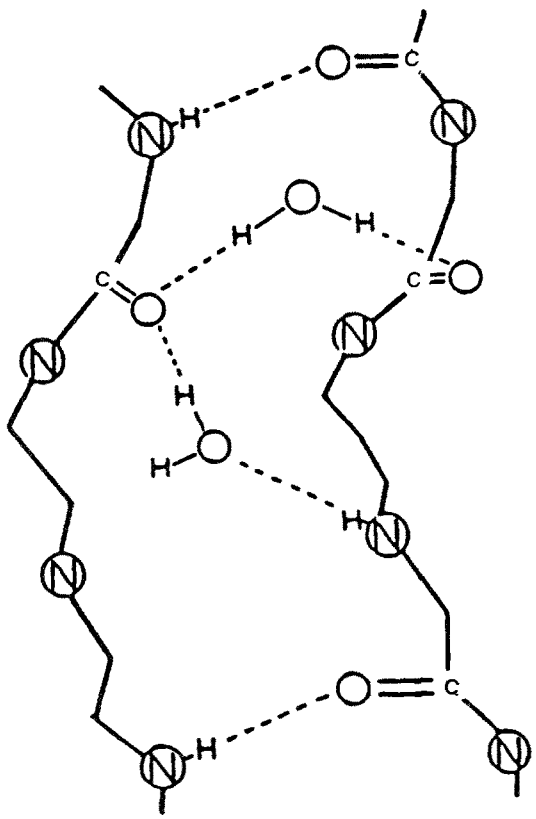

Figure 1 Schematic representation of the binding of water molecules in collagen.

In *in vivo* conditions, collagen molecules are bathed in ionic (tissue) fluid. To correlate *in vitro* studies to the actual *in vivo* condition and to develop a comprehensive understanding of the electromechanical mechanism *in vivo*, the interaction of collagen with its ionic bathing medium must be elaborated. In aqueous media, free penetration of ions into the collagen structure is possible and by interaction with charged side chains this reduces fibrillar potential. Fibrillar potential is thus influenced by the ionic strength of the fluid environment. In fact Katz *et al.* [4] reported that at physiological ionic strength, the fibrils have an average potential that is approximately 4 mV more positive than the incubating solution. It is therefore concluded, that the electrostatic potential difference between a collagen molecule and its bathing solution is proportional to the nett charge on the fibrils and the ionic strength of the solution [10]. The former, is a function of the state of ionization of the charged residues of the protein, as well as the size of the intermolecular spaces of the fibril. Inclusion of an intermolecular space factor accounts for changes in electrostatic charge density resulting from mechanical perturbation of the fibrils. The dependence of the fibrillar potential on the ionic strength of the solution is governed by an exponential function [10], such that it decreases as the ionic strength is increased and reaches an asymptote at physiological concentrations, i.e., collagen fibrils retain a nett positive charge at physiological ionic concentration.

A positively charged collagen molecule in an aqueous ionic solution has a double electrical layer at the surface/solution interface. The extent of this double/diffuse layer (often called Debye length) will depend on the ionic concentration of the fluid. The double/diffuse layer contains excess anions compared to the bulk fluid. If a mechanical compression (sufficient to cause fluid flow but insufficient for fibrillar deformation to occur) is applied, the resulting potential would derive solely from the streaming potential mechanism. However, in practice this is not the case because fluid flow occurs as a consequence of fibrillar deformation, i.e., the compression is applied to the collagen matrix and not directly to the fluid. Thus the concept of streaming potential, in its classical form, is not valid in the case of fluid-filled bone. Therefore, a combined effect resulting from compression of the matrix (piezoelectric response) and flow of fluid (streaming potential) is expected.

The proposed sequence of events leading to an expression of a generated potential in fluid-filled bone is summarized as follows: In mechanically unstressed collagen an equilibrium potential difference is present resulting from the ionic environment within the fibrils and in the surrounding fluid and this gives rise to the double/diffuse layers of ions (Fig. 2), the electrostatic effect. Applied stress changes the electrostatic charge

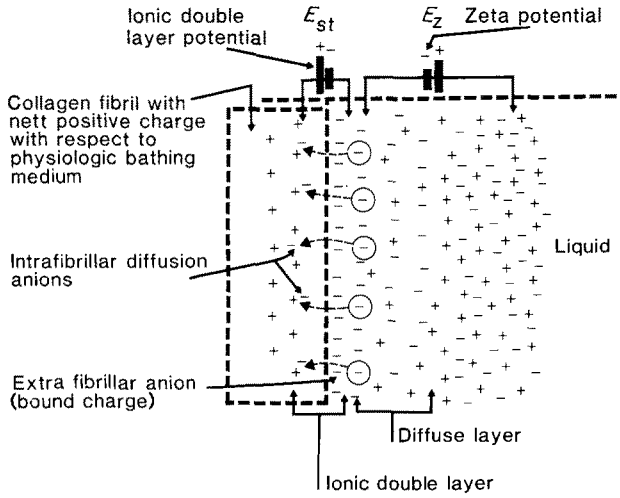

Figure 2 The interaction between collagen fibril and ionic bathing medium, electrostatic effect.

density which thus leads to a variation in ion flow already initiated by pressure gradient (streaming potential). Increased stress levels lead to greater charge density change and more fluid expulsion and together these enhance the ion double layer. Mechanically derived ionic fluid-flow (streaming potential) is thus modified and controlled by the additional charge density change in the collagen fibrils (piezoelectric effect), Fig. 3, the electrodynamic effect.

3. Materials and methods

Bone specimens used throughout the present investigation were machined from adult bovine tibiae obtained in fresh condition. On receipt of bone, adventitious tissues were removed immediately and the epiphyseal ends of each tibia were cut using a band saw. The marrow was then removed from the diaphyseal shafts, which were then sectioned, by using a hand saw running at low speed, in the axial direction to produce two to four longitudinal sections. Each section was further cut to obtain rectangular shaped specimens, which were then smoothed on a milling machine. Dripping distilled water was used throughout cutting and machining to minimize thermal damage. The finished size of the specimens measured approximately

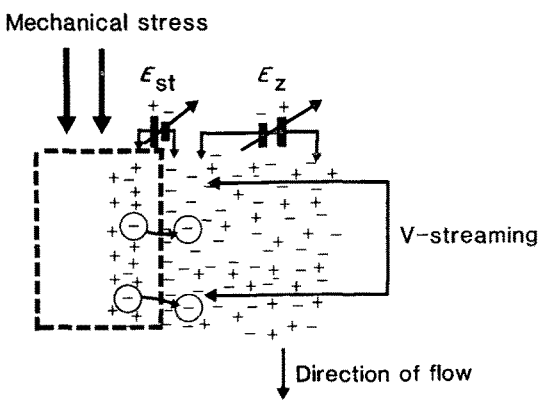

Figure 3 The electrodynamic effect, applied stress changes the electrostatic charge density (E_{st}) which thus leads to a variation in ion flow already initiated by pressure gradient (streaming potential).

50 mm × 10 mm × 2 mm. Removal of the fat from the specimens was achieved by immersion in trichloroethylene for several days until it became clear that no more fatty spots were appearing on the surface of the liquid. The specimens were then washed several times in deionized water to remove any debris remaining from the machining procedure. Finally, the specimens were kept immersed in deionized water at 4 °C until they were used. Storage time did not exceed 24 h, during which time degeneration of the bone collagen is not expected to have occurred.

Dynamic deformation of the specimens was achieved by a specially designed four-point bending rig and the use of a hydraulic fatigue testing machine (ESH Testing Limited, Brierley Hill, West Midlands, UK). The four-point bending rig has been designed and manufactured to house a pair of non-contacting electrodes. Details of the measuring apparatus, including the construction of the non-contacting electrode, can be found in a previous publication [11]. The four-point bending rig was enclosed in a shielded and sealed humidity chamber whilst being loaded in the ESH machine, Fig. 4.

4. Experimental

To obtain recordings of the strain generated potential (SGP) at various hydration levels, specimens previously washed with deionized water were allowed to equilibrate, in the humidity chamber, at the desired RH for 24 h prior to testing. This procedure ensured uniform hydration throughout the thickness of the specimen at the desired level. The testing protocol involves applying 30 N preload to the specimen, which is placed in the four-point bending rig, followed by a square wave loading signal of 40 N magnitude and 1 s duration. The SGP is then recorded simultaneously with the loading signal and is shown in trace (a) of Fig. 5 for the highest relative humidity level (98% RH). The specimen under test is kept in the humidity chamber while the RH is lowered to the next desired level (85% RH). After allowing 24 h equilibration time, the above loading procedure is repeated and the resulting SGP at 85% RH recorded (Fig. 5, trace (b)). Traces (c), (d), (e) and (f) are obtained by following the above procedure for the relative humidities, 75%, 55%, 35% and 15%, respectively. Traces in Fig. 5 give a representative record of the SGP waveform from one specimen, while the results from 10 specimens are tabulated in Tables I and II. Fig. 6 shows a plot of the peak-to-peak value (V_{pp}) of the SGP versus relative humidity as obtained from the results in Table I. The relationship between the magnitude of the SGP and RH is illustrated in the three zones in Fig. 6. It is clearly evident from this figure that the V_{pp} is relatively independent of humidity level in Zones I and III (% RH is either lower than 35 or higher than 85% RH, respectively) and is strongly dependent on RH in Zone II (% RH is between 35 and 85%). The other aspect of the dependence of the SGP on the RH level is the shape of the recorded waveform, Fig. 5. This is expressed as the ratio of the positive phase to that of the negative phase, Table II. The ratio approximates to unity at very high RH (98%)

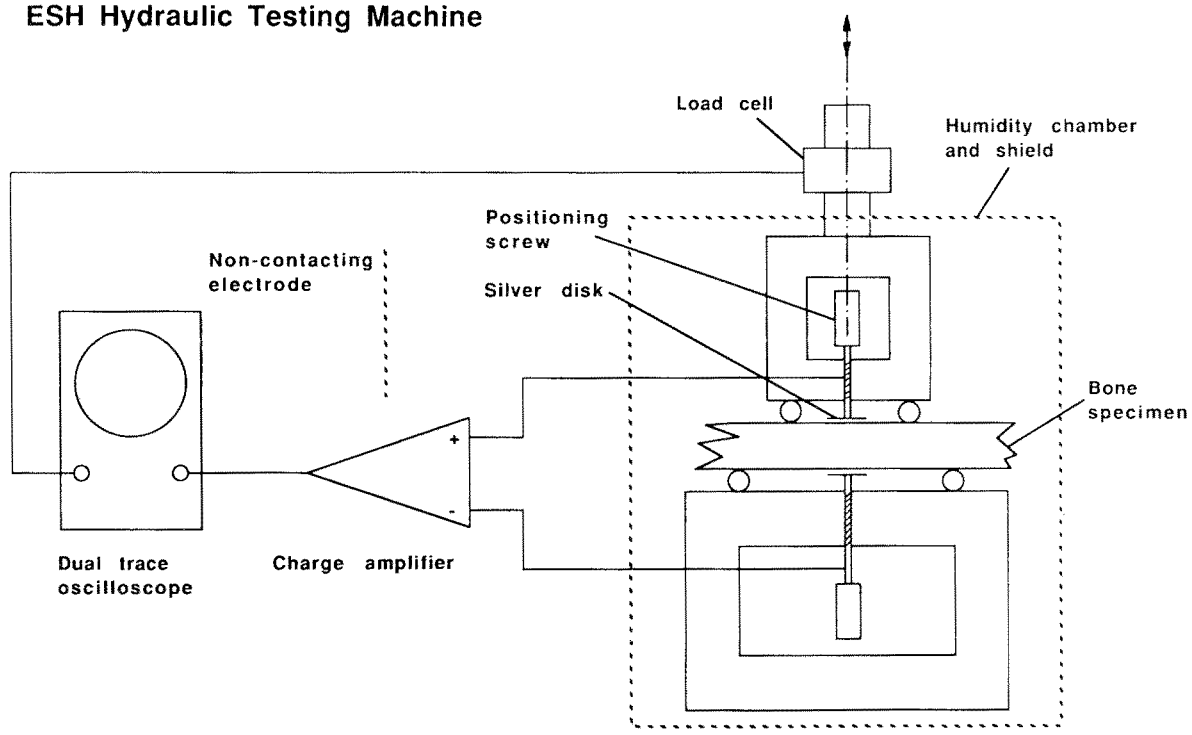

Figure 4 Schematic diagram showing the non-contacting electrode system and measuring circuit (reproduced from Hastings *et al.* [11]).

and very low RH (15%), indicating waveform symmetry. In contrast with this is the considerable variation of the ratio at other RH levels which always exceeds unity. Full consideration of these findings will be given in the discussion section.

One of the main objectives of the present study was to establish whether or not the mineralized collagenous matrix of bone retains its electrical activity in conditions of full hydration. The following experiment has been designed to verify the above postulation by recording the SGP from specimens in fluid-filled and fully hydrated conditions. A number of hydrated specimens which were kept immersed in deionized water for several weeks were employed. Each specimen was loaded in four-point bending mode and tested by the application of a square wave loading signal of 50 N magnitude. A relative humidity of 98% was maintained throughout testing. Trace (a) in Fig. 7 shows the SGP signal from a specimen saturated with deionized water, thus representing the electromechanical response from the collagenous matrix, since virtually no ionic fluid is present. In trace (b), the signal is recorded from the same specimen after immersion in physiological saline and application of similar loading conditions. The peak-to-peak value of the SGP signal in both traces approximate each other and the longer decay time in the signal in trace (b) is due to a streaming potential effect (ionic flow).

Traces (a) and (b) in Fig. 8 represent the SGP response from a specimen saturated with physiological saline and loaded by square wave mechanical loading (trace (c)) in four-point bending mode. Trace (a) is obtained immediately after the specimen is

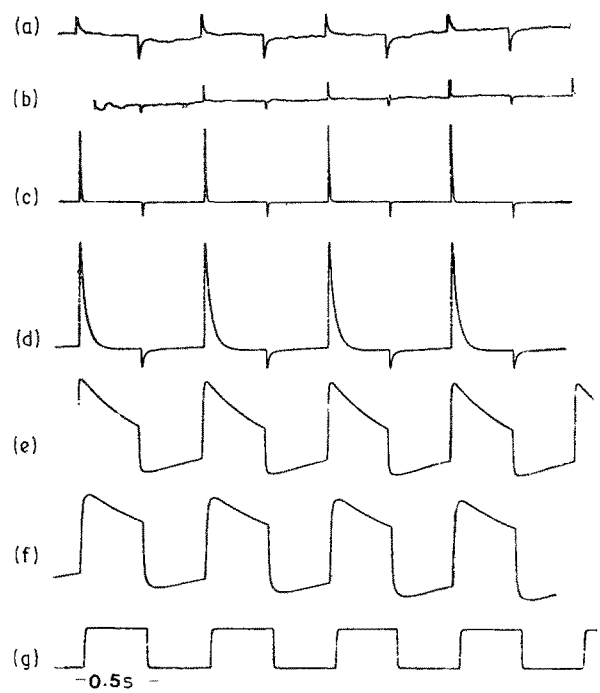

Figure 5 The SGP from fluid filled bone at different levels of RH: (a) 98% (0.16 mV); (b) 85% (0.10 mV); (c) 75% (1.70 mV); (d) 55% (4.80 mV); (e) 35% (7.00 mV); (F) 15% (7.00 mV); (g) Loading signal (40 N).

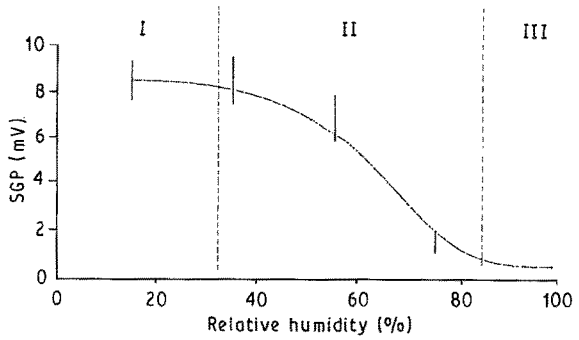

Figure 6 Effect of relative humidity on the SGP of fluid filled bone.

TABLE I The effect of relative humidity (%) on the SGP of bone

Sample No.	98% V_{pp} (mV)	85% V_{pp} (mV)	75% V_{pp} (mV)	55% V_{pp} (mV)	35% V_{pp} (mV)	15% V_{pp} (mV)
1	0.16	0.10	1.70	4.80	7.00	7.00
2	0.15	0.12	0.51	7.00	6.80	8.60
3	0.12	0.10	0.90	8.00	8.50	8.50
4	0.10	0.10	1.00	8.40	10.0	9.80
5	0.14	0.13	0.80	6.30	9.00	9.00
6	0.15	0.14	1.40	6.40	8.00	8.00
7	0.12	0.26	1.50	6.50	9.00	9.60
8	0.20	0.10	1.90	7.60	9.50	8.40
9	0.10	0.25	1.90	5.80	7.50	8.00
10	0.20	0.12	0.90	6.30	8.50	7.50
Average	0.144	0.142	1.25	6.71	8.38	8.44
s.d. ±	0.03411	0.05810	0.4679	1.0173	0.9997	0.8296

TABLE II The effect of relative humidity (%) on the shape of the SGP signal (indicated by the ratio of the positive/negative phases of the signal)

Sample No.	98% $V_p^{(+)}/V_p^{(-)}$	85% $V_p^{(+)}/V_p^{(-)}$	75% $V_p^{(+)}/V_p^{(-)}$	55% $V_p^{(+)}/V_p^{(-)}$	35% $V_p^{(+)}/V_p^{(-)}$	15% $V_p^{(+)}/V_p^{(-)}$
1	1:1	2.3:1	4.6:1	5:1	1.6:1	1.6:1
2	1:1	3.2:1	4.9:1	5.1:1	2:1	2:1
3	1:1	2.7:1	5.1:1	3.8:1	1.6:1	1.1:1
4	1:1	2.4:1	4.2:1	4.5:1	1.3:1	1.1:1
5	1:1	3.1:1	3.5:1	4.4:1	1.7:1	1.2:1
6	1:1	2.5:1	4.5:1	5.8:1	1.9:1	1.6:1
7	1:1	2.1:1	4.8:1	5.5:1	1.7:1	1.3:1
8	1:1	2.3:1	3.8:1	4.9:1	2:1	1.6:1
9	1:1	3.3:1	4.5:1	5.6:1	1.5:1	1.5:1
10	1:1	2:1	4.5:1	4.3:1	1.9:1	1.2:1

placed in the humidity chamber and therefore represents the signal from fluid filled bone. In trace (b), the signal is recorded after allowing the specimen to drain for several h in the humidity chamber at 98% RH. Thus representing the SGP signal from hydrated bone, i.e., little ionic flow, and is therefore mainly contributed by the electromechanical effect of the collagenous matrix.

5. Discussion

The main objective of the present study was to determine whether the mineralized collagenous matrix of bone retains its electromechanical activity in fully hydrated conditions. If so, the interaction of the ionic tissue fluid with the electrically active matrix can be elucidated and finally a comprehensive understanding of the mechanism(s) underlying the electromechanical effect in fluid filled bone and *in vivo* can be achieved.

Fibrous collagen macromolecules are packed together in bundle-like formations with open channels running along the fibre direction into which chains of water molecules can fit. Bone matrix at full hydration can only absorb 26 wt % water and may be considered as essentially low conductivity mineralized collagen cores separated by conducting layers and channels of water molecules. This excess water will act as a plasticizer and also increase the effective dielectric permit-

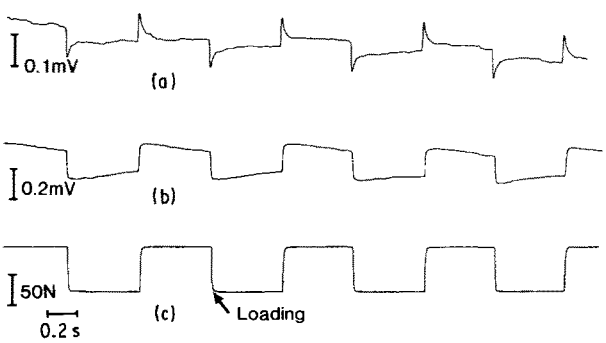

Figure 7 The SGP signal from fluid filled bone: (a) deionized water; (b) physiologic saline; (c) loading signal.

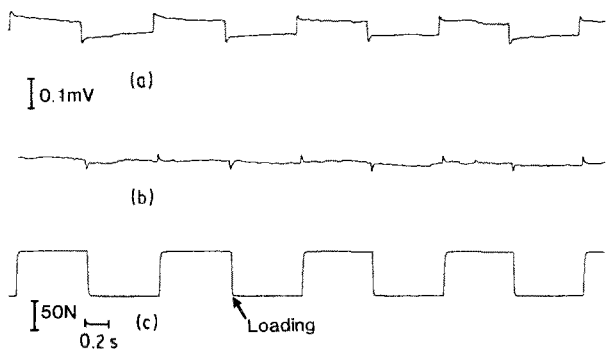

Figure 8 The SGP signal from fluid filled bone (a) and hydrated bone (b). Trace (c) represents the loading signal.

tivity of the specimen. Mechanical deformation of the matrix may result in changed equilibrium of the charge density of the less conductive protein, due to side-chain motions, and also in breakage and reformation of hydrogen bonds holding the adsorbed water molecules to the structure. The resultant electromechanical signal may therefore be a superposition of these two effects. Results in Fig. 7(a) confirm the above hypothesis of two mechanisms. Examination of the decay curve of the SGP signal reveals the presence of two decay (relaxation) times, τ_{fast} and τ_{slow}, where τ_{fast} is the relaxation time constant of the (solid protein phase) and τ_{slow} is the relaxation time of the fluid phase (water). This implies a contribution to the recorded signal from at least two mechanisms. Pollack *et al.* [12] referred to such analysis of the relaxation time when interpreting the SGP signal from step loading, and commented that a second contributory mechanism (in addition to streaming potential) may exist in fluid filled bone.

At lower hydration levels (i.e., below 26 wt % water), water molecules are displaced from the channels and become preferentially adsorbed to localized sites within the helical structure, thus breaking the water conductive layer in the channels, and as dehydration is further increased, it becomes difficult to be certain as to the extent to which the adsorbed water penetrates into the structure. Consequently, the permittivity and the electrical conductivity of the dehydrated structure becomes more dependent on the local hydration level. The relationship is such that the permittivity decreases exponentially with increasing dehydration levels [13]. The heterogeneous structure of bone is therefore expected to have a large influence on the characteristics of the SGP signal. This is reflected in the shape and magnitude of the SGP signal, Figs 5(b)–(d) and 6. For the range of humidity between 35–85% RH, the peak-to-peak value of the SGP signal increases with increasing dehydration, Fig. 6, due to the increase of the contribution from the piezoelectric effect. A characteristic feature of the shape of the SGP signal at the above mentioned range of relative humidities is the asymmetry in the negative/positive half cycles of the recorded signal (corresponding to the loading/unloading phases, which is also expressed by the ratio $\pm$ in Table II), Fig. 5(b)–(d). This feature of the SGP signal may be interpreted in the light of the work of Reinish and Nowick [14] on the electrical properties of bone. They reported that the moisture content of a sample of bone is not a unique function of the relative humidity of the atmosphere with which the sample has equilibrated. This is to say that a hysteresis-like relationship during drying and wetting cycles of the sample at various RH levels exists. Similar behaviour has been reported by the above authors for the d.c. conductivity, dielectric constant, and the piezoelectric coefficient (d) of rehydrated bone. Such hystereses have been attributed to the condensation of water in the capillaries and pores and also to surface tension effects which prevented the removal of previously condensed moisture. It may therefore appear that the asymmetry of the SGP signal in traces (b)–(d) of Fig. 5 is due to a combined effect from the conductivity of the surface and the mobility of the condensed water in the porous structure behaving differently during the loading/unloading cycles. At hydration levels below 35% RH, the SGP signal is typical of that obtained from a piezoelectric material, Fig. 5, traces (e) and (f). Dehydration of the specimen causes the SGP signal to decay at a lower rate due to the decrease in charge leakage within the specimen.

The above results, Fig. 5, have established that a piezoelectric effect is present in fully hydrated bone matrix saturated with deionized water, i.e., in the absence of freely mobile ions. The introduction of freely mobile ions (such as Na^+ and Cl^-) to the bathing fluid, as far as the electromechanical effect is concerned, leads to a series of events which can be best understood when dealt with at the molecular level.

The first of these events is the reduction in the electrostatic fibrillar potentials which concern the interaction of collagen fibrils with the ionic bathing solution. This is characterized by the penetration of free ions into the inter- and intra-molecular spaces which is confirmed by the observed dielectric dispersion at low frequencies (< 10 Hz) reported by Kosterich *et al.* [15]. The dielectric dispersion has been related, by them, to the presence of diffusion-controlled ionic polarization which is a manifestation of an ionic diffusion mechanism taking place along collagen fibrils.

The second event is the formation of the double/diffuse layers near the charged surface of the fibril, as shown in the model of Fig. 2. This is related to the electrical interaction between the charged collagen fibril and its extrafibrillar ionic fluid occupying the vascular structure of bone matrix. The electromechanical response following mechanical compression of such a model incorporates fibrillar deformation (change in spatial charge density), and the flow of ions out of the fibrils as well as in the microvoids of the vascular structure (streaming potential effect). Since the solid phase receives all of the mechanical compression impulse and deforms faster than the fluid phase, it is conceivable that the generated potential represents the response from the two phases in the above sequence of events, i.e., change in spatial charge density followed by ionic movement (streaming potential). Thus one can see how the response from the solid phase influences streaming potential through its controlling role on the ionic concentration of the double/diffuse layers. Moreover, it should be emphasised here that the flow of ions takes place only as a result of matrix deformation and a distinction must, therefore, be made from the classical streaming potential mechanism where ions flow in response to direct application of mechanical pressure to the fluid phase, i.e., no deformation occurs in the solid phase.

The electromechanical response from the combined effects of the solid and fluid phases, for a specimen saturated with physiological saline and in response to step-loading mechanical deformation, is shown in trace (b) of Fig. 7. The peak-to-peak value of this signal compares with that obtained from the same specimen saturated with deionized water indicating structural (solid phase) saturation. The decay (relaxa-

tion) time of the signal in trace (b) is longer than that in trace (a). This is due to the presence of Na^+ and Cl^- ions which have lower diffusion coefficients than that of pure water. The longer decay time in (b) is comparable to that from the streaming potential mechanism (0.4–0.5 ms).

The in-phase relationship between the two traces (relative to the loading signal, trace (c)) confirms the earlier postulation of the electrical response from the solid phase leading the streaming potential in the sequence of events, which is in total agreement with the model proposed in Fig. 2.

When the ionic fluid is left to drain from a specimen under high relative humidity (98% RH), the new condition is considered ionic saturation at full hydration, i.e., ions are less mobile due to the lower level of fluid in the vascular structure and ionic adsorbtion to charged side-chains. The electromechanical effect from such a specimen would therefore be that of the solid phase. However, the generated potential is expected to be of small magnitude and short decay time due to the shunting effect of the adsorbed ions. Trace (b) in Fig. 8 represents the SGP signal from an ionic saturated fully hydrated specimen subjected to step loading deformation. Once again, this signal has a similar phase relationship with that from the same specimen in the fluid filled condition (relative to the loading signal, trace (c) in Fig. 8), indicating that the generated potential is that of the solid matrix of bone.

6. Conclusions

The following conclusions can be drawn from the present study:

1. The collagenous matrix of bone retains its electrical activity in full hydration and in fluid-filled conditions and an electromechanical response in the form of a potential generated in response to mechanical deformation can still be produced.

2. Although the presence of free mobile ions in the bathing fluid reduces the electrostatic potential of the collagen fibrils, it does not completely neutralize it. At physiological ionic strength, the fibrils retain a nett positive electrical potential relative to their bathing medium.

3. The overall electromechanical response from fluid-filled bone and possibly from bone *in vivo*, contains at least two components; the first arises from the deformation of the collagenous matrix (solid phase) and the second from the motion of ions within the fibrillar spaces and in the vascular structure (fluid phase).

4. The electromechanical response from the solid phase leads that from the fluid phase in the time domain, thus exerting a controlling function upon the flow of ions.

5. The classical definition of the streaming potential mechanism does not totally apply to the electromechanical response from fluid-filled bone for the reason that the mechanical pressure required to cause fluid flow is applied to the bone matrix and not directly to the fluid, and the controlling effect of this matrix must be considered.

6. Due to the heterogeneous nature of bone matrix and the discontinuity of the fluid in the matrix from the perivascular fluid (PVF) in the capillaries, a unidirectional flow of fluid from the compression side of bent specimen to the side undergoing tension, as has been postulated by many investigators, is not likely to occur. Instead, localized relative motion of ions takes place within the fibrillar structure and limited fluid flow in the vascular network occurs.

7. Although the electromechanical response from the solid phase may be related to a piezoelectric mechanism in the sense that a mechanical deformation of the matrix leads to the generation of a potential signal, through changes in the charge density of the structure, the exact interpretation in terms of a classical piezoelectric theory may require further investigation in view of the effect of adsorbed ions from the bathing fluid. The authors would therefore recommend the use of the general term "electromechanical response" when referring to the strain generated potentials from fluid filled bone.

References

1. G. HASTINGS and F. MAHMUD, *J. Biomed. Eng.* **10** (1988) 515.
2. S. SINGH and J. KATZ, *J. Bioelectr.* **7** (1988–9) 219.
3. D. CHAKKALAKAL, *J. Mater. Res.* **4** (1989) 1034.
4. E. KATZ, N. WAX and S. LI, in "Electrical Properties of Bone and Cartilage" edited by C. Brighton, J. Black and S. Pollack (Grune and Stratton, New York, 1979) p. 131.
5. F. MAHMUD, G. HASTINGS and M. MARTINI, *J. Biomed. Eng.* **10** (1988) 54.
6. H. MAEDA and E. FUKADA, *Biopolymers* **21** (1982) 2055.
7. J. ANDERSON and C. ERIKSON, *Nature* **227** (1970) 491.
8. G. REINISH and A. NOWICK, *ibid.* **253** (1975) 626.
9. A. MARINO and R. BECKER, *ibid.* **253** (1975) 627.
10. S. LI and E. KATZ, in "Electrical Properties of Bone and Cartilage" edited by C. Brighton, J. Black and S. Pollack (Grune and Stratton, New York, 1979) p. 119.
11. G. HASTINGS, F. MAHMUD and M. MARTINI, *J. Biomed. Eng.* **11** (1989) 403.
12. S. POLLACK, R. SALZSTEIN and D. PIENKOWSKI, *Ferroelectrics* **60** (1984) 297.
13. R. PETHIG, in "Dielectric and Electronic Properties of Biological Materials" (Wiley and Sons, New York, 1979) pp. 15, 70, 100, 329.
14. G. REINISH and A. NOWICK, *J. Retic. Soc.* **123** (1971) 283.
15. J. KOSTERICH, K. FOSTER and S. POLLACK, *IEEE Trans. Biomed. Eng.* **BME-31** (1984) 369.

Received 11 October
and accepted 19 October 1990