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Experimental determination of short-term creep of femoral sheep bones and theoretical estimation of the dissipated energy during walking

A. G. BEZERIANOS, G. C. NIKIFORIDIS
Department of Medical Physics, School of Medicine, University of Patras, 26110 Patras, Greece

The short-term creep of ten fresh femoral sheep bones was determined through an analysis of their strain response in the time interval 10^{-2}–10^{3} s, under constant load corresponding to 100% body weight. The strain in the direction of minimum principal stress was measured by strain gauges attached at two points located on the ventral dorsal level at one-half and three-quarters of the bone's length, beginning from the knee joint. The experimental data were analysed using a three-parameter viscoelastic model consisting of a spring and a Kelvin element in series, and the material constants were calculated. The results indicate that the viscoelastic behaviour of femoral compact bone presents a transition phase with a time constant comparable with that of the normal walking period. Approximating the stress developed under walking conditions by a sinusoidal function, the corresponding strain time function and the dissipated energy due to viscoelastic mechanisms were estimated. According to our estimations the dissipated energy during normal walking conditions (speed 2.5 $\mathrm{km\,h^{-1}}$ and 60 steps $\mathrm{min^{-1}}$) was about 12%.

1. Introduction

Throughout life the femur is subjected to repeated stresses and strains during millions of cycles of mechanical loading. The repeated loadings over a specific time period constitute the local stress (or strain) histories of the bone. The importance of bone stress histories in skeletal biology has been well appreciated by some investigators who have viewed it as a central issue in skeletal biology [1–3]. On the other hand, the time-dependent mechanical properties of bone are of significant importance in the analysis of the coupling between bone and implant [4, 5]. Indeed, differences in the time behaviour of these components can produce relative motion (referred to as micromovements) in dynamic loading and may accelerate the loosening of endoprostheses [6].

Lakes and Katz [7] examined the torsional dynamic and relaxation properties of cortical bone on wet-machined specimens. Their experiments were performed at body temperature over wide time intervals from 1 to 10^5 s, and they formulated a constitution equation characterizing these data. Currey [8] analysed the long-term creep behaviour of wet bone specimens in the time interval from 2 min to 10 days. Dynamic experiments performed in bone machined specimens by Black and Korostoff [9] showed an increasing storage modulus in the frequency range 35–353 Hz. However, no particular attention had been paid to the study of the viscoelastic behaviour of bone in time intervals within the normal period ($T = 1$ s) of walking, and no attempts had been made for the extrapolation of the *in vitro* data to the *in vivo* behaviour of bone during walking. Additionally, all of the experiments were carried out on bone machined specimens which may present structural alterations, caused by the cutting, with respect to the whole bones [10].

In this study, through quasi-static experiments performed on whole bones, the short-term creep of sheep femur under compressive load was measured using strain gauges. The creep strain was determined using an appropriate analytical model, according to which the bone is considered as an isotropic two-phase viscoelastic material. Furthermore, the calculation of the dissipated energy during walking was attempted by simulating the stress developed on the bone through a sinusoidal function.

2. Materials and methods

2.1. Material handling

The creep characteristics of the left femoral bones removed from 10 killed sheep ranging in age from 24 to 36 months and weighing from 18 to 30 kg were studied. Each bone was cleaned of muscles, wrapped in saline-soaked swabs and kept frozen at $-20\,^{\circ}$C. Before testing the bones were removed from the deep freeze and allowed to thaw for 14–16 h at room temperature. During the thawing period they remained sealed in a plastic bag in order to minimize tissue dehydration [11]. This is indispensable for a good approach to the *in vivo* viscoelastic behaviour of bone, since liquids and gels in the pores of the bone influence its viscoelasticity.

2.2. Testing apparatus and procedure

Constant loading was applied through a Wolpert testing machine (TZM-Z-748) on the head of the femur along its longitudinal axis. The amplitude of the load varied from 180 to 300 N (100% body weight of the donor), which is similar to the load observed under normal walking conditions [12]. The axes of principal stresses were determined using rosettes of three strain gauges (FRA-3-2-23; TML Tokyo Sokki Kenkyjo) by static experiments with loads included in the range from 50 to 1000 N. After this determination the strain measurements were performed using rosettes of two strain gauges (FCA 6 11; TML Tokyo Sokki Kenkyjo) with a very short time response (< 10 ms for a strain gauge of length 2 cm) in a 0° and 90° arrangement. The axes of the strain gauges were in the same direction as the principal stresses.

The rosettes were glued in two points of the bone surface located on the ventral–dorsal level at one-half and three-quarters of the bone's length, measured from the knee joint (see Fig. 2, below). Particular attention was paid to the attachment and protection of the strain gauges against humidity. Cyanoacrylate adhesive (CN adhesive, Tokyo Sokki Kenkyjo) was used for pore insulation and strain gauge gluing. Finally, everything was covered with microcrystalline wax (W1, Tokyo Sokki Kenkyjo). Each strain gauge was connected to a bridge amplifier (HBM KWS 3073). A similar rosette, attached to an unloaded bone, was used as the passive resistance of the bridge. The analogue signals of the amplifiers and of the load cell of the testing machine were digitized and stored in the computer memory (HP 9825B) for further processing. For the determination of the creep curve, strain measurements for a total time of 1000 s were performed. A better evaluation of the creep was obtained by recording the data with a variable sampling rate. Thus, initially the sampling rate was one measurement per 10 ms and thereafter it was decreased since the creep phenomena were slower (see Fig. 2, below).

3. Analytical model

In an outline based on its microstructure, compact bone can be considered as a two-phase viscoelastic material [13, 14]. The behaviour of such material can be described by a rheological model of three parameters consisting of a Kelvin body connected in series with a Hooke body, where E_1 and E_2 are the elastic moduli and η_1 is the coefficient of viscosity (Fig. 1). When a constant stress σ is applied the creep strain is determined by [15]

$$\varepsilon(t) = (\sigma_0/q_0)\,[1 - (1 - p_1\lambda)\,e^{-\lambda t}] \quad (1)$$

where $\lambda = q_0/q_1$, $p_1 = \eta_1/(E_1 + E_2)$, $q_0 = E_1E_2/(E_1 + E_2)$ and $q_1 = \eta_1 E_2/(E_1 + E_2)$ and for $t \gg 1/\lambda$ the creep strain is asymptotic and given by

$$\varepsilon(t) = \sigma_0/q_0 \quad (2)$$

When an oscillating stress $\sigma(t) = \sigma_0 \cos \omega t$ of cycling frequency ω is applied to the model the strain response will be an oscillating function of the same frequency but lagging behind by a phase angle δ [16]

$$\varepsilon(t) = \varepsilon_0 \cos(\omega t - \delta) \quad \text{with} \quad \varepsilon_0 = \sigma_0/J(\omega) \quad (3)$$

where $J(\omega) = J_1(\omega) - jJ_2(\omega)$ is the complex compliance and $\tan\delta = |J_2/J_1|$ is the loss factor. J_1 and J_2 are the storage and loss compliances, respectively, which are related to the constants p_1, q_0 and p_1 by

$$J_1(\omega) = (q_0 + p_1q_1\omega^2)/(q_0^2 + q_1^2\omega^2) \quad \text{and}$$

$$J_2(\omega) = (q_1 - q_0p_1)\omega/(q_0^2 + q_1^2\omega^2) \quad (4)$$

The storage compliance J_1 is associated with the energy stored, whereas the loss compliance is associated with the energy dissipated and transformed into heat during the stress oscillation. The ratio $J_2/J_1 = \tan\delta$ is used as a parameter of the damping capacity of the material.

During one loading cycle the energy dissipated (ΔW) is [16]

$$\Delta W = \pi\sigma_0 J_2(\omega) \quad (5)$$

whereas the total energy supplied to the bone is

$$W = \varepsilon_0\sigma_0/2 \quad (6)$$

The ratio $\Delta W/W$ gives the fraction of the total mechanical energy dissipated in the bone.

According to Bergmann *et al.* [17], during the stance phase of sheep walking the resultant force acts

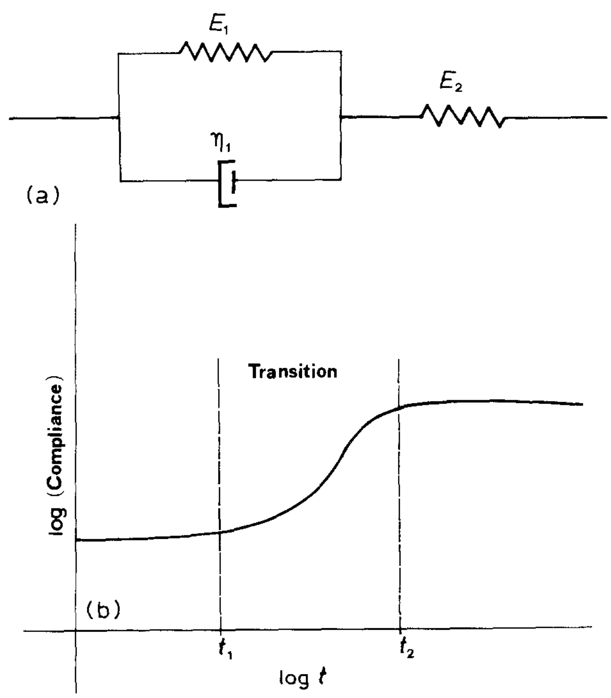

Figure 1 Viscoelastic solid model of (a) three parameters and (b) its creep compliance. The model consists of a Kelvin element, constants E_1 and η_1, and a spring of constant E_2 in series and it is capable of absorbing energy if the load is applied in time interval t_1 to t_2.

in the longitudinal direction of the femur with small anterior–posterior and medio-lateral components, and its typical values vary between 90 and 140% body weight. Thus, the following can be assumed. First, the force applied in the hip joint of the sheep during walking acts in the longitudinal axis of the femur and varies sinusoidally between zero and a maximum value. Secondly, the stress distribution developed in the transversal section of the diaphysis is homogeneous. Finally, the stress in the diaphysis during walking can be expressed as the sum of a constant and a sinusoidal component:

$$\sigma(t) = \sigma_0(1 + \cos \omega t) \qquad (7)$$

After the first seven to ten steps, when the transient phenomena disappear [18], the strain response of the bone under the aforementioned stress is determined using the Boltzmann superposition principle:

$$\varepsilon(t) = \varepsilon_1(t) + \varepsilon_2(t) \qquad (8)$$

where $\varepsilon_1(t)$ is the strain response to the constant stress σ_0 and $\varepsilon_2(t)$ is that to the sinusoidal stress $\sigma_0 \cos \omega t$.

Combining Equations 2, 3 and 8, it can be deduced that

$$\varepsilon(t) = (1/q_0)\sigma_0 + \varepsilon_0 \cos(\omega t - \delta) \qquad (9)$$

Equation 9 indicates that strain during walking is the sum of two components, the first of which is constant whereas the second is oscillating and entirely responsible for the dissipation of energy.

4. Results

In Table I the length and the cross-sectional area (S) of the examined bones in the sites a and b are given. The mean bone length was 21.1 cm, whereas the mean cross-sectional area at sites a and b was 59.7 and 56.1 mm^2, respectively. The imposed constant stress was computed assuming a homogeneous stress distribution all over the transversal section. The strain–time behaviour of the specimens was determined by fitting the experimental data according to the exact solution of a three-parameter model (Equation 1) using the least-squares technique. Typical strain–time responses in the axis of minimum principal stress for points a and b are shown in Fig. 2. The time variation of the strain in the maximum principal axis of stress was comparable with the noise, so the creep strain in this direction could not be revealed.

The experimental data are in good agreement with the strain–time response predicted by the three-parameter model (< 1% average absolute error) for both points. The values of the constant terms, concerning the viscoelastic behaviour of the ten femoral bones of our sample, were calculated on the basis of Equation 1 and are given in the fourth, fifth and sixth columns of Table I. From the comparison of the constant terms it can be deduced that there is no statistically significant difference in the time behaviour of the specimens at points a and b.

The complex compliance (J) as well as the storage and loss compliances (J_1 and J_2) and the relative dissipated energy ($\Delta W/W$) were calculated using the mean values of the viscoelastic constants ($p_1 = 0.90$ s, $q_0 = 9.7$ GPa and $q_1 = 9.8$ GPa s) of the examined specimens at both points a and b. Fig. 3 shows graphically the ratio $\Delta W/W$ versus cyclic frequency as well as the complex, storage and loss compliances. The ratio $\Delta W/W$ expresses directly the dissipated fraction of the total mechanical energy supplied. From the diagrams it is deduced that the transition regime of the tested material occurs between $\log \omega = -1$ and $\log \omega = 1$ (i.e. in the frequency range 0.016–16 Hz) and that the relative dissipated energy shows its maximum value at

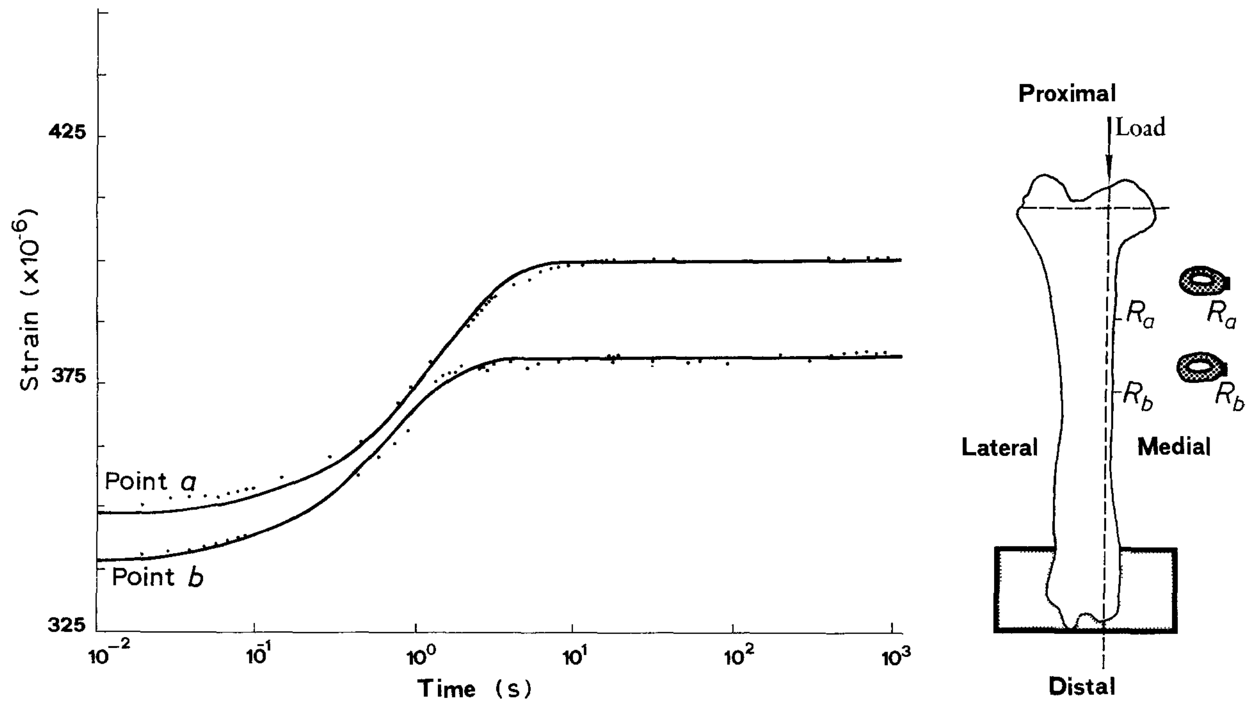

Figure 2 Experimental measurements and fitted curves (specimen 6) of creep strain for two points a and b located at one-half and three-quarters of the bone length measured from the knee joint. The inset on the right is a diagrammatic drawing of the rosette positions.

TABLE I Material properties predicted by three-parameter model from experimental strain values

Specimen no	Body weight (kg)	Bone length (m)	Cross-sectional area ($\times 10^{-6}$ m^2)		Applied stress (MPa)		p_1 (s)		q_0 (GPa)		q_1 (GPa s)	
			a	*b*	*a*	*b*	*a*	*b*	*a*	*b*	*a*	*b*
1	18	0 195	50.2	47.6	3.58	3.78	0.82	1.03	9.9	8 1	10 4	8 8
2	30	0.231	72.2	67.0	4.15	4.48	1.12	0.92	10.5	10.2	13.0	9.6
3	25.5	0 218	62.6	59.5	4.07	4.28	0.87	1.05	8.9	9.3	9 5	10 1
4	20.5	0.204	55.0	51.9	3.73	3.95	0.80	0.90	8.6	8.9	8.6	9.2
5	22	0.208	57.5	54 3	3.82	4.05	0 89	0.93	9.8	9.1	10.1	8.7
6	20	0.202	54.0	51 1	3.70	3 92	1 18	0.61	9.3	10.3	12.6	7 0
7	28	0.226	69.8	64.1	4.01	4.36	0.97	0.72	9 4	12.5	11.0	9.5
8	24.5	0 215	60.4	57 8	4.06	4.24	1.02	0.95	10.3	10.5	10 8	10 4
9	20	0.202	56.1	51.7	3.56	3.87	0 75	0.78	9.2	9.6	8.6	9.0
10	23	0.212	59.6	56.2	3.85	4.09	0.97	0.76	8.9	10.7	9.2	10.2
Mean	23.1	0 211	59 7	56.1	3.85	4 102	0.939	0.865	9.48	9.92	10.38	9.25
SD	3.8	0.011	6.9	6 0	0 21	0.23	0.14	0 142	0.63	1.22	1.52	0 99

loading frequencies encountered in normal activities. Finally, the values of the dissipated energy per volume unit and loading cycle, as well as the percentage of the total energy supplied under different walking conditions, are given in Table II. Under normal walking conditions, corresponding to a speed of 2.5 km h^{-1} or a walking rate of 60 steps min^{-1} the dissipated energy per loading cycle was calculated to be 85.5 J m^{-3} tissue, representing 11.7% of the strain energy.

TABLE II Dissipated energy and percentage of dissipated energy per loading cycle by the femoral sheep bone due to viscoelastic mechanisms under three different walking conditions

Walking conditions	Speed (km h^{-1})	Rate (steps min^{-1})	ΔW (J m^{-3})	$\Delta W/W$ (%)
Slow	1 0	34	14.3	19.5
Normal	2.5	60	85.5	11 7
Fast	5 5	100	52.0	7.15

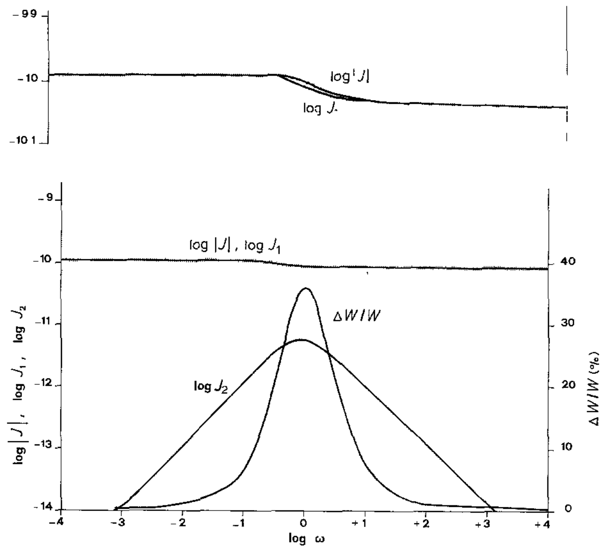

Figure 3 Graphic representations of the percentage dissipated energy ($\Delta W/W$) and the complex (J), storage (J_1) and loss (J_2) compliances versus log ω.

5. Discussion and conclusions

No statistically significant differences were observed with regard to the strain–time behaviour at two different sites of the femur diaphysis. On this basis we assume that the viscoelastic behaviour of the whole compact bone is expressed by the mean values of the viscoelastic constants (p_1, q_0 and q_1) of the examined specimens, at both points *a* and *b*. Our experimental results indicate that this behaviour presents a transition phase (Fig. 2) with a time constant $\tau = 1/\lambda = q_1/q_0$ comparable with that of the normal walking period (varying from 0.92 s for specimen 2 to 1.08 s for specimen 3). This is the reason to extend our considerations from the short-term creep of the femur to its strain response during walking.

According to the analysis of the $\Delta W/W$ versus frequency curves (Fig. 3) and the numerical data of Table II, we may assume that the dissipated energy in load-bearing bones (such as the femur) occurs in a relatively narrow frequency spectrum which definitely includes walking frequencies. These findings support the hypothesis that the rate of imparting energy to bones depends on the walking frequency. This conclusion is in accordance with the view that bone remodelling is sensitive to the rate and the amplitude of the load, as well as to the strain distribution in the whole bone [19].

Bergman *et al.* [17] measured the forces in sheep hip joints during gait, and found that the resultant joint force acts mainly in the long direction of the femur with typical maximum values between 90 and 140% of body weight. According to similar measurements in man [20] the resultant hip joint force in normal subjects, during gait, acts parallel to the ideal longitudinal axis and ranges from two to six times body weight. The differences in the load direction between sheep and man do not appear to be greater than the differences between different patients, whereas the force magnitude is smaller in sheep than in man. It must be borne in mind, however, that the bone dimensions are also smaller. Comparing stress calculations for human hip joints [21] and our estimations for the stress on sheep hip joints [5], it could be deduced that the absolute stresses are almost the same whether sheep femora are loaded with 100% body weight or human ones with 350% body weight. Therefore, our considerations concerning the sheep femoral bone could be extended to the imparting of energy in human bones during cyclic loading.

The behaviour of the bone as a composite viscoelastic material and the calculation of the relative dissipated energy during walking are the objective of further studies and must be considered in close relation to the factors controlling the function of osteocytes [22].

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