PROTECTION FROM STRESS IN BONE AND ITS EFFECTS

EXPERIMENTS WITH STAINLESS STEEL AND PLASTIC PLATES IN DOGS

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In a preliminary experiment the paired radii and femora of dogs were tested for bone mineral mass and mechanical properties including the load at break, the ultimate bending strength and the modulus of elasticity; symmetry was observed for most of the parameters determined. The influence of the elasticity of materials used for the internal splintage of bone and its relationship to bone remodelling were then investigated for stainless steel and plastic plates applied to the femora of dogs. A significant decrease in bone mineral mass per centimetre length of bone and in mechanical properties was demonstrated for the femora plated with steel, and microradiography showed that this was due to massive endosteal resorption. A model for determining the influence of protection from stress in bone is presented.

Mechanical stimulus or stress in the skeleton is probably the major factor in the maintenance of the normal balance between the dynamic processes of bone formation and resorption. When physiological stress is diminished, as in astronauts or in limbs affected by poliomyelitis, osteoporosis rapidly develops (Vaughan 1970). Protection from stress occurs if two or more components with different moduli of elasticity form one mechanical system. When a load is imposed on such a

Some investigators have suggested that in the fixation of fractured long bones by rigid plates this mechanism leads to osteoporosis, with the possibility of fatigue fracture after removal of the plate (Cochran 1969; Köbler and Wiechell 1971; Uhthoff and Dubuc 1971; Diehl and Mittelmeier 1974). Although at first the rigid fixation must favour primary bone healing, the protection from mechanical stress during the later stages of healing and during weight-bearing may lead to osteoporosis with decreased bone strength. For this reason a material with a considerably lower modulus of elasticity than that of the metals in general use, the plastic polytrifluoromethoxyethylene or PTFCE, was selected for a study of the mechanical and histological consequences of the application to cortical bone of plates made of two materials of markedly different elasticity (Table I).

In order to ensure that the results could not be influenced by asymmetry of the properties of paired bones, the mechanical parameters and the bone mineral mass of paired long bones in dogs were first compared.

INVESTIGATION OF SYMMETRY IN PAIRED BONES

MATERIAL AND METHODS

The radii and femora of seven normal mongrel dogs (weight 15 to 25 kilograms) were collected within one hour of death, paired and stored in 70 per cent ethyl alcohol for twenty-four hours and then under deep-frozen conditions at -20 degrees Celsius. Storage in alcohol has no major effect on the mechanical properties of bone (McElhaney, Fogle, Byars and Weaver 1964).

Bone mineral mass—Within five days this was estimated by means of photon absorptiometry (Cameron, Grant and McGregor 1962). From measurements at a site five millimetres

TABLE I

SOME MECHANICAL DATA OF METALS USED FOR IMPLANTS OF CORTICAL BONE AND OF PTFCE EXPRESSED IN MEGANEWTONS PER SQUARE METRE

	Modulus of elasticity (MN/m ²)	Ultimate tensile strength = UTS (MN/m ²)	Fatigue limit as percentage of UTS
316 stainless steel .	200,000	960	45
Cast Vitallium .	210,000	650	43
Wrought Vitallium	220,000	900	52
Titanium (T 130) .	110,000	400	50
Cortical bone .	6,000–20,000	80–150	25–40
PTFCE .	1,200	37	

system the component with the higher modulus bears more of the load and so protects the other component. This situation is present when a rigid metal plate is fixed to a long bone. The protection from stress occurs because the modulus of elasticity of cortical bone is less than a tenth of that of stainless steel (Table I).

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proximal to the midpoint of the bone, the bone mineral mass in grams per centimetre length of bone could be calculated. The technique of the scanning and the calculations used have been described by Rösingh, Hart, van der Schoot and Bon-Nijssen (1975). When the mechanical test was done a ring of bone was cut out at the site of the scanning for measuring the cross-sectional area. The bone ring was washed thoroughly with distilled water and then ground parallel to a thickness of 3.0 ± 0.1 millimetres on SIC paper (grit 300). The cross-sectional area was calculated by registering the shadow area of the ring on a lux-meter (Tonino 1974). The bone mineral mass in grams per cubic centimetre of bone is the ratio of the mineral mass per centimetre length of bone to the cross-sectional area.

Mechanical parameters—The bones were tested at room temperature in a Hounsfield tensometer by bending in three-point loading, using rounded pivots (Fig. 1). The deformation was at a constant speed of 6.25 millimetres per minute. The active length of the bone was 100 millimetres. The deformation as well as the load were registered simultaneously on a Honeywell X-Y recorder, using electronic displacement transducers. The test pieces were kept wet during each experiment, which usually lasted ten to fifteen seconds. With a vernier caliper the width and height of the bones were measured before testing at the midpoint and at a distance of five millimetres on each side of it. After the bones had broken, the internal dimensions were recorded at the same levels.

The features determined and calculated were: the load at break (F_b), the ultimate bending strength (UBS) and the

modulus of elasticity (E) using the formulae for whole bones (Yamada 1970):

$$\text{UBS in MN/m}^2 = 8F_bHL / \pi (BH^3 - bh^3)$$

$$E \text{ in MN/m}^2 = 4F_pL^3 / 3\pi d_p (BH^3 - bh^3)$$

These formulae assume a linear distribution of stress across the section of the bone, but as Swanson (1971) and Burstein, Currey, Frankel and Reilly (1972) have pointed out, this assumption is not totally valid. However, as our main aim was to show differences within the pairs rather than to measure absolute values, that argument is not relevant to the results of this experiment.

Student's t-test for the mean of left and right (t_m) and for the difference of paired observations (t_{po}) was used.

RESULTS

The bone mineral mass per centimetre length of bone is given for all pairs of bones in Table II. Means and standard deviations of the right and left radii and femora are shown, as well as the mean and standard deviation of the difference. Observations on the bone mineral mass per cubic centimetre of bone are given in Table III. No significant differences in bone mineral mass were found between the paired specimens except for the bone mineral mass per centimetre bone length of the paired radii.

Measurements of the breaking load (F_b), the ultimate bending strength (UBS) and the modulus of elasticity (E) are given for the paired radii in Table IV and for the paired femora in Table V. Means and standard deviations are shown, as is the mean difference with standard deviation. The modulus of elasticity of the paired radii and femora from one dog could not be calculated for technical reasons. No significant differences in mechanical parameters were found between the paired specimens, except for the ultimate bending strength of the paired femora.

Comment

In experiments using quadrupeds where the influence of some kind of treatment on a long bone is studied, it is common practice to use the unaltered member of the opposite side as a control. However, only in a few experiments has the

FIG. 1

Detail of the Hounsfield tensometer. The femur is bending under three-point loading. Note the line of exit screw holes down the medial aspect of the shaft.

TABLE II
BONE MASS/CM BONE LENGTH ESTIMATED BY PHOTON
ABSORPTIOMETRY

	n	Right	Left	t_m	$\bar{d}$	t_{po}
Radius	7	0.6334	0.6543	0.3925	0.0210	2.5760
Sd		0.1085	0.0899	n.s.	0.0215	$p \leq 0.05$
Femur	7	0.8901	0.8875	0.0348	0.0027	0.2120
Sd		0.1383	0.1412	n.s.	0.0128	n.s.

Student's t-test for the mean of left and right (t_m) and for paired observations (t_{po}) is applied.

TABLE III
BONE MASS/CM² OF BONE ESTIMATED BY PHOTON ABSORPTIOMETRY

	n	Right	Left	t_m	$\bar{d}$	t_{po}
Radius	7	0.9089	0.9413	0.3151	0.0324	1.3715
Sd		0.1819	0.2025	n.s.	0.0631	n.s.
Femur	7	1.0172	1.0159	0.0163	0.0010	0.0360
Sd		0.1637	0.1337	n.s.	0.0286	n.s.

TABLE IV
OBSERVATIONS ON PAIRED RADII

	n	Right	Left	t _m	$\bar{d}$	t _{po}
F _b (N)	11	997.03	1002.78	0.0470	6.57	0.3072
Sd		286.32	287.99	n.s.	70.9	n.s.
UBS (MN/m ²)	11	198.87	195.45	0.2259	3.4	0.8047
Sd		35.34	35.25	n.s.	14.0	n.s.
E (MN/m ²)	10	11611.5	11306.6	0.2493	328.0	1.5044
Sd		2653.8	2783.8	n.s.	690.0	n.s.

Means and standard deviations for the load at break (F_b in Newtons), the ultimate bending strength (UBS in megaNewtons per square metre) and modulus of elasticity (E in megaNewtons per square metre) of eleven pairs of radii are given. Student's t-test for the mean of left and right (t_m) and for paired observations (t_{po}) is applied.

TABLE V
OBSERVATIONS ON PAIRED FEMORA

	n	Right	Left	t _m	$\bar{d}$	t _{po}
F _b (N)	13	1763.24	1742.04	0.1201	108.51	0.7057
Sd		443.81	455.96	n.s.	30.10	n.s.
UBS (MN/m ²)	13	184.45	177.69	0.4749	6.75	2.3369
Sd		37.33	35.21	n.s.	10.42	p ≤ 0.05
E (MN/m ²)	12	9727.3	9418.5	0.2836	308.7	1.4854
Sd		2867.9	2473.2	n.s.	720.0	n.s.

Means and standard deviations for the load at break (F_b), the ultimate bending strength (UBS) and modulus of elasticity (E) of thirteen pairs of femora are given.

required symmetry been established. Mather (1967) showed that the average load needed to break twenty-eight left human femora did not differ significantly from that needed to break the right femora, but the average energy-absorption capacity of failure was significantly higher for the left femora. Henry, Freeman and Swanson (1968) tested nineteen pairs of rabbit radii in respect of bending strength and modulus of elasticity: the difference between the two strengths was expressed as a percentage of the mean of these values. For all pairs the mean of these percentages was 5.7 per cent, with a standard deviation of 5.6 per cent. The corresponding figures for the modulus of elasticity were 5.1 per cent and 5.4 per cent. On bending twenty-three pairs of right and left rabbit metatarsals Currey (1969) found a positive correlation between the two sides for the load causing failure in static loading, the ultimate stress and the modulus of elasticity. Also Puhl, Piotrowski and Enneking (1972) and White, Panjabi and Hardey (1973) showed the symmetry of the mechanical parameters in torsional loading of respectively paired canine fibulae and rabbit limbs. Despite considerable variation within the pairs, no pattern of right or left dominance emerged. These observations support a theory of symmetry. However, on weighing the bones of paired limbs, Dogra and Singh (1971) and Singh (1971) found a statistically significant one-sided dominance. On this basis it is reasonable to suppose that the amount of mineral would also show a one-sided dominance.

In the experiments described here dogs of different breeds and sizes were used and large differences in the several parameters were found. When one compares only the means of right and left, as most authors do (Mather 1967; Currey 1969), there is no significant difference for all the observed para-

eters. But because of the heterogeneity of biomaterials it seems more valuable to compare the measurements in the same animal by means of Student's t-test for the difference of paired observations. No significant difference was found for the parameters studied as shown by the calculation of t-values for paired observations (t_{po}), except for the bone mineral mass per centimetre length of bone of the radii (which was 3.2 per cent higher for the left side) and for the ultimate bending strength of the femora (which was 3.7 per cent higher on the right side). This means that there is a very high degree of symmetry in mechanical properties and mineral mass between the left and right long bones of a dog. As White *et al.* (1974) stated, one can make use of this quality of symmetry in planning biomechanical experiments, where the variable can be studied in one long bone while the paired bone is used as a control.

The values found for the bending strength and modulus of elasticity are in agreement with other reports (Currey 1969; Yamada 1970).

INVESTIGATION OF PROTECTION AGAINST STRESS

MATERIAL AND METHODS

In six mongrel dogs (weight 20 to 30 kilograms) one femur was plated with a six-hole AO plate made of stainless steel (No. 223.06) and the other with a similar plate made of PTFCE. Before operation all plates were weighed on an analytical balance.

FIG. 2

Dog No. 537. A radiograph showing the AO plate on the right femur and the PTFCE plate on the left.

Under general anaesthesia and aseptic conditions the lateral aspect of the femur was approached dorsal to vastus lateralis and the periosteum was stripped off. After drilling and tapping of both cortices with standard AO instruments, the AO plate was fixed with AO cortical screws. All screws were inserted with a tension-calibrated screw-driver, the torque applied being always 1.96 metre Newtons (≈ 20 kg_fcm). After haemostasis and rinsing of the wound with saline, the subcutaneous tissue and the skin were closed with catgut and

nylon respectively. The opposite femur was treated in the same way but with the plastic plate (Fig. 2). The next day the dogs were released and could walk normally. They were killed after ten, twelve, fourteen, sixteen (two dogs) and

mineral mass was determined by means of photon absorptiometry and the mechanical parameters were measured.

Microradiographs were made by the technique of Jowsey, Kelly, Riggs, Bianco, Scholz and Gershon-Cohen (1965). At

TABLE VI
THE TORQUE NEEDED TO RELEASE THE SIX SCREWS IN
METRE NEWTONS (N.M)

		Range	Total	Average
H 537	AO	0.196 to 1.764	5.488	0.915
10 weeks	PTFCE	0.784 to 1.176	5.292	0.885
H 536	AO	0.392 to 1.176	4.508	0.752
12 weeks	PTFCE	0.588 to 0.784	4.508	0.752
H 543	AO	0.392 to 1.764	5.488	0.915
14 weeks	PTFCE	0.588 to 1.372	6.076	1.013
H 533	AO	0.588 to 1.764	7.056	1.176
16 weeks	PTFCE	0.784 to 0.980	5.292	0.885
H 535	AO	0.588 to 1.176	5.096	0.849
16 weeks	PTFCE	0.588 to 1.568	5.880	0.980
H 534	AO	0.588 to 1.568	6.076	1.013
18 weeks	PTFCE	0.784 to 0.980	5.488	0.915

TABLE VII
DETAILS OF BONE MINERAL MASS

	n	AO	PTFCE	t _m	$\bar{d}$	t _{po}
Bone mineral mass/cm of bone	6	0.8545	1.0229	2.0375	0.168	4.1588
Sd		0.154	0.131	p ≤ 0.05	0.099	p ≤ 0.05
Bone mineral mass/cm ³ of bone	6	0.8826	0.9896	1.0282	0.107	1.9075
Sd		0.188	0.172	n.s.	0.137	n.s.

The mean, standard deviation and difference of paired observations ($\bar{d}$) of the bone mineral mass per cm and per cm³ of bone is given for the AO plated and PTFCE plated femora. Only the difference in bone mineral mass per cm length of bone is significant (p ≤ 0.05).

TABLE VIII
DETAILS OF THE AO PLATED AND PTFCE PLATED FEMORA

(n=6)	AO	PTFCE	t _m	$\bar{d}$	t _{po}
F _b (N)	1338.58	1911.23	3.6247	572.58	6.6795
Sd	321.04	216.09	p ≤ 0.01	209.98	p ≤ 0.05
UBS (MN/m ²)	130.19	165.82	1.5341	35.61	4.3988
Sd	41.49	38.84	n.s.	19.83	p ≤ 0.05
E (MN/m ²)	5846.01	7304.02	1.0895	1458.0	3.8872
Sd	2282.86	2251.11	n.s.	918.8	p ≤ 0.05

The means, standard deviations and difference ($\bar{d}$) of paired observations of the load at break (F_b), the ultimate bending strength (UBS) and modulus of elasticity (E) are given for the AO plated and PTFCE plated femora. By Student's t-test for paired observations the difference was significant for all observed mechanical parameters (p ≤ 0.05).

eighteen weeks and the femora were disarticulated. The screws were released with the tension-calibrated screw-driver, the torque required being measured (Table VI). The plates were again weighed. The bones were fixed in 70 per cent ethyl alcohol for twenty-four hours, after which the bone

a point 6.5 millimetres proximal to the midpoint of the plated area a ring 1 millimetre thick was cut. After cutting sections of 0.1 millimetre, microradiography was performed on a Kodak 649-0 plate (20 kilovolts; 20 milliamperes; 10 minutes).

RESULTS

No infection was observed up to the time of death. The average torque needed to release the screws from the femora was 0.92 metre Newtons (Table VI). No significant difference in torque was noted between the screws of the AO plates and those of the PTFCE plates. This observation is very important, because if a difference were present the two limbs could not be compared further as representing the same kind of mechanical system. The weights of

plated femur there were areas of extensive subendosteal trabeculation (Fig. 3). There was evidence of new bone formation in these enormous resorption cavities, but edge sclerosis indicated that this had already ceased in a number of the cavities (Fig. 4). In the PTFCE plated femur the resorption cavities were much smaller and of normal size (Figs. 3 and 5), the lamellae generally were thicker and there was some formation of woven bone beneath the plate. The same kind of woven bone, but less of it, could be seen in the AO plated femur.

FIG. 3

Dog No. 533. Microradiographs of transverse sections of the femora made sixteen weeks after the application of lateral plates between the arrows. ($\times 4$.)

all the plates were the same before operation and after removal.

Measurements of the bone mineral mass per centimetre and per cubic centimetre of bone are given for all pairs of bones in Table VII. Means and standard deviations of the AO and PTFCE plated femora are shown, as well as the mean value and standard deviation of the difference. There was a significant decrease in bone mineral mass per centimetre of the AO plated femora, for the mean value, and for the difference of the paired observations ($P \leq 0.05$). The difference in bone mineral mass per cubic centimetre was not significant (Table VII).

The mechanical findings are presented in Table VIII. All AO plated femora showed much lower mechanical properties than the PTFCE plated femora; for paired observations the decrease was significant at the $P \leq 0.05$ level for all the mechanical parameters measured.

Histologically, great differences were seen between the bones plated with AO steel plates and those plated with PTFCE. As an example the microradiographs of the femoral diaphyses of a dog are shown, sixteen weeks after the application of the plates. They present very clearly what was observed in all the bones. In the AO

DISCUSSION

Because it was established in the first group of experiments that there is symmetry in the mechanical properties and mineral mass of paired long bones in dogs, it must be concluded that the differences in mineral mass, mechanical parameters and histological pictures observed between the femora with steel plates and the femora with plastic plates, were entirely due to the different mechanical properties of the two materials. It must be emphasised that not only the modulus of elasticity but also the thickness and length of the plate relative to the bone influence the amount of protection from stress. Diehl and Mittelmeier (1974), performing bending tests on human tibiae after steel plates had been removed, found an average loss of function of one-third. This figure is in agreement with the decrease of 30 per cent in load at break which we found for the femora to which AO plates had been fixed.

The significant decrease in bone mineral mass per centimetre length of the AO-plated femora is explained by the massive resorption and by the fact that the newly formed bone in the resorption cavities was still not fully

mineralised (Fig. 4). The fact that the difference in mineral mass per cubic centimetre was not significant means that the cross-sectional area of the diaphysis of the bones with AO plates must also have been diminished, for it was not possible to measure the influence of the resorption cavities on the cross-sectional area because of the thickness of the bone rings. Therefore widening of the medullary cavity or periosteal resorption or both must have taken place in those femora: indeed both processes were observed in all cases in the microradiographs. This confirms the findings of Uthoff and Dubuc

by the findings in the microradiographs. It may be assumed that after a certain period of time the processes of massive resorption and new bone formation will be in balance again, but any such time-dependent relationship requires further investigation. In contrast with other studies (Uthoff and Dubuc 1972; Diehl and Mittelmeier 1974), where the formation of resorption cavities was related to the area beneath the plate, these experiments showed resorption mainly on the anterior and posterior aspects of femora plated laterally. This observation may be explained on a mechanical basis; if the metal plate is

FIG. 4
Detail of the diaphysis of the AO plated femur from Figure 3 above. Large resorption cavities, mostly past the active stage, and young Haversian systems are seen. The arrows indicate edge sclerosis in a few places. ($\times 80$.)

FIG. 5
Detail of the diaphysis of the PTFCE plated femur from Figure 3 above. The resorption cavities are small and the young Haversian systems are of normal size. The arrows point to areas of edge sclerosis. ($\times 80$.)

(1971) who showed three important changes in the structure of compact bone following compression fixation with metal plates: cortical thinning, cavitation of the cortex and the presence of woven bone.

As Currey (1969) has pointed out, there exists a relationship, possibly linear, between the ultimate bending strength and modulus of elasticity on the one hand and the mineral content of bone on the other. The observed decrease in mechanical properties of the femora with AO plates is in agreement with the observed decrease in bone mineral mass. Other experiments, such as those in which a nerve was cut (Gillespie 1954) or one limb was immobilised in plaster (Haike, Heymann and Wagner 1966), show the same decrease in mechanical properties of the treated limb, but because of the unknown influence of these manipulations on muscle and other soft tissues, these experiments are only of limited value in relation to protection from stress.

The observed mechanical differences between the PTFCE plated and AO plated femora are also explained

lateral, as in these experiments, the stress and the strain in the bone are diminished mainly in the anterior and posterior cortices.

The results of this study may have some clinical significance in relation to the appropriate time for removal of metal plates after osteosynthesis. Müller, Allgöwer and Willenegger (1965) advocated removal after one and a half to two years, but the present findings suggest that this should be done much earlier, as soon as there is bony union.

In conclusion, these experiments show that the phenomenon of protection from stress in a plated bone does exist and can be of paramount importance to the strength of the bone. After removal of a metal plate fatigue fracture can and does occur as a consequence of diminished bone strength (Köbler and Wiechell 1971; Uthoff and Dubuc 1971). How the reduced mechanical stress on the bone is transformed into the biological stimulus of increased resorption is not yet known, but the theory of piezo-electricity may offer an answer (Bassett 1968).

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