Fatigue Behavior of the Equine Third Metacarpus: Mechanical Property Analysis

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Summary: This is the first in a series of experiments to study the fatigue properties of equine cannon (third metacarpal) bone specimens from Thoroughbred racehorses. Monotonic and fatigue tests to failure were performed in four-point bending on diaphyseal specimens in a 37°C saline bath to answer three initial questions. (a) Will a linear variable differential transducer yield the same elastic modulus as strain gauges? (b) Will fatigue results depend on whether the periosteal or endosteal side of the beam is in tension? (c) Are there regional variations in the monotonic and fatigue properties of the cannon bone midshaft? Eighteen left-right pairs of specimens from six horses were used. One beam of each pair was fitted with strain gauges. Fatigue tests were conducted on 24 specimens under load control at 2 Hz: an initial range of 0-10,000 microstrain was used so as to produce failure in a reasonable period of time. There were no left-right differences in the fatigue or monotonic properties, and the presence of a gauge had no effect on modulus measured by a linear variable differential transducer. However, gauge-measured moduli were about 1 GPa less than transducer-measured values. Fatigue life was independent of which side of the beam was in tension, and there were significant variations in mechanical properties around the cortex. The lateral region was stiffer than the dorsal region but the latter had a longer fatigue life. The fixed cylindrical supports used in this experiment eventually produced slight wear grooves, causing artifactual stiffening at the end of the load cycle in some

specimens. A second experiment using roller supports confirmed the reason for this stiffening. It also showed that fatigue life was shorter when roller supports were used but regional differences were similar.

Pathology related to fatigue damage in bone is an important problem in athletes and military trainees (12). Stress fractures are even more problematic among racehorses. Approximately 70% of young Thoroughbred racehorses are victims of stress fracture or "bucked shins," a painful inflammation of the third metacarpal bone due to a fatigue response (18). The third metacarpal or "cannon" bone is the most frequent site of such injuries.

Fatigue failure of bone from species other than horses has been extensively studied (2-5,8,9,13,24,25). Studies of human and bovine bone have shown that the number of cycles to failure, N_{FALL} , is related to the applied stress (or strain) range. As, by a power law

$$N_{FALL} = k \Delta s^q \quad (1)$$

where k and q are variables that depend on the type of bone, the method of loading, and whether s is stress or strain (8). The exponent q is quite large, in the range of 5 to 15. Therefore, small changes in load or strain magnitude greatly diminish bone's fatigue life. Most vertebrates produce maximal bone strains of 2,000-3,000 microstrain (23); however, strains in young Thoroughbred racehorses range from 4,400 to 5,600 microstrain (19). If $q = 5$, the low end of its observed range for bone tested *in vitro*, the fatigue life (number of cycles to failure) at 4,500 microstrain is only 13% of that for a strain range of 3,000 microstrain. Higher q values result in a greater difference in fatigue life. Thus, it seems that racehorses can

sustain a fatigue fracture much more quickly than other animals simply because they generate higher peak strains. Although repair of fatigue damage by remodeling would obviously enter this picture, this simple calculation is nonetheless in concert with the high prevalence of stress fractures in Thoroughbred racehorses.

Our long-term goal is to elucidate the relationships between two forms of fatigue damage: microcracks (1,17) and modulus reduction (5). Preliminary experiments showed that loading similar to that producing isolated microcracks in canine bone does not produce cracks in equine bone. We therefore decided to test the hypothesis that equine specimens would not exhibit isolated microcracks even after fatigue failure. Given the reported sensitivity of fatigue life to loading frequency (13,14), we used super-physiologic strains to produce failure in a reasonable time.

Several other important research questions arose, which are the subject of this paper (with the results for crack damage to be reported elsewhere). (a) It would be easier to obtain modulus data from linear variable differential transducers than strain gauges. We therefore sought to compare modulus data obtained both ways and to determine whether gluing gauges on the specimen would affect the fatigue results. (b) We were concerned that our results would depend on whether the periosteal or endosteal side of the beam was in tension, so we tested for that. (c) We wanted to find out whether fatigue life, monotonic strength, or the elastic modulus varies regionally within the diaphysis of the cannon bone. In the course of analyzing the data from this experiment, we found that wear at the fixed four-point bending supports influenced the behavior of the specimens. We therefore

conducted a second experiment to test the hypothesis that roller supports would prevent this effect and to determine how it had affected modulus and fatigue life.

METHODS

Specimens were taken from the mid-diaphysis of six pairs of equine cannon bones. The bones were obtained fresh from necroptied Thoroughbred racehorses. There were five females and one castrated male: two were in race training and four had raced numerous times. Their ages ranged from 2 to 5 years (mean, 3.7 years). All bones were examined grossly to avoid any with evidence of acute periosteal response, and were stored frozen at -4°C .

Three rectangular beams *were* machined from each bone so that their long axes corresponded to that of the bone. A band saw was used to isolate the mid-diaphysis and separate the medial, lateral, and dorsal regions. The beams were then milled to a final size of 10 x 4 x 100 mm using continuous irrigation with normal saline. In all cases, one 10 x 100 mm aspect of the beam faced the periosteal surface and was marked by beveling one corner. The actual dimensions of each beam were measured to the nearest 0.01 mm for use in subsequent calculations. Beams from each horse were assigned randomly **to** three test groups such that each group contained an equal number of beams from each horse, region, and side (i.e., left or right). Prior to storage, a longitudinally oriented uniaxial strain gauge was applied to the top or bottom surface (or both) of one beam from each contralateral pair, using M-Bond 200 adhesive (Measurements Group, Raleigh, NC, U.S.A.). Prepared beam specimens were stored frozen at -4°C and then equilibrated in normal saline at 37°C before testing. Some strain gauges partially de-bonded and were removed and replaced immediately prior to testing. Testing was performed in a bath of circulating normal saline at

37°C on a model 809.10 MTS servohydraulic testing machine running under Test ware SX software (MTS, Eden Prairie, MN, U.S.A.). on a 386 computer (Compaq, Houston, TX, U.S.A.). Testing was performed in accordance with the four-point bending test guidelines outlined in American Society for Testing and Materials Standard D790M-82. The inner and outer supports were 32 and 64 mm apart, respectively. All were fixed stainless-steel cylinders. 9.53 mm in diameter. Maximum beam deflection was measured at the beam center with a linear variable differential transducer (model DC-E 125: Schaevitz Engineering, Pennsauken, NJ, USA) rigidly mounted to the outer support fixture.

There were 12 beams in each of three experimental groups in the first experiment. Group M was tested monotonically to failure at a displacement rate of 1 mm/sec with the periosteal side of each beam in tension. Elastic modulus was determined by linear regression of the linear region of the load-deflection curve between 20 and 200 N. Failure load and deflection values were taken where the load values were maximum.

The fatigue tests were divided into two groups, P (periosteal) and F (endosteal), performed with the periosteal and endosteal sides loaded in tension. Testing was performed under load control at a frequency of 2 Hz with a sinusoidal waveform. The maximum load to be applied to each beam was determined by loading the beam to the deflection associated with 10,000 microstrain, as calculated from beam theory, and noting the corresponding load. The specimen was then cycled between 10 N and that load until failure (defined by fracture) occurred, and the number of cycles to failure was recorded. Data on load, deflection, and from the strain gauges, when they were present, were acquired for the first 100 cycles of each fatigue test. Subsequently, two cycles of such data were **recorded** at logarithmically spaced intervals (i.e.,

every 100 cycles for the first 1,000 cycles, every 1,000 cycles for the first 10,000 cycles. and so on) until failure. The data acquisition rate was 200 Hz.

The nature and location of each failure site were recorded. The initial elastic (tangent) modulus was calculated for the monotonic tests and the first cycle of the fatigue tests. For the cyclic tests, both tangent modulus and secant modulus were calculated for the recorded cycles in order to monitor the accumulation of mechanical damage.

The slope, m_0 , of each recorded load-deflection curve was calculated with use of linear regression over that portion *between 100 N* and approximately 20% of the maximum load. In all cases, tangent modulus (E) was calculated using m_0 and the equation

$$E_{LVD} = 11 L^3 m_0 / 464 b d^3 \quad (2)$$

For those specimens that were fitted with strain gauges, the slope of the load-strain curve, m_s , was similarly calculated and the gauge-measured modulus ($E_{en(i)}$) was determined from the equation

$$E_{GA} = 31, m_s / (4 b d^3) \quad (3)$$

where L = outer support span, b = beam width, and d = beam thickness. The secant modulus was calculated using Eq. 2 with m_n equal to the point-to-point slope from the minimum to maximum load in the cycle.

In the second experiment the methods were the same. with the following exceptions. A single group of 18 beams was machined from the dorsal, medial, and lateral cortices of the left cannon bones of six additional horses (age range, 3-5 years and mean age 4.0 years: three male. three female: one in race training and five racing). No strain gauges were applied to any of these beams. They were tested in fatigue in the same manner as Group P in the first

experiment except that the supports were replaced with 9.53 mm diameter cylinders that were supported by ball bearings and thus were free to rotate. Finally, the periodic data collection program was changed so that modulus was averaged over nine cycles at each sampling point, and the last 20 cycles before failure were recorded also.

The results are reported as mean $\pm$ SD. Values for bone fatigue life are not normally distributed (11); therefore, medians are also reported for this variable, and these data were log-transformed for statistical analysis. All but two of the beams failed before 8,000 cycles. One test was stopped at 20,000 cycles. Mixed model analyses of variance (ANOVAs) containing the independent variables of horse, leg (left or right), region (dorsal, lateral, or medial), tension side (periosteal or endosteal), presence of strain gauges (yes or no), and support (fixed or roller), and as many interactions as the analysis would support, were used to evaluate these effects on fatigue life and initial modulus (OW procedure; SAS, Cary, NC, U.S.A.). Horse and horse interaction terms were included as random effects. Terms were deleted from the model on the basis of high p values, and error terms used to test each hypothesis were constructed either with the random statement or with specific terms that included the horse interaction. A split plot analysis was used to compare fixture supports, with horse the main plot, fixture the main plot effect, and other terms as subplot effects. *Post hoc* comparisons were evaluated by specific contrast statements or least square means using specific error terms for the parent analysis. Significance was set at $p \leq 0.05$. [Figure 1]

RESULTS

Analysis of the data from the first experiment allowed us to reach a number of conclusions with regard to the effects of strain gauges on the measurement of elastic modulus and fatigue life (Table 1). When the monotonic and fatigue tests in the first experiment were combined, a paired *t* test showed that the initial elastic modulus calculated using strain gauge data was

significantly less than that obtained from linear variable differential transducer data ($p < 0.0005$). The regression equation relating them was

$$E_{\text{LVDT}} = 2.19 + 0.933 E_{\text{GAGE}} \quad (4)$$

with $R^2 = 0.83, p < 0.0001$. On the other hand, ANOVA showed that the presence of a gauge had no effect on ELVDT. (Hereafter, "initial modulus" will refer to Et.VDT•)

A paired t test showed that initial modulus was similar in left-right pairs of beams (Table 1). Similarly, there were no significant differences in monotonic maximum bending load or maximum beam deflection, or in fatigue life, between the specimens with or without strain gauges, or between left and right specimens.

With regard to the question of beam orientation during fatigue testing, fatigue life was not affected by testing with the periosteal side in tension as compared with the endosteal side in tension ($p > 0.64$). Initial elastic modulus was ($p = 0.0046$) or was not quite ($p = 0.059$) significantly affected, depending on how the ANOVA error term was selected.

Before proceeding to the question of regional differences, it is convenient to digress now to the issue of fixed supports compared with roller supports. Initially, working with fixed support data from the first experiment, we normalized elastic (tangent) and secant modulus values measured over the fatigue life to the corresponding initial value for each specimen. The tangent modulus generally decreased over the fatigue life in all three anatomic regions, and the secant modulus of the lateral specimens behaved similarly. However, the secant modulus of the dorsal specimens consistently increased after an initial reduction (Fig. 1). Medial specimens exhibited mixed secant modulus behavior. (Because data were collected only periodically, these curves do not necessarily show the extent of modulus changes just before failure.)

This unusual behavior of the secant modulus was found to be associated with very shallow grooves worn into the surfaces of the beams by the fixed supports as the beams were repeatedly bent. When the supports contacted the end of their wear grooves at maximum load, they applied longitudinal as well as perpendicular loads to the beam. Since the beams were stiffer in this mode of loading, the slope of the load-deformation curve reflected this interaction by curving upward at the end of the load cycle, increasing the secant modulus. We have documented these effects of wear grooves using physical and finite element models.

The second experiment tested the hypothesis that such effects would not occur if the supports rolled, preventing wear. This was indeed the case. The grooves were essentially eliminated, and plots of secant modulus versus fatigue life now curved downward rather than upward near the failure point (Fig. 2). Roller supports *reduced* fatigue life ($p = 0.0142$) (Table 2), but initial modulus was not affected by the support change ($p = 0.336$). [Figure 1]

Returning now to the question of regional differences, the ANOVA model for the two experiments combined showed that both initial modulus ($p = 0.0001$) and fatigue life ($p = 0.0003$) were affected by region, regardless of which supports were used (Tables 1 and 2). The initial modulus was significantly different in all three regions, being greatest laterally and least dorsally. Fatigue life was significantly different in the dorsal and lateral regions, and in the medial and lateral regions, but there was not a significant difference between the dorsal and medial regions. The differences in fatigue life were the reverse of those for initial modulus: the dorsal region had the greatest fatigue resistance and the lateral region the least.

The monotonic tests showed that there were statistically significant regional differences in maximum bending load. All three regions were different from one another, with lateral specimens strongest and dorsal specimens weakest. No significant differences among regions were found for the deflection at failure load.

Finally, we note an interesting aspect of our data in relation to Eq. 1. This equation is based on experiments in which applied stress or strain is varied over a wide range and the resulting fatigue life is measured. Our experiments looked at fatigue life for one particular initial strain range: 10,000 microstrain. However, due to the variability in initial elastic modulus of the specimens, the stress ranges applied under load control varied among the specimens. When we plotted fatigue life versus applied stress range, we found that Eq. 1 fit the data quite well for both fixed and roller supports (Fig. 3). This result suggests that fatigue life is a function of both stress and strain. Furthermore, the values of q obtained from regression analysis (6.8 for fixed supports and 4.7 for roller supports) fell within the lower portion of the range (5-15) for experiments in which applied stress (or strain) was varied widely. [Table 2]

DISCUSSION

The answers to the research questions posed at the beginning of this paper are as follows. First, elastic modulus measured using linear variable differential transducers was different from that measured using strain gauges. but the application of gauges had no discernible effect on initial modulus or fatigue life. Second, fatigue life did not depend on which side of the beam (periosteal or endosteal) was in tension, hut we remain uncertain about the effect on initial modulus. Finally, fatigue life, monotonic strength, and elastic modulus all varied regionally within the diaphysis of the cannon bone. Dorsal specimens were less stiff and strong than

lateral specimens but had longer fatigue lives. We also learned that cyclic loading in four-point bending produced wear grooves at fixed supports but not at roller supports, and the two methods of loading resulted in different fatigue lives. [Figure 2] [Figure 3]

There are several strengths to our study. We studied the fatigue properties of equine cannon bone tissue at a physiologic load repetition rate (2 Hz) for the first time, to our knowledge, previous work having been done at 40 Hz (20). We worked out several details concerning the fatigue testing of *bone* in four-point bending, including the importance of roller supports in preventing artifacts due to wear grooves at fixed supports. Regional variations have been demonstrated in the mechanical properties of the cannon bone, with the lateral region being strongest and stiffest but the dorsal region having the best fatigue resistance. Thus, there seems to be a trade-off between monotonic rigidity and fatigue resistance.

There are also a number of limitations to our study. One is that the applied strain ranges employed were significantly higher than those measured *in vivo* by Nunamaker et al. (20) for the cannon bones of Thoroughbred racehorses running at race speeds. We explained above why we loaded at this strain range. While it is twice the strain magnitude normally experienced in the cannon bone, it is comparable with that used in early human and bovine bone fatigue studies, which were similarly super-physiologic relative to normal strains (3,540). Studies now in progress will investigate the fatigue behavior of equine bone for other strain ranges.

Another limitation is that the stresses in the experimental beams were different from those that each particular region of tissue would experience in intact, functioning bone. The direction of the ground reaction force during gait, and strain gauge data, indicate that the largest bending moments in the cannon bone are dorsopalmar (19). Our dorsal specimens were bent in this manner, but the medial and lateral beams were bent mediolaterally. This mismatch between

experimental loading and that for which the bone is presumably adapted could have contributed to the regional differences. However, had we machined the specimens to match the putative physiologic loading direction, the histologic layering within them would have been parallel to the neutral plane in the dorsal beams and perpendicular to it in the beams from the other two regions. This seemed to be an equally serious problem. It would not be practical to match the loading in the beams to that normally incumbent on the material in the running horse, because this loading is unclear. This problem is common to many studies of bone's mechanical properties. Since bending is a common mode of clinical fracture, and many fractures are thought to initiate on the tensile side of bent bones, we chose four-point bending to obtain a relevant yet relatively simple stress state requiring simple fixtures.

The regional differences in mechanical properties reported here are consistent with data on compositional and mechanical properties reported elsewhere for equine cannon bone. Results of static compression tests (15) have indicated a trend toward a weaker, less stiff dorsal region as compared with the more lateral and medial areas. Sutthipong et al. (27) reported statistically significant regional differences in porosity and radiographic mineral density. The most dorsal region exhibited the highest porosity and mineral density, and the lateral specimens showed the lowest porosity, with mineral density similar to that of the medial specimens. Thus, regional differences in mechanical performance may be related to microstructural differences mediated by remodeling.

We are aware of only one other report of the fatigue behavior of equine cannon bone. Nunamaker et al. (20) conducted fully reversed rotating bending fatigue tests on 1.5 mm diameter specimens of cannon bone under deflection control, at a frequency of 40 Hz and a temperature of 20°C. Extrapolation of their strain-life curve for Thoroughbred racehorses

suggests a static failure strain of less than 10,000 microstrain, the effective strain range employed for fatigue tests in the current experiment. Carter and Hayes (3) showed that bovine bone fatigue life measured at 20°C is approximately twice that measured at 37°C. Therefore, an even larger discrepancy would be predicted had the tests run by Nunamaker et al. been conducted at 37°C. The inconsistency between the results in the current study and those in the study by Nunamaker et al. is likely explained by the different test frequencies and loading modes (i.e., fully reversed and multidirectional bending under deflection control compared with unreversed and unidirectional bending under load control). In rotating bending tests, a larger volume fraction of the specimen is subjected to peak loads (and strains) than for a specimen in four-point bending. Thus, there may be a greater probability of crack initiation from a defect in a rotating bending test than in four-point bending. Frequency dependence of fatigue behavior is not completely understood. In studies of frequency dependence on fatigue life conducted under load control (13,14), fatigue life increased with increasing frequency. Nunamaker et al. (20) found no difference in fatigue life between 10 and 40 Hz, but in other fatigue studies performed under strain control, increased strain rate resulted in a reduced fatigue life (24). Because elastic modulus increases with strain rate (6), load-controlled tests will result in lower peak strains at increased test frequencies, whereas deflection-controlled tests will result in higher peak stresses at increased test frequencies. It appears from these results that fatigue life is a function of both stress and strain and that frequency effects will depend on the control mode of loading. Since the study by Nunamaker et al. used deflection control, one may postulate that, if those tests had been performed at 2 Hz instead of 40 Hz, peak stress would have

been lower and fatigue life would have been higher and more comparable with those in our study.

Comparisons of the present data with those for bovine and human bone suggest that the dorsal and medial cortices of the equine cannon bone have superior fatigue resistance at high strains. Carter and Hayes (3) tested bovine femoral bone in fully reversed (i.e., rotating cantilever) and compressive fatigue. Using 22.3 GPa as the modulus of elasticity of bovine compact bone (22), and extrapolating from their maximum strain values of 4,843 microstrain, the estimated fatigue life at 10,000 microstrain is only 46 and 31 cycles, respectively, for reversed and compressive fatigue. Carter and Hayes (7) subjected bovine bone specimens with reduced central sections to four-point bending at high strains. They showed a 53% reduction in secant modulus after only six load cycles at approximately 10,000 microstrain.

In the case of human femoral bone, Carter and Caler (10) measured the fatigue life at 10,000 micro-strain for uniaxial, fully reversed tension and compression, obtaining, a value of slightly more than 100 cycles. Their data for uniaxial tensile fatigue indicate that failure would occur after only a few cycles at strain ranges well below 10,000 microstrain. Fatigue characteristics of human bone tested in tension and compression at various strain levels were reported by Patten (21), who measured changes in secant modulus during uniaxial fatigue tests on waisted cylindrical specimens from human femora at various effective strain ranges in tension and compression. Representative curves show 25% reductions in modulus after less than 15 cycles at higher strains (4,600-6,700 microstrain). On the other hand, some specimens tested at lower strains (2,800-4,100 microstrain) did not exhibit 25% modulus reductions until over 10,000 cycles had been accumulated. In summary, data from various

experiments suggest that equine metacarpal bone has superior resistance to fatigue compared with human and bovine bone. However, this interpretation is by no means certain, and it would be premature to state categorically that equine bone has superior fatigue resistance. The data may instead show how different modes of loading affect fatigue resistance. Experiments need to be done to distinguish between the effects of species, bone structure, and type of stress on fatigue life.

If the equine cannon bone does have exceptional fatigue resistance, why is it so susceptible to stress fractures and "bucked shins" in living animals? We propose that the answer to this question lies in the fact that *in vitro* laboratory fatigue testing does not account for *in vivo* biological responses to fatigue damage. It is postulated that bone remodeling repairs fatigue damage, so that tolerance of cyclic loading should be better *in vivo* than *in vitro*; however. This is not necessarily the case. Remodeling has been hypothesized to have negative as well as positive effects on fatigue resistance because it introduces additional porosity to the bone (26). This is another source of modulus degradation, increasing strains and shortening fatigue life according to Eq. 1. Martin (16) has used a numerical model to show that there are distinct limits to the amount of additional loading that such a system can tolerate. The model predicts that the fatigue repair system is highly nonlinear, due in part to the exponent in Eq. 1. When pushed beyond its limits, it may enter a "vicious circle" in which porosity (due to remodeling) and fatigue damage rapidly exacerbate one another, leading to failure. "Bucked shins" may represent the early stages of this behavior, and the frequency of complete stress fractures in the face of exceptional *in vitro* fatigue resistance may be entirely attributable to ever increasing

porosity and fatigue damage as remodeling tries to keep up with continued racing and training.

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Test group	Elastic modulus (GPa)			Fatigue life (cycles)				Maximum load (N)			Maximum deflection (mm)		
	Mean	SD	N	Mean	SD	Median	N	Mean	SD	N	Mean	SD	N
Gauge	19.7	3.1	18	5,245	10,741	2,099	11	895	87	6	7.18	1.17	6
No gauge	19.5	2.2	18	2,664	2,490	1,538	12	899	95	6	6.86	0.63	6
Left	19.8	3.1	18	5,523	10,273	1,614	12	892	92	6	6.97	1.15	6
Right	19.5	2.1	18	2,125	1,813	2,099	11	901	90	6	7.07	0.71	6
Group P (periosteal)	20.1	2.6	24	2,318	2,327	1,326	11	897	87	12	7.02	0.91	12
Group E (endosteal)	18.6	2.5	12	5,347	10,239	2,175	12	—	—	—	—	—	0
Dorsal	17.1	1.3	12	8,107	12,979	2,349	7	798	27	4	6.98	0.53	4
Medial	20.4	2.9	12	3,564	2,212	3,179	8	902	33	4	6.29	0.96	4
Lateral	21.3	0.9	12	550	269	505	8	990	35	4	7.79	0.60	4
All	19.6	2.6	36	3,898	7,568	1,750	23	897	87	12	7.02	0.91	12

TABLE 1. *Data from experiment 1: fixed supports*

Test group	Initial modulus (GPa)			Fatigue life (cycles)			
	Mean	SD	N	Mean	SD	Median	N
Dorsal	17.4	1.9	6	1,628	2,375	686	6
Medial	19.2	1.4	6	1,482	1,374	944	6
Lateral	20.8	1.9	6	310	95	332	6
All	19.1	2.2	18	1,140	1,608	579	18

TABLE 2. *Data from eaperiment 2: roller supports*

FIG. 1. Secant moduli are normalized to their initial values and plotted over fatigue life for each fatigue specimen in the first experiment. The dorsal, medial, and lateral regions are plotted separately ($n = 6$ for each plot). The different symbols represent individual specimens.

FIG. 2. Graphs of elastic modulus (normalized by the initial value) versus load cycle to the point of failure, for all the specimens in the second experiment. Six specimens are plotted for each region. The reduced fatigue life in the lateral region is seen, as well as the increasing modulus degradation near failure in the medial and lateral specimens relative to the dorsal specimens. The different symbols represent individual specimens.

FIG. 3 Graphs of fatigue life versus applied stress range for fixed and roller supports. The short lines represent regression for regional groups, and the long lines show regression for all groups combined. The latter are statistically significant: fixed supports, $r = 0.72$, $p < 0.001$; roller supports, $r = 0.59$, $p < 0.011$.