Ultrasonic determination of the elastic modulus of human cortical bone

K. D. Hunt[1] V. Dean O'Loughlin[2] D. W. Fitting[3] L. Adler[4]

[1]Department of Anthropology, Indiana University, Bloomington, IN 47405, USA
[2]Department of Anatomy, Indiana University, Bloomington, IN 47405, USA
[3]National Institute of Standards and Technology, Materials Reliability Division, 325 Broadway, Boulder Colorado 80303, USA
[4]Department of Welding Engineering, Ohio State University, 190 W. 19th Avenue, Colombus, OH 43210, USA

Abstract—*The elastic modulus (C_{ii}) of the cortical bones of 19 individuals (14 femurs and 16 tibias, fixed in formalin) was determined ultrasonically. Elastic moduli were measured at four anatomical positions (anterior, posterior, medial and lateral) and in all three planes of orientation (transverse, longitudinal and radial). The mean tibial C_{ii} (34.11 GPa) was greater than that obtained for femurs (32.52 GPa). The tibial longitudinal plane C_{ii} (34.1 GPa) was significantly greater than the femoral longitudinal plane C_{ii} (32.5 GPa). C_{ii} was significantly higher in the tibia than the femur in both the medial and posterior anatomical positions. The anterior tibia had a significantly lower C_{11} compared to other positions. C_{ii} was significantly higher in the longitudinal plane than the transverse or radial planes in both the femur and the tibia. There was no consistent difference in modulus between left and right sides. No age effects were observed. There were no significant differences between males and females, or between African Americans and European Americans.*

Keywords—*Sex, Age, Race, Femur, Tibia*

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1 Introduction

ALTHOUGH BONE shape is often emphasised when modelling bone response to dynamic loading (cf. CURREY and ALEXANDER, 1985; RUFF, 1984, 1987), the elastic modulus of bone remains among its most important physical attributes. Accurate estimation of the elastic modulus may be important for prosthesis design (LANGCAMER *et al.*, 1993), and it provides important insight into variation in structural demands on long bone shafts in different planes and at different anatomical positions.

Although the elastic modulus has been rather accurately established for the bones of a number of non-human species, physical properties of human bone (BURR, 1980; KEAVENY and HAYES, 1993) are known from smaller sample sizes (Table 1).

RAUBER'S quite early (1876) measurement of the elastic modulus of human bone at 20 GPa has proven consistent with most later static-measurement studies (Table 1; DEMPSTER and LIDDICOAT, 1952; KIMURA, 1952; KO, 1953; SEDLIN and HIRSCH, 1966; BURSTEIN *et al.*, 1975; CARTER and HAYES, 1977; YOON and KATZ, 1976*a*, *b*). Static methods of measuring the elastic modulus may not reflect the true stiffness of whole bone due to bias introduced by the exposure of the Haversian system, machine error, edge effects, heterogeneity bias, shape bias, stress/strain disproportion and stress/time dependencies (BURSTEIN *et al.*, 1975; EVANS and BANG, 1967; ABENDSCHEIN and HYATT, 1970; REILLY and BURSTEIN, 1974). To remedy these biases, a number of investigators have examined the elastic modulus in human bone using ultrasonic methods (ABENDSCHEIN and HYATT, 1970; YOON and KATZ, 1976*a*,*b*; ASHMAN *et al.*, 1984; RHO *et al.*, 1993). Three studies have measured elastic modulus in fresh (i.e. unfrozen, undried, recently extracted) human bone (Table 1). ABENDSCHEIN and HYATT (1970) measured the elastic modulus in femurs and tibias ($n = 4$ individuals). ASHMAN *et al.* (1984) examined femurs only ($n = 5$). RHO *et al.* (1993) examined femurs, tibias and humeri ($n = 8$). These workers did not compare femoral and tibial moduli, presumably because sample sizes were too small to allow it. Only ASHMAN *et al.* (1984) reported the anatomical position of specimens.

Bone density in human femurs is greater medially and laterally than anteriorly and posteriorly (ATKINSON and WEATHERALL, 1967; AMTMANN, 1971*a*). KO (1953) found the medial quadrant of the femur to be the strongest, though other studies have found no such difference (EVANS and LEBOW, 1952; S and HIRSCH, 1966). These differences are consistent with observations that stresses are greatest on the medial and lateral sides of a loaded femur (AMTMANN, 1971*a*), and raise the possibility that elastic modulus might vary with anatomical position as well (ABENDSCHEIN and HYATT, 1970; BURSTEIN *et al.*, 1975; CARTER and HAYES, 1976, 1977; CURREY, 1969, 1970, 1975; DUCHEYNE *et al.*, 1977; EVANS, 1973; GOODBREAD, 1976; VIANO *et al.*, 1976; VOSE and KUBALA, 1959; WRIGHT and HAYES, 1976).

Correspondence should be addressed to Dr. Hunt

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Table 1 Elastic modulus of human cortical bone[a]

Loading	Loading rate (s^{-1})	E (GPa)	Reference
tension	low?	20.0	RAUBER, 1876
tension	low	14.1 (rewetted)	DEMPSTER and LIDDICOAT, 1952
compression	low	10.4	KIMURA, 1952
tension	low	17.3	KO, 1953
tension	low	17.2	SWEENEY *et al.*, 1965
bending	not reported	15.5	SEDLIN and HIRSCH, 1966
tension	not reported	6.0	SEDLIN and HIRSCH, 1966
tension	0.1	14.1	BURSTEIN *et al.*, 1975
tension	0.1	22.3	BURSTEIN *et al.*, 1976
tension	5.3×10^{-4}–237	17.7–40.4	WRIGHT and HAYES, 1976
ultrasonic	–	24.5 (fresh)	($n = 4$)[b], ABENDSCHEIN and HYATT, 1970
ultrasonic	–	32.5 (dry)	YOON and KATZ, 1976b
ultrasonic	–	27.6 (fresh)	($n = 5$), ASHMAN *et al.*, 1984
ultrasonic	–	20.7 (fresh)	($n = 8$), RHO *et al.*, 1993
ultrasonic	–	$C_{33} = 32.51$	($n = 14$), femur, this study
ultrasonic	–	$C_{33} = 34.07$	($n = 16$), tibia, this study

[a] Values compiled for wet bone only, except where noted
[b] Sample sizes are given for number of individuals, not number of bone samples

Here, ultrasonically determined values for the elastic modulus (C_{ii}) are reported for human femoral and tibial bones, preserved in 10% formalin. Bone samples were obtained from a slightly larger number of individuals than previous studies (19 individuals, 16 tibias and 14 femurs), allowing for statistical comparisons between anatomical positions, among longitudinal, transverse and radial planes, between the tibia and the femur, and between races and sexes.

2 Methods

Equipment and methods for ultrasonic measurements were substantially similar to those of YOON and KATZ (1976*b*) and ASHMAN *et al.* (1984). Bone specimens were obtained from amputations at the University of the Tennessee Hospital, Knoxville, Tennessee.

Amputation was indicated because of arteriosclerosis in 12 cases, diabetes in three, thrombosis in two and severe injury resulting in vascular insufficiency in two. All patients were ambulatory at the time of surgery. Samples were obtained from 19 individuals, ten males and nine females, aged 55 to 98, and a total of 126 separate bone samples were analysed. Modulus values were determined for 16 femurs (14 individuals) and 16 tibias. For the two individuals for which both left and right femur samples were available, elastic moduli were calculated for both sides and averaged to obtain a single value for each plane and each anatomical position. Sample sizes in tables represent the number of individuals, not the number of bone samples.

Limbs were refrigerated after amputation, but not frozen. In all cases bone samples were removed within 48 h of amputation. Femoral samples were taken from the diaphysis (i.e. only cortical bone was sampled) as close to the midshaft as could be ascertained on the amputated limbs. The areas from which femoral samples were taken ranged from 10–23 cm from the distal articular surface, with a median of 15 cm. All tibial samples were taken exactly from the midshaft, defined as that point halfway between the distal patella and the medial malleolus. After an ~ 5 cm section of the diaphysis was extracted, a tracing of the cross-section was made, on which the anatomical position was noted. This procedure took less than a minute, after which the bones were immediately stored in 10% formalin to avoid drying. A variety of methods of preservation, including storage in formalin, seem to increase the elastic modulus (LANG, 1969; SMITH and WALMSLEY, 1959; EVANS and LEBOW, 1952; SEDLIN and HIRSCH, 1966; EVANS, 1973). Values here are therefore expected to be higher than *in vivo*.

The periosteum was stripped from the samples and four roughly cubical samples were taken from anterior, posterior, medial and lateral aspects of the cortex using a Buehler low-speed diamond saw. The mean dimensions of the specimens ($n = 116$) were 4.86 ± 1.36 mm (transvers dimension) 2.89 ± 1.16 mm (radial) $\times$ 4.67 ± 0.80 (longitudinal). Samples were taken as close to the centre of cortical bone (i.e. as far from the periosteal and endosteal surfaces) as possible, and in no case was the distance to either surface less than 2–3 mm. The saw blade was kept cool by constant application of Buehler Isocut fluid, which also kept specimens wet during cutting. The saw speed was kept at < 150 rpm to prevent scratching and heating. No scratches were observed on the specimens under $50 \times$ magnification. Higher magnification revealed scratches, but they were considerably smaller than the Haversian system. Machined specimens were then returned to formalin in individual containers until measurements were made. Storage times between amputation and the elastic modulus measurement ranged from 5 days to 5 months, although it was less than one month in most cases. Specimens were measured in transverse, radial and longitudinal dimensions with an outside micrometer* calibrated to 0.01 mm.

Specimens were secured in a device in which two 4 mm diameter, 5 MHz, compressional-mode transducers† were affixed with phenyl salicylate cement to two parallel ground and polished (1 micron alpha alumina) aluminium buffer discs (5.04 mm thick). The top buffer rod and its transducer could be moved with a threaded rod to apply clamping pressure to secure the bone specimen between the buffer discs. A pulser/receiver‡ delivered a 230 V pulse of 0.2 ms duration to the piezoelectric transmitting transducer to create an ultrasonic

*Starrett #436
†Valpey-Fischer PZT-5A
‡Panametrics 5055 PR

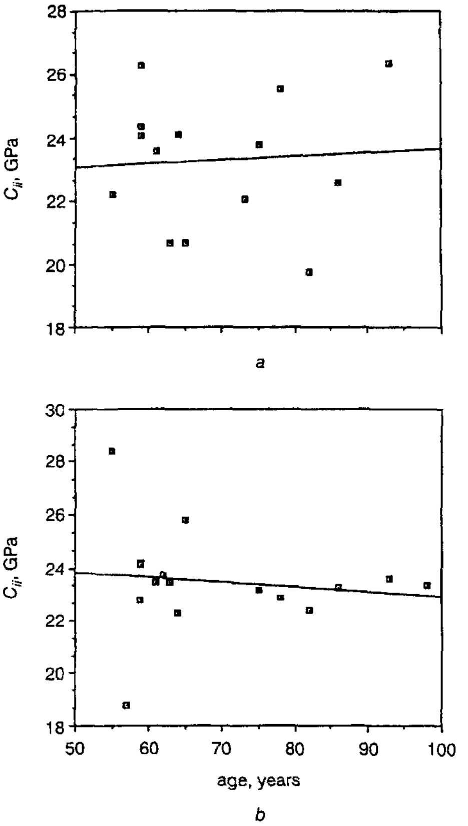

Fig. 1 *A comparison of elastic modulus (C_{ii} values for anatomical position and plane averaged with age shows no significant change in C_{ii} in either the femur (A, $n = 14$) or the tibia (B, $n = 16$)*

pulse. The receiving transducer was connected to the receiver input of the pulser/receiver. The receiver gain was adjusted for each bone specimen to yield roughly the same amplitude displayed receiver waveform.

Ultrasonic pulse transit times through the buffer rods and bone samples were measured using an oscilloscope*, with a calibrated time base. The digital 'delta-time' function of the oscilloscope was used for making all time-interval measurements. In all cases the ultrasonic wavelength (λ) was ~ 0.3 mm, much larger than the Haversian system of the bone. The ultrasonic pulse transit time through the buffer discs was measured by placing the transmitting and receiving transducer buffer discs in contact with a very thin layer of ultrasonic couplant (cellulose gel) between them. The transit time was taken as the time interval between the pulser trigger and the first excursion of the receiver waveform from zero. Measurement of the ultrasonic transit time through the bone specimen was made by applying a thin ultrasonic couplant layer to opposite sides of the bone cube and placing it between the buffer discs in the measurement fixture. The total transit time (through discs and bone) was measured. The transit time through the buffer discs was subtracted from the total transit time to yield the ultrasonic pulse transit time through the bone sample.

The longitudinal wave speed in the bone was calculated by dividing the sample thickness by the pulse transit time through it. Longitudinal wave speed measurements were made with three orientations of the bone sample; longitudinal, radial and transverse. These axes correspond to the axes of symmetry of bone (transversely isotropic) and so are pure-mode propagation directions. That is, longitudinal waves propagate with particle motion parallel to the wave propagation direction. Because the measurements are in pure-mode directions the elastic stiffnesses (C_{11}, C_{22} and C_{33}) along the diagonal of the stiffness matrix can be computed by multiplying the bone density by the square of the measured wave speed. These are the moduli reported in this paper as C_{ii} though this measure is often reported as E using the same formula. A common value of 1.86 g cm^{-3} was used as the density of the specimens (YOON and KATZ, 1976*a*), a technique virtually identical to that of RHO *et al.*, 1993. Such an assumption, while clearly inferior to measuring density on individual samples, is justified because bone density of healthy individuals typically does not change dramatically until the ninth decade (LINDAHL and LINDGREN, 1967). Specimens were examined under the microscope at 50 x for pathology, including lowered bone density due to osteoporosis. All samples analysed appeared normal.

One way analysis of variance and multiple comparison procedures were used instead of *t*-tests where possible. The former methods set up more stringent criteria for declaring significant differences than *t*-tests.

3 Results

There was no significant correlation between age and C_{ii} in this sample (Fig. 1) in either the femur ($R^2 = 0.001$, $n = 14$), or the tibia ($R^2 = 0.014$, $n = 16$), and therefore no indication in this sample that the elastic modulus changes significantly with age.

There was one significant difference between left and right bones (Table 2). C_{ii} was determined in each plane by an average of the values in that plane in the four anatomical positions (anterior, posterior, medial and lateral). C_{ii} in the radial plane was significantly greater in the right femur than the left (Table 2). There were no side differences by anatomical position. When C_{ii} was averaged for all three planes an analysis of variance revealed no statistically significant differences between left and right femurs for any of the four (anterior, posterior, medial or lateral) anatomical positions.

The mean elastic moduli for femoral radial and transverse planes (averaged over anatomical position) were comparable (25.23 against 23.43 GPa; see Table 3) and did not differ significantly (Table 4). The value for the longitudinal plane

**Hewlett-packard Model 1740A

Table 2 Analysis of variance of elastic modulus by side

Bone	Plane	Left C_{ii} GPa[a]	n	Right C_{ii}, GPa[a]	n	F ratio	p-value
femur	transverse	26.98	6	24.12	5–6	2.092	0.182
femur	radial	22.25	5	24.80	5–6	7.354	0.024
femur	longitudinal	32.82	6	32.39	6	1.180	0.306
tibia	transverse	25.01	6–7	23.68	8–9	3.517	0.082
tibia	radial	22.94	5–7	23.60	8–9	0.360	0.558
tibia	longitudinal	34.72	6–7	33.47	9	1.361	0.263

[a] Average of anterior, posterior, medial and lateral values

Table 3 *Average femoral elastic modulus by anatomical position*

Plane	n	C_{ii} GPa[a]	S.D.	Minimum	Maximum
transverse	14	25.23	3.57	22.78	36.18
radial	13	23.43	20.25	20.25	26.36
longitudinal	14	32.51	0.87	31.33	34.69

[a] Average of anterior, posterior, medial and lateral values

Table 4 *2-tailed t-tests for elastic modulus between planes*

Bone	Planes compared	*t*-value	*p*-value
femur	transverse vs. radial	1.37	0.196
femur	transverse vs. longitudinal	−7.80	0.000
femur	radial vs. longitudinal	−16.16	0.000
tibia	transverse vs. radial	1.37	0.190
tibia	transverse vs. longitudinal	−17.92	0.000
tibia	radial vs. longitudinal	−14.22	0.000

(C_{33}) was significantly larger (in a 2-tailed *t*-test (see Table 4). The same differences were found in the tibia: the elastic modulus was significantly greater (2-tailed *t*-test; Table 4) in the C_{33} (longitudinal plane) (34.07 GPa), but was similar in the transverse C_{11} and radial C_{22} planes (23.44 against 24.3 GPa, respectively).

An LSD multiple comparison test was conducted to determine whether there were differences in elastic modulus in the four anatomical positions. The null hypothesis, that all four anatomical positions have the same C_{ii} in the various planes, was rejected for the tibial transverse plane (Tables 5 and 6), but not for the femur (Table 6). The LSD multiple comparison test showed that in the transverse plane the elastic modulus of the anterior tibia was significantly lower than the modulus of the lateral and posterior positions. There were no significant differences by anatomical position for the other planes.

The C_{33} longitudinal elastic moduli (averaged over all anatomical positions) were greater in the tibia than the femur; there were no significant differences for transverse and radial samples for any anatomical position.

C_{ii} of the tibia was significantly greater than in the femur in the medial ($p = 0.004$) and posterior positions ($p = 0.005$; Table 7). The anterior value approached significance ($p = 0.087$).

For the elastic modulus averaged over all anatomical positions, females had higher elastic moduli in all three planes in the femur, and in two of the three planes for the tibia, though none of the differences were statistically significant (Table 8; C_{ii} was nearly identical in males and females in the transverse plane).

African Americans had higher elastic moduli in two of three planes for both the femur and the tibia, but the differences were not significant (Table 9). Note, however, that the longitudinal C_{ii} (i.e. C_{33}) approaches significance ($p = 0.06$). The consistency of sex and race differences suggest that further examination of population differences in C_{ii} is warranted.

4 Discussion and conclusions

Age, sex and race differences were not significant, but sample sizes were small enough that biologically significant differences may not have been detected. Although tibial specimens showed some suggestion of decreasing C_{ii} with age, the slope of C_{ii} over age was not significantly different from zero for either the tibia or the femur. Females had higher moduli than males, though not significantly. African Americans had consistently higher moduli compared to European Americans, though again not significantly. Further studies of sex and population differences in humans seem warranted.

One might expect modulus differences by side in leg bones due to population-wide leg preferences. The trends exhibited here cannot be easily reconciled with this reasoning, since side differences were not dramatic in either the femur or the tibia. The right femur was significantly stiffer in the radial plane, but there was no overall trend, as the elastic modulus was higher in other planes for the left femur. Neither were there side differences when analysed by anatomical position.

Table 6 *Oneway ANOVA of elastic modulus by anatomical location*

Bone	Plane	*f*-ratio	*p*-value
femur	transverse	0.600	0.688
femur	radial	0.950	0.424
femur	longitudinal	0.906	0.445
tibia	transverse	3.522	0.021
tibia	radial	1.768	0.164
tibia	longitudinal	1.584	0.203

Table 5 *Average tibial elastic modulus by anatomical position*

Plane	Anatomical position	*n*	C_{ii}, GPa	S.D.	Minimum	Maximum
transverse	medial	15	24.38	1.38	22.76	28.26
transverse	lateral	16	25.44	3.74	21.91	38.33
transverse	anterior	14	22.13	3.72	18.83	34.49
transverse	posterior	15	25.24	2.87	22.20	34.79
mean transverse	14–16	14–16	24.30	2.93	18.83	38.33
radial	medial	13	23.44	1.69	21.33	26.78
radial	lateral	15	23.42	7.36	5.53	42.51
radial	anterior	16	21.58	0.93	19.85	24.05
radial	posterior	15	25.30	4.69	21.34	41.65
mean radial		13–16	23.44	3.69	5.53	42.51
longitudinal	medial	15	34.26	1.99	30.13	38.45
longitudinal	lateral	16	32.05	5.55	16.75	37.21
longitudinal	anterior	16	35.88	7.67	31.53	64.21
longitudinal	posterior	15	34.07	1.74	30.08	37.43
mean longitudinal		15–16	34.07	4.31	16.75	64.21

Table 7 Longitudinal plane elastic modulus comparison in femur and tibia

	Femoran C_{ii} GPa	n	Tibial C_{ii} GPa	n	F-ratio	p-value
medial	32.33	14	34.26	15	9.657	0.004
lateral	33.35	14	32.07	15	0.569	0.457
anterior	32.01	14	36.03	15	3.145	0.087
posterior	32.37	14	34.07	15	9.360	0.005
mean	32.52	14	34.11	15		

Table 8 Elastic modulus (GPa) by sex

Bone	Plane	Male	n	Female	n	F-ratio	p-value
femur	transverse	24.88	7	25.71	6	0.152	0.705
femur	radial	22.99	7	23.94	6	0.676	0.429
femur	longitudinal	32.49	7	32.63	6	0.074	0.791
tibia	transverse	24.30	7	24.23	9	0.007	0.935
tibia	radial	23.16	7	23.42	9	0.055	0.818
tibia	longitudinal	33.58	7	34.36	9	0.494	0.494

Table 9 Elastic modulus (GPa) by race

Bone	Plane	African	n	European	n	F-ratio	p-value
femur	transverse	27.93	3	24.46	10	2.210	0.165
femur	radial	23.70	3	23.35	10	0.063	0.807
femur	longitudinal	32.47	3	32.58	10	0.032	0.861
tibia	transverse	24.63	4	24.14	12	0.288	0.600
tibia	radial	22.74	4	23.50	12	0.369	0.553
tibia	longitudinal	35.75	4	33.44	12	4.180	0.060

The elastic modulus was greatest in the longitudinal plane in both the femur and the tibia, whereas there were no significant differences between the transverse and radial planes. This accords with traditional understanding of bone as a transversely isotropic material (DEMPSTER and LIDDICOAT, 1952; BONFIELD and GRYNPAS, 1977). This quality of bone makes it particularly important that researchers accurately orient specimens correctly when measuring the elastic modulus.

The tibia appears to be stiffer of the two bones. The elastic modulus in the longitudinal plane of the tibia (34.1 GPa) was significantly greater than that of the femur (32.5 GPa). When analysed by bone and anatomical position (averaging the values for the three planes) C_{ii} was greater in the tibia at both the medial and posterior positions.

There was only one significant differences by anatomical position. In one plane (transverse) the anterior aspect of the tibia had a lower C_{ii} than other anatomical positions. Although no significant differences in anatomical position were noted for the femur, C_{ii} was highest in the femur laterally.

This observation is intriguing, since the tibia has a higher elastic modulus than the femur medially (Table 7; $p=0.004$), posteriorly ($p=0.005$) and anteriorly (not significantly, $p=0.087$), and it was only in the lateral position that C_{ii} was statistically indistinguishable in the femur and tibia (Table 7). The higher C_{ii} for the lateral part of the femur is not unexpected, since there is evidence (ATKINSON and WEATHERELL, 1967; AMTMANN, 1971*b*; BURR, 1980) that during locomotion stresses are greater medially and laterally in the femur than anteriorly and posteriorly, presumably tension laterally and compression medially. This pattern suggests that during routine stressing of the lower limb the greater stiffness of the tibia and its relatively vertical orientation means that it should experience little lateral bending. Because the femur is stiffer laterally, under compressive stress it might be expected to bow medially more by medial compression than by lateral tension. Perhaps the femur has evolved to be the principal shock-absorbing element in the lower limb during weight bearing, whereas the tibia is a less flexible strut. These variable physical properties in the lower limb bones hint at an interesting dynamic shock absorbing function that warrants examination *in vivo*.

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Authors' biographies

Kevin D. Hunt is an associate professor of anthropology at Indiana University. He received a BA from the University of Tennessee in 1980, and a Ph.D. from the University of Michigan in 1989. His principal research interest is the ecological morphology of early hominids, particularly the evolution of human bipedalism. He has studied locomotion and posture in primates at the Gombe Stream and Mahale Mountains National Parks in Tanzania and at the Kibale Forest Reserve in Uganda. He is currently studying dry-habitat chimpanzee locomotion and posture at the Semliki Valley Wildlife Reserve, Uganda.

Valerie Dean O'Loughlin received her Ph.D. in Biological Anthropology from Indiana University, Bloomington in 1995. She is currently a visiting Assistant Professor of Anatomy in the Medical Sciences Program at Indiana University. Her interests include craniofacial morphology, anatomy, skeletal biology and paleopathology. Her Ph.D. thesis used New World archaeological populations to examine the effects of cranial deformation on the endocranial vasculature. Recent publications include 'Sinus and memingeal vessel pattern changes induced by artificial cranial deformation' (*Int. J. Osteoarchaeol.*, 1995) and 'Comparative endocranial vascular changes due to craniosynostosis and artificial cranial deformation' (*Am. J. Phys. Anthropol.*, 1996).

Dale W. Fitting received a BS in 1971 from the Colorado School of Mines, an MS from the University of Colorado, and a Ph.D. in engineering science and mechanics from the University of Tennessee. He is a materials research engineer at the National Institute of Standards and Technology, in Boulder, Colorado, where he is investigating X-ray diffraction, X-ray energy-dispersive techniques and ultrasonic point-source/point-received measurements. Dr. Fitting has worked on ultrasonic arrays for high-resolution imaging, quantitative materials characterisation and ultrasonic computed tomography. He has done research and taught at the University of Michigan, Ohio State University, Oak Ridge National Laboratory and the University of Tennessee. He co-authored a book reviewing ultrasonic spectral analysis techniques for nondestructive testing.

Laszlo Adler is the Taine McDougal Professor of welding engineering and engineering mechanics, and Director of the Nondestructive Evaluation Program at Ohio State University. His Ph.D. is from the University of Tennessee, in physics. He as written a book on ultrasonic spectroscopy, and has published over 200 scientific and technical papers. He is a member of the editorial board for the *Journal of Nondestructive Evaluation*, of the Physical Acoustics Technical Committee, and of the National Academy of Sciences Advisory Committee. He is a Fellow of the Acoustical Society of America. He was awarded a Distinguished Research Professorship in 1984, and is a Distinguished Scholar at Ohio State University.