SYNTHETIC BONE FOR IMPACT STUDIES

Dr. John N. Snider
Alumni Gym
Industrial Engineering Dept.
University of Tennessee
Knoxville, TN 37996

Dr. Peter M. Fuller
500 S. Preston Street
Health Sciences Center
University of Louisville
Louisville, KY 40292

Dr. Jack Wasserman
Perkins Hall
Engr. Sci. & Mech. Dept.
University of Tennessee
Knoxville, TN 37996

Dr. Roberto Benson
Dougherty Hall
Materials Science Dept.
University of Tennessee
Knoxville, TN 37996

Mr. Tyler Kress
Perkins Hall
Engr. Sci. & Mech. Dept.
University of Tennessee
Knoxville, TN 37996

Mr. Guy Tucker
Perkins Hall
Engr. Sci. & Mech. Dept.
University of Tennessee
Knoxville, TN 37996

ABSTRACT

Use of crash test dummies has become the standard method for the study of human biodynamics during impact loading. They are used to study both the mechanism of injury and methods of protection from injury in the aircraft and automotive industries. Studies repeatedly demonstrate the limitations of the current dummies when they have been used for the simulation of motorcycle and pedestrian accidents. There is a definite need for the development of replacement limbs for the existing dummies. Research over the past few years has provided the insight necessary for the design of a breakable lower limb to be incorporated into the Hybrid III test dummy. This paper discusses the preliminary design of the artificial bone structure that is to be used in the Analogue Human Leg Structure(AHLS). From a material science viewpoint, natural bone has an excellent strength to weight ratio because it is a fiber-reinforced composite material. Therefore, a polymer composite or a specifically processed liquid crystalline polymer is most appropriate. Our laboratory findings have indicated that a fiber/Hydroxyapatite(HAP) reinforced polyamide can be designed to have certain mechanical properties which are close to those measured for bone.

PURPOSE

An objective of this study is to develop an artificial bone structure which has the same dynamic response to impact loading as the natural bones of the human leg. It is believed that an artificial bone can be developed which will also have useful properties for orthopedic applications.

From a material science viewpoint, natural bone has an excellent strength to weight ratio because it is a fiber reinforced composite material. Therefore, a polymer composite, or a specially processed liquid crystalline polymer is most appropriate. The advantage of using reinforced composites is the ease with which their properties can be tailored to specific applications, thus making them ideal choices for this type of high strength application. Preliminary studies from our laboratory have shown that a fiber/Hydroxyapatite (HAP) reinforced polyamide can be designed to have certain mechanical properties which are close to those measured for bone. Polymer processing and fiber selection are important considerations in the final design of the bone simulant. Particular attention will be given to the use of injection molding since it generally introduces molecular and fiber orientation which provide favorable mechanical properties. Radial variation of the porosity might aid in the development of the final leg model, which will probably include selected foams or polyurethanes to simulate the soft tissue. The stiffness and void fraction will be controlled during processing. The availability of these porous prostheses will permit modeling of all bone tissue.

LITERATURE REVIEW OF MECHANICAL PROPERTIES OF BONE

A good characterization of leg bone properties is essential to the development of a synthetic bone for impact studies. If leg bone properties were completely characterized for both static and dynamic conditions, then a polymer or some other material could be "designed" to have those properties. "Designed" in this context means that the synthetic bone material properties can be altered by using blends and fillers in such a way that the properties eventually resemble those of real bone. The mechanical properties of bone are well discussed in the literature. Researchers Yamada [1], Evans [2] and McElhaney [3] pioneered the study of mechanical properties of bone. The data have been continually updated by Currey [4], Reilly and Burnstein [5], Carter and Spengler [6], Fung [7], Van Buskirk and Ashman [8], and Cowin et al. [9] This bone-property overview is a very brief sketch of their work. For a thorough and detailed discussion of the mechanical properties of bone refer to Bone Mechanics [10] by Cowin.

In biological terms bone is described as a connective tissue, an aggregation of similarly specialized cells united in the performance of a particular function. In bioengineering terms bone can be viewed as a nonhomogeneous anisotropic composite. In the literature, bone is often divided into two categories, especially with reference to its mechanical properties. These categories are dehydrated and hydrated, and are often referred to as old and fresh or dry and wet. In general, dry bone is brittle and fails at a strain of approximately 0.4% and wet bone fails at a strain of about 1.2%. Wet, of course, is of most interest to this paper, because it best represents the in vivo bone.

The volumetric composition of bone tissue can be divided into almost equal thirds: water, minerals, and collagenous matrix. Even among like bones from human to human, this composition can vary. When just considering human leg bones, variations exist with the age, the sex, and whether or not the individual has experienced a bone disease. About two-thirds of the weight of bone, or half its volume, is inorganic material with the composition of hydroxyapatite which is present as tiny elongated crystals of the order of 200 Å long with an average cross-section of 2500 Å [2]. The rest of the bone is collagen fibers. Different water content and salt content have significant effects on the mechanical properties. The role of water in bone is somewhat obscure as discussed by Timmins and Wall [11]. However, variation in water content with age is fairly well documented, so a correlation might be drawn between water content and ductility.

Bone has been assumed to be transversely isotropic [12,13,14,15] and also to be an orthotropic [16,17,18] material. In order to obtain technical constants for human bone, researchers have used a couple of methods: 1) ultrasound, in which the measured velocities are used to determine elastic coefficients and technical constants are then found by matrix inversion, and 2) standard testing in which load machines are used to make direct measurements. Table 1 presents the technical constants for human bone measured by various investigators. The material symmetry generally assumed is that of transverse isotropy (TI) or orthotropy (ORTH). An important observation is that stiffness in the circumferential direction is always greater than the stiffness in the radial direction. Yanson, et al. [19], suggest that the lower stiffness in the radial direction is associated with the greater permeability in that direction. Blood flow is less in the circumferential direction as opposed to the radial direction. So, for cortical tissue of long bones, an orthotropic assumption might be more accurate.

Viscoelasticity, the effect of strain rate on the stress-strain curve, is an important phenomenon of bone. McElhaney [3] indicated that, for the embalmed human femur in compression, bone is stiffer and stronger at higher strain rates. Carter and Hayes [20] found that both strength and modulus of elasticity were approximately proportional to the .06 power of strain rate.

Some mechanical properties of human leg bones are presented in Table 2 (data adopted from Yamada [1] and Fung [7]). In general, it is well known that the strength of bone varies with the age and sex of the human, the location of the bone, the orientation of the load, the strain rate, and the specimen condition (whether it is dry or wet). The higher strain rate effect may be especially significant, with higher ultimate strength being obtained at higher strain rate. Another note is that the strength and modulus of elasticity of spongy bone are much smaller than those of compact bone (Yamada [1] presents human vertebrae data as support).

The literature provides a basis for comparison of real bone properties to those of simulant bone. Motoshima [21] tested long wet leg bones of 13 fresh cadavers ranging from 20 to 83 years old. Some of Motoshima's results are presented in Table 3 and will serve as an excellent static comparison in the search for a bone simulant.

Table 1[a]

Technical Constants for Human Bones

Group	Relly and Burnstein	Yoon and Katz	Knets et al.	Ashman et al.
Bone	Femur	Femur	Tibia	Femur
Symmetry	TI	TI	ORTH	ORTH
Method	M	U	M	U
E_1 (GPa)	11.5	18.8	6.91	12.0
E_2 (GPa)	11.5	18.8	8.51	13.4
E_3 (GPa)	17.0	27.4	18.4	20.0
G_{12} (GPa)	3.6[b]	7.17	2.41	4.53
G_{13} (GPa)	3.3	8.71	3.56	5.61
G_{23} (GPa)	3.3	8.71	4.91	6.23
ν_{12}	0.58	0.312	0.49	0.376
ν_{13}	0.31[b]	0.193	0.12	0.222
ν_{23}	0.31[b]	0.193	0.14	0.235
ν_{21}	0.58	0.312	0.62	0.422
ν_{31}	0.46	0.281	0.32	0.371
ν_{32}	0.46	0.281	0.31	0.350

E = Modulus of Elasticity
G = Modulus of Rigidity
ν = Poisson's Ratio

Note: The "three" direction is coincident with the long axis of the bone; the "one" and "two" directions are radial and circumferential, respectively. Method U is ultrasound and method M is standard machine testing. TI, transverse isotropy; ORTH, orthotropy.

[a]Cowin, 1989
[b]Not measured

Table 2[a]

Mechanical Properties of Wet Compact Human Bone (20-39 Yrs.)

Mechanical Property	Value
Ultimate Tensile Strength (Femur)	124 MPa
Ultimate Tensile Strength (Tibia)	174 MPa
Ultimate Percentage Elongation (Femur)	1.41
Ultimate Percentage Elongation (Tibia)	1.50
Modulus of Elasticity in Tension (Femur)	17.6 GPa
Modulus of Elasticity in Tension (Tibia)	18.4 GPa
Ultimate Compressive Strength (Femur)	170 MPa
Ultimate Percentage Contraction (Femur)	1.85
Ultimate Shear Strength (Femur)	54 MPa
Torsional Modulus of Elasticity (Femur)	3.2 GPa

[a]Yamada, 1970; Fung, 1981

Table 3[a]

Static Properties of Bone

Mechanical Property	Value
E_t, Modulus of Elasticity (tension)	1.0×10^{10} Pa
σ_{y_t}, Yield Stress (tension)	1.3×10^{8} Pa
σ_{y_b}, Yield Stress (bending)	4.3×10^{7} Pa
σ_{u_b}, Ultimate Bending Stress	5.9×10^{8} Pa

[a]Motoshima, 1959

APPROACH

It is believed that the impact fracture of bone and the resulting extent of damage is describable in terms of classical elastic/plastic mechanics of non-isotropic composite materials. Consequently, it is proposed to develop a 3-D finite element computer representation of bone structures in which the material properties are allowed to be dependent on strain rate, spatial position, and direction. Strain rate dependent properties (modulus of elasticity, yield stress, ultimate stress, ultimate strain, Poisson's ratio) could be measured for real bone samples oriented in the radial, circumferential, and axial directions. These properties could be input, initially uniformly, into the computer model which would be exercised under impact loading simulation conditions. The model results could then be compared with actual impact experiments on real complete bone specimens in terms of energy absorbed to failure, extent of damage, stress/strain distributions, and displacements. The model property values and distributions could be adjusted to obtain a "best" correspondence between the model and the test results. The "validated" model could then be subjected to a sensitivity study to determine the most sensitive parameters and how their variations affect the calculated results. The results of this would be a set of desirable properties and their distributions for a physical model.

The ranges of possible properties attainable by composites consisting of fiber-reinforced polyamide containing different quantities and types of fiber could then be measured experimentally. This could allow a determination of whether or not the desirable range of physical properties can be matched sufficiently well that the synthetic bone could be expected to be a good representative of real bone under impact loading conditions.

Physical models could then be constructed from the material of the desirable make-up and then subjected to impact testing that could be compared with both the actual bone test results and with the computer finite element results.

This paper presents results of exploratory studies in which impact data on human leg bones are developed and compared with similar impact data on carbon and Kevlar fiber-reinforced polyamide (Nylon 6,6).

The impact apparatus used in this research consists of three main components; a specimen holding device, the impactor support cart and its associated guideway, and the cart accelerator system. In operation, the cart accelerator system accelerates the impactor support cart to the desired velocity after which the impactor strikes the test specimen. The impactor support cart is then stopped by means of direct impact with energy-absorbing bales of wood fiber.

Four embalmed human tibias and four fiber-reinforced polyamide specimens were impacted transversely at midshaft by the constant velocity ($\Delta v<3\%$) impactor. The bones were simply-supported and impact velocity was approximately 7.5 m/s.

A force transducer was mounted on the backside of the impact pipe in such a way that the force that the bumper exerted on the bone was transmitted directly to the transducer.

PRESENT RESEARCH PROGRAM STATUS

A facility was developed to simulate impact conditions on cadaver, animal, and model specimens with impact velocities up to 13.41 m/s (30 miles per hour). This facility and other complementary laboratories provide the capability of: 1) testing a variety of specimens ranging from bone to a full cadaver (or dummy), 2) state-of-the-art date acquisition, 3) determining mechanical properties of various materials including bone, and 4) developing bone simulant specimens.

A variety of dynamic response experiments have been conducted to date. These include intact cadaver legs, human tibias, human femurs, intact goat legs, dog bones (humeri, femora, tibiae - mechanical properties only), horse bones, bakelite as a bone simulant, and fiber-reinforced polyamide as a bone simulant.

These studies have provided guidance in determining load characteristics and injury severity. After complete characterization, the ultimate objective is to develop a dummy which approximates the human body. The lower leg simulant will consist of a fiber-reinforced composite for a bone replacement, and, most likely, a polyurethane foam as a soft tissue substitute.

Additional impact tests and complementary laboratory experiments will provide important mechanical property information for use in developing the computer model and the final bone simulant. If material properties were known, it would be entirely possible to develop a complete finite element computer model of the human leg's response to impact loading. Such a model is believed necessary for the design of a simulant that behaves in the same manner as the human leg.

An eventual goal of this research is to develop an entire dummy leg including the hip, femur, knee, tibia, ankle, foot, and surrounding soft tissue mass. This leg could be used in place of the existing legs of modern crash test dummies.

RESULTS

Figure 1 shows a typical example of a force versus time curve from the transducer during an impact test of a polyamide specimen. The first encircled point marks the time at which initial contact is made between the bumper and the specimen. The second encircled point marks the time, t_f, and force level at which the specimen ultimately fails. Similar data were obtained for each of the polyamide specimens and each of the bones. The energy absorption capacity, U/V, was obtained from these data by evaluating the following relation:

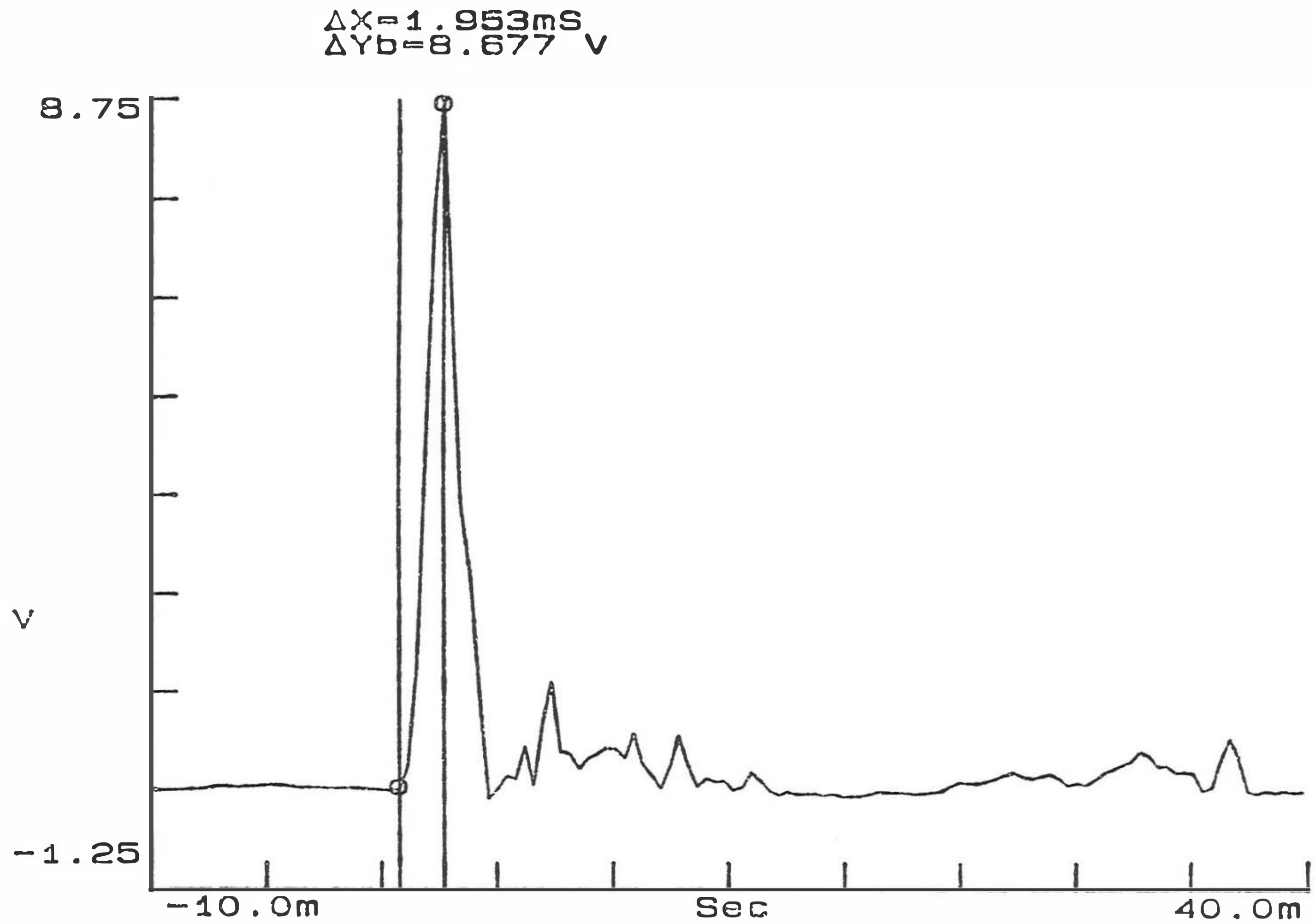

Figure 1. Force versus time data for a 4.1275-centimeter pipe high-speed impact onto a fiber-reinforced polyamide specimen (force constant = 2.248 mV/N).

$$\frac{U}{V} = \frac{v_o \int_0^{t_f} F\, dt - KE}{V}$$

where U = internal energy absorbed to time t_f,
v_o = impactor velocity,
t_f = time from instant of contact to failure,
F = impact force measurement,
t = time,
V = cortex volume between support points,
and KE = the kinetic energy of the bone specimen, at the instant of failure.

The kinetic energy was obtained by assuming a linear velocity profile from v_o at the specimen midshaft down to $v=0$ at each end. A mean cross-sectional area, $\bar{A}_{cs}$ (for the tibias), was approximated by the product of an average circumference and an average cortex thickness. Circumferences for the tibia were measured at three axial positions and averaged as was the cortex thickness. With these approximations

$$KE = \frac{\bar{A}_{cs}\,\rho}{g_c} \int_0^L v_y^{\,2}\, dy$$

where v_y = local transverse vel. at position y
$= v_o(1 - y/L)$,
and y = longitudinal direction with zero at the midshaft,
ρ = bone density = avg. value of 1900 kg/m^3 (Cowin, 1989),
g_c = proportionality coefficient $= 1\ \dfrac{kg\text{-}m}{N\text{-}s^2}$,
and L = length from midshaft of specimen to point at which specimen contacts support.

The volume was obtained by multiplying the average cross-sectional area defined above by 2L.

The polyamide specimens were cylinders with a length of 17.780 cm and a diameter of 1.905 cm. One to three centimeters at the ends were utilized for the simple support so the volume calculated was the product of the cross-sectional area and the length from support to support. In order to make the specimens "useful" under torsional loading conditions and avoid making them hollow, a unique method of processing was used. Three small polyamide rods (≈0.4 cm diameter) were wrapped with long Kevlar and carbon fiber and then wrapped together into a "triangular" shape with more Kevlar and carbon fiber. Polyamide was then poured around the grouped fiber-reinforced rods by injection molding to make the finished product. The fibers added substantial bending strength yet will slip in the circumferential direction if any torsional loads are applied.

The average energy absorption capacity for the bone specimen was approximately 18,610 J/m^3 as compared to 28,123 J/m^3 for the polyamide specimens. When comparing these numbers, note that the sample size was small and that there were some possible sources of error that include: 1) the calculated volumes of the tibias are just approximations. They are relative from specimen to specimen, but their variance is not linear, hence an error can be introduced, and 2) the areas under the force-time curves for all specimens were approximated. The approximations were obtained by assuming linear force application until rupture. These values probably only have a small error because the plots did look linear from $t = 0$ to t_f.

CONCLUSIONS

The energy absorption capacity for the number of bones tested was relatively constant, ranging from 16,239 J/m^3 to 20,592 J/m^3. The polyamide impact results also yielded a relatively constant value with three of the specimen values between 16,847 J/m^3 and 28,347 J/m^3 and the other at 40,189 J/m^3. Therefore, it appears as if the energy absorption capacity is a good correlation parameter for "normalization" of specimen fracture behavior.

The close correspondence of the average energy absorption capacities of bone and polyamide gives support to the belief that fiber-reinforced polyamide might be a good choice as a bone simulant material under impact loading conditions. This would be true if the geometry of the polyamide specimens can be qualified so as to result in an ultimate failure force comparable to that of healthy bone. This belief is reinforced by the comparison of the mechanical properties of bone to those of different types of polyamide with varying fiber content and length. These values are shown in Tables 4 [22] and 5 [21,23].

The polyamide data under static and dynamic loading conditions sufficiently resembles that of bone so additional testing and work will continue. In addition to the comparable energy absorption capacity values, bone and polyamide specimens seem to produce similar types of fractures after impact loadings.

Table 4[a]

Effects of Fiber Length and Content on Properties of Reinforced Polyamide (Nylon)

Physical Property	ASTM test method	Unfilled nylon-6	30% Short glass fiber (<3.175 mm)	30% Long glass fiber (>6.350 mm)	40% Long glass fiber
density, g/cm^3	D 792	1.13	1.34	1.34	1.52
water absorption,%,24 hr.	D 570-577	1.6	1.1	1.1	0.8
tensile strength, MPa	D 638-58T	82	153	83	208
ultimate elongation,%	D 638-58T	200	2	5	5
flexural modulus, MPa	D 790-56T	2639	6944	6597	8333

[a]Encyclopedia of Polymer Science and Technology, 1988

Table 5[a]

Static Properties of Bone and Nylon 6-6 (Polyamide), 30-33% Short Glass Fiber Reinforced

Tensile Physical Property	Bone	Fiber-reinforced Polyamide
E, Modulus of Elasticity, Pa	1.89×10^{10}	8.9×10^{9}
σ_y, Yield Stress, Pa	1.26×10^{6}	170,352,000

[a]Cowin, 1989; Modern Plastics Encyclopedia, 1988

REFERENCES

1. Yamada, H. Strength of Biological Materials. Evans, F.G., Ed., Williams & Wilkens, Baltimore: 1970.
2. Evans, F.G. Mechanical Properties of Bone. Charles C. Thomas, Springfield, Illinois: 1973.
3. McElhaney, James H. "Dynamic Response of Bone and Muscle Tissue." ed. Sara F. Leslie. Bethesda, Md. Journal of Applied Psychology, 21(4), 1966, pp. 1231-1236.
4. Currey, J.D. "Differences in the Blood Supply of Bone of Different Histological Types." Q.J. Microsc. Sci., 101, 351, 1960.
5. Reilly, D.T. and A.H. Burnstein. "The Mechanical Properties of Cortical Bone." J. Bone Jt. Surg., 56A, 1001, 1974.
6. Carter, D.R. and D.M. Spengler, "Mechanical Properties and Composition of Cortical Bone." Clin. Orthop., 135, 192, 1978.
7. Fung, Y.C. Biomechanics: Mechanical Properties of Living Tissues. New York: Springer-Verlag New York, Inc., 1981.
8. Van Buskirk, W.C. and R.B. Ashman. "The Elastic Moduli of Bone." In ed. S.C. Cowin, Bone Mechanics. New York: American Society of Mechanical Engineers, p.45.
9. Cowin, S.C., et al. "The Properties of Bone." in Handbook of Bioengineering. Skalak, R. and S. Chien, Editors-in-Chief, McGraw-Hill, New York, 1987, 2.1.
10. Cowin, S.C. Bone Mechanics. Boca Raton, Florida, CRC Press, Inc., 1989.
11. Timmins, P.A., and J.C. Wall. "Bone Water," Calcif. Tissue Res., 23, 1, 1977.
12. Lang, S.B. "Ultrasonic Method for Measuring Elastic Coefficients of Bone and Results on Fresh and Dried Bovine Bones." IEEE Trans. Biomed. Eng., 17, 101, 1970.
13. Reilly, D.T., et al. "The Elastic Modulus for Bone." J. Biomech., 7, 271, 1974.
14. Relly, D.T. and A.H. Burstein. "The Elastic and Ultimate Properties of Compact Bone Tissue," J. Biomech., 8, 393, 1975.
15. Yoon, H.S. and J.L. Katz. "Ultrasonic Wave Propagation in Human Cortical Bone: Measurements of Elastic Properties and Micro-Hardness." J. Biomech., 9, 459, 1976.
16. Van Buskirk, W.C., et al. "Ultrasonic Measurement of Orthotropic Elastic Constants of Bovine Femoral Bone." J. Biomech. Eng., 103, 67, 1981.
17. Ashman, R.B., et al. "A Continuous Wave Technique for the Measurement of the Elastic Properties of Bone, J. Biomech., 17, 349, 1984.
18. Knets, I., and A. Malmeisters. "Deformality and Strength of Human Compact Bone Tissue." Mechanics of Biological Solids. Proc. Euromech Colloquium 68. Ed. G. Brankar. Sofia: Bulgarian Academy of Sciences, 133, 1977.
19. Yanson, K.A., et al. "Physiological Significance of Changes in Bone Volume Associated with Deformation." Polymer Mech. USSR, [Mekh. Polim.]: 10, 695, 1974.

20. Carter, D.R. and W.C. Hayes. "Bone Compressive Strength: the Influence of Density and Strain Rate." Science. 194, 1174, 1976.
21. Motishima, T. "Studies on the Strength for Bending of Human Long Extremity Bones." The 64th Japan Anatomy Conference, Kyota, Japan: The Kyoto Prefectural University of Medicine, 1959.
22. "Polyamide Plastics." Encyclopedia of Polymer Science and Technology, 1969, 10, 466.
23. Modern Plastics Encyclopedia, 1988.