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FOREWORD

The 1981 Advances in Bioengineering provides an annual forum for the publication and timely dissemination of technical information of the Bioengineering Division of ASME. The extended abstracts contained in this symposium volume have been screened for technical suitability. Contributors are encouraged to develop a full-length article on their technical achievements and submit the manuscript to our division's journal, Journal of Biomechanical Engineering, edited by Dr. Y. C. Fung.

The Bioengineering Division of ASME is pleased to publish an invited paper by Dr. F. Gaynor Evans, 1980 Herbert R. Lissner Awardee in Biomedical Engineering. Dr. Evans has generously contributed a summary of relevant biomechanics with new perspectives on the historical achievements. It is fitting that his paper is first in this symposium volume and it is with deep appreciation that the Division publishes his article: "A Review of Some Pioneering American Research in Biomechanics."

A technical session has been developed on cardiovascular pathophysiology as related to prosthetic valve performance and evaluation. Paul D. Stein, M. D. has graciously accepted the Division's invitation to make a keynote presentation for the session summarizing the current bioengineering opinion on cardiovascular pathophysiology.

Two technical sessions at the 1981 WAM are co-sponsored with the Heat Transfer Division: "Heat and Mass Transfer in Living Systems" and "Cryogenics in Biomedical Engineering". Papers in these sessions are published as individual, full-length papers available at the Winter Annual Meeting and are only referenced in this symposium volume. Our other sessions cover a wide range of bioengineering activities, including Bioinstrumentation and Devices, Bioheat and Transport, Biomechanics, Tissue Mechanics, Kinematics and Bioengineering. Published papers are in compact format with a limit of two pages of text and two pages of figures/tables.

I would like to extend my appreciation to the management of the General Motors Research Laboratories for the opportunity to develop the technical program in the 1981 Advances in Bioengineering for the American Society of Mechanical Engineers. I would also like to extend my appreciation to many of the staff of the Biomedical Science Department, GM Research Laboratories, who assisted in finalizing the technical program, especially Ms. Jody Simonetti.

David C. Viano
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THE HERBERT R. LISSNER AWARD IN BIOMEDICAL ENGINEERING

Dr. Max Anliker
Professor of Biomedical Engineering
University of Zurich

Dr. Max Anliker, Professor of Biomedical Engineering and Director of the Institute for Biomedical Engineering, University of Zurich, Zurich, Switzerland, has been awarded the 1981 H. R. Lissner Biomedical Engineering Award for his outstanding contributions to the field of biomedical engineering.

Max Anliker was born in Zurich on December 25, 1927. He received his M. S. in Experimental Physics in 1952 and his Ph. D. in Natural Sciences in 1955 from the Swiss Federal Institute of Technology. From 1951 to 1955 he was a teaching and research assistant in mechanics at the Swiss Federal Institute of Technology.

In 1955 he joined the Polytechnic Institute of Brooklyn as a post-doctoral research associate, and from 1956 to 1958 he worked as a senior research engineer at the Convair Division of General Dynamics in San Diego. Between 1958 and 1971 he was a faculty member of the Department of Aeronautics and Astronautics at Stanford University, where he initiated a research program in biomechanics and engaged in a close collaboration with the Ames Research Center of the NASA in the area of biomedical engineering.

In 1971 Dr. Anliker accepted an offer from his alma mater and the University of Zurich to build up and direct a new institute for biomedical engineering. As a professor of the medical school of the University of Zurich and a faculty member of the Swiss Federal Institute of Technology his academic responsibilities also reflect the partnership of the two universities in the Institute for Biomedical Engineering.

During the past 10 years he directed and actively participated in a number of research projects in biomedical engineering which were jointly pursued with various departments of the medical school. These efforts were primarily focussed on the development of the new noninvasive methods of quantifying relatively small changes in the geometric and physical properties of biological structure and in the functional parameters of such structures. The motivation for projects with such a goal is to provide the means for documenting the gradual development of diseases and their response to treatment and to provide a better basis for an increased prevention of diseases such as strokes, pulmonary embolisms, osteoporosis and cancer with metastasis by way of an early diagnosis and subsequent treatment.

Dr. Anliker is author of 104 full-length publications and 90 short-papers and abstracts. He presented the "Konrad Witzig Memorial Lecture" in 1976 and the "Alza Distinguished Lecture" in 1977. He is a member of the American Physiological Society, the Biomedical Engineering Society and the American Society for Engineering Education.

THE HERBERT R. [illegible] AWARD IN BIOMEDICAL ENGINEERING

Dr. Max Anliker
Professor of Biomedical Engineering
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Dr. Max Anliker, [illegible] of [illegible] Engineering and Director of [illegible] Biomedical Engineering, [illegible] University [illegible] Switzerland, has been awarded the 1981 [illegible] Biomedical Engineering Award for his outstanding [illegible] to the field of biomedical engineering.

[illegible] A[illegible] was [illegible] on December 20, 19[illegible]. He received [illegible] in theoretical Physics in 1952 and his Ph.D. [illegible] Sciences in 1958 from the Swiss Federal [illegible] Technology. From [illegible] to 1956 he was [illegible] research assistant in [illegible] at the [illegible] Institute of Technology.

In 1956 he [illegible] from 1956 to 1960 [illegible] research engineer at the [illegible] of General Dynamics in San Diego. Between [illegible] and 1971 he [illegible] member of the [illegible] at Stanford [illegible], he initiated a research program [illegible] with [illegible] NASA in the area of [illegible] engineering.

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A REVIEW OF SOME PIONEERING AMERICAN RESEARCH IN BIOMECHANICS

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The term "Biomechanics" appears to have been first used in American literature by Gratz (11), an American surgeon, in a publication entitled "Biomechanics. A new method of studying physical disabilities."

Trabecular orientation, as seen in sections of intact bones, was the first aspect of bone structure to be studied from the biomechanical viewpoint. The first American attempt was made by Wyman (40) who interpreted the trabecular arrangement in femoral, vertebral, talar, and calcaneal sections as "studs" (compressure resisting) and "braces" (tension resisting) bars. His paper was originally presented at the Boston Society of Natural History in 1848.

After this there was little interest in the biomechanics of bone until Koch (31) published his classic paper on "Laws of Bone Architecture". In this paper Koch calculated the stress and strain, under assumed vertical loads of 100 lbs. (standing), 160 lbs. (walking) and 320 lbs. (running), in 75 cross sections of the right femur of a 35 year old, 200 lb. Negro man who was accidentally killed. Koch was highly qualified for his study as he was a practicing professional engineer for many years before getting his M.D. at Johns Hopkins. Afterwards he interned at Henry Ford Hospital and was in general practice in Detroit until his death. The author knew Koch and got him to have a demonstration at the 64th annual meeting of the Amer. Assoc. of Anat. in Detroit in 1951.

Carey (1) made x-ray studies of cancallenous bone organization in different planes of hip, knee, and ankle joints plus intact pelvis and hallux. He believed that trabeular orientation depended on back pressure vectors produced by muscles acting on the joints.

Interest in biomechanics had a great surge in the 1940's when a group of engineers (Herbert R. Lissner, Lawrence M. Patrick, and Milton Lebow), neurosurgeons (Elisha S. Gurdjian, et al.) and an anatomist (F. Gaynor Evans) at Wayne State University, Detroit, Michigan, began a series of collaborative studies on injury mechanisms, under various controlled conditions, with techniques from the Sciences of Experimental Stress Analysis and Materials Testing.

Gurdjian and Lissner (12) were the first to study the mechanism of head injury with strain guages, and a cathode ray oscilloscope. Skull deformation and intracranial pressure changes, at the time of head injury, were studied in living anesthetized dogs. Skull deformation was measured with Sr-4, Type-5, electric strain guages, with a half inch guage length and a resistance of 350 ohms, bonded directly to the left parietal region of the skull. Intracranial pressure changes in the parietal region of the brain and in the cerebrospinal fluid were measured with small plastic "pressure plugs" screwed into drill holes in the right and the left temporal region of the skull. Two platinum wire electrodes traversed the plugs lengthwise and passed through the opened dura so they were in contact with the brain and the cerebrospinal fluid. Pressure readings from both sides of the skull were taken simultaneously by an electronic switch operating at a frequency of 1,000 cycles per second.

Intracranial pressure change measurements were based on the fact that compression of an electrolyte decreases its electrical resistance while pressure increases near the "pressure plug" is accompanied by a decrease in the electrical resistance of the cerebrospinal fluid between the platinum electrodes. Electrical resistance changes were recorded on an oscilloscope. Electrical resistance changes in the strain guages were measured with a potentiometer circuit.

The strain guage data showed that skull compression, from inbending of the bone, occurs from a blow near the guage while tension, from outbending of the bone, occurs on the opposite side of the skull. Approximately 1/2000 sec was required for maximum skull deformation. Varying the intensity of the blow increased the magnitude of the deformation but the time for the initial deformation remained essentially the same. Thus, the time period (1/2000 sec) for the initial deformation is independent of the intensity of the blow but is dependent on the size and shape of the skull.

After a hammer blow to the skull on the same side as the "pressure plug" there was an initial pressure decrease in the area of the plug, followed, in the case of the skull, by 2 to 4 pressure oscillations before equilibrium is reached. Simultaneously there was a pressure increase in the region of the "pressure plug" on the opposite side of the skull. About 1/1400 sec, on the average, was required to reach the initial pressure peak. This difference is due to the facts that maximum acceleration of the skull usually does not occur until after maximum acceleration. Because the pressure and deformation waves are out of phase they dampen each other and bring both pressure and deformation to equilibrium in a short time period.

The above experiments showed that at least two factors are involved in head injury - (1) skull deformation and (2) increased intracranial pressure giving rise to dynamic stresses caused by pressure gradients.

Gurdjian and Lissner (13) were also the first to study bone deformation with "Stresscoat", a strain sensitive lacquer that cracks in response to tensile strain in the material upon which the lacquer has been sprayed. Cracks in the "Stresscoat" lacquer are oriented transverse to the direction of the strain and arise first in the region of highest tensile strain where failure of the material will occur if sufficient force is applied to it. For photographic purposes the lacquer cracks, constituting a "stresscoat" pattern, were traced with India ink. The first problem investigated was the correlation between "Stresscoat" patterns on (1) the skull of living dogs and monkeys under nembutal anesthesia; (2) the skull of the same animal dead with undisturbed intracranial contents, and (3) the dry skull of the same animal with the intracranial contents removed.

The study showed that, under the same experimental conditions, the "Stresscoat" pattern was essentially similar in each of the three conditions. The main difference noted that the extent of the pattern in the living condition was greater than that in the dead condition or in the dry skull. From these experiments Gurdjian and Lissner concluded that valid deductions, regarding the biomechanical behavior of living bones, could be drawn from "Stresscoat" patterns on dry bones.

They also studied "Stresscoat" patterns in 5 dry human skulls and 3 cadaver skulls produced by blows in different regions of the skull.

From their Stresscoat experiments Gurdjian and Lissner concluded that (1) the Stresscoat method is well suited for studying skull deformations in head injury at the site of impact as well as in remote areas, (2) Stresscoat patterns on dry bones are useful for determining tensile strain and areas of weakness in living bones under dynamic loading, (3) tensile strain deformation may be greater some distance from than they are at the site of impact, (4) deformations of the skull base, from a blow on the occiput, have sufficient magnitude to cause pressure waves in the brain stem and medulla, (5) bone fracture occurs from failure of the bone from tensile stress and may originate on either the outer or the inner surface of the skull, and (6) strain propagation characteristics are a function of skull shape and thickness differences which may cause variation in the strain patterns in different skulls.

The mechanical behavior of the skull and its contents when subjected to injuring blows was discussed by Lissner and Gurdjian (32). Data obtained from SR-4 strain guages cemented to the skull of living anesthetized dogs showed that the time for maximum deformation was 0.0005 sec and that equilibrium was reached in 0.005 sec after the blow. Maximum pressure in the brain and cerebrospinal fluid was reached 1/1400 sec after the start of the blow.

Quantitative determinations of the amount of absorbed energy required to produce a threshold Stresscoat pattern in various regions of dry human skulls were also made by Gurdjian and Lissner (14). Energy was applied to the skull by dropping it on a 160 lb. steel slab and catching it on the rebound.

When the blow was applied to the midfrontal region of the skull 14 in. lbs. to 18 in. lbs. of energy were required to produce a stresscoat pattern of a magnitude similar to that produced in the mid-occipital region by a blow of 8 in. lbs. of energy. Sometimes a small amount of energy was dissipated in deformation and fracture of the skull. Contrecoup deformations of the fronto-spheno-temporal regions resulted from a midoccipital blow in only one skull.

The distribution and direction of the Stresscoat deformations generally parallels clinical fractures. However, deformations may be given directional differences by variations in skull shape and the velocity of the injurying object.

In another study Gurdjian and Lissner (15) used "stresscoat" to study tensile strain patterns and fractures produced in human skulls by (1) hammer blows to various regions of the skull, (2) blows of constant energy, and (3) blows of varying energy. They found that maximal tensile strain patterns were produced by blows to different parts of the skull and that tensile deformation is greater at some distance from the impact point. Skull deformations produced by blows of a given amount of energy in the occipital region are far more severe than those resulting from similar blows to the forehead. The "stresscoat" patterns indicate some differences in behavior of different skulls but the direction is the same. Stresscoat patterns in dry and in living bone are essentially the same.

The mechanism of linear fractures in fresh cadaver skulls was investigated by Gurdjian, Lissner, and Webster (18). The skulls were "stresscoated" inside and out and then dropped on the steel slab, used in earlier experiments, in such a way that the point of skull-impact was vertically below the center of gravity of the skull. As before the skull was caught by hand on the rebound so that there was only one impact. In each of the 25 tests skull weight (lbs.), distance dropped (inches), lacquer sensitivity (in/in) and site of impact were recorded.

The "stresscoat" patterns on the outside and the inside of the skull indicated that they arose from tensile strain created in each region by outbending and inbending of the bone, respectively. On both the outer and the inner aspect of the skull the cracks forming the "stresscoat" pattern were oriented in a radial direction around the impact site. On the outer surface of the skull, where outbending occurred as a result of the impact, the stresscoat pattern was not continuous around the impact site and was at some distance from it. On the inner surface of the skull, where inbending occurred, the stresscoat pattern was continuous around the impact site and was close to it.

Many of the linear fractures on the outer surface of the skull were initiated by outbending of the bone, at some distance from the point of impact, as predicted from the "stresscoat" pattern in the area. Close correspondence between the location and direction of the fractures and experimentally produced "stresscoat" patterns indicates that the fractures arose from failure of the bone from the tensile stresses and strains created within it by bending. From the location of the site of impact the direction of the fracture line may be deduced and vise versa.

Skull fracture, with special reference to engineering factors, was studied by Gurdjian, Webster, and Lissner (19) in 55 intact cadaver heads. Each head was weighed (lbs) as dropped, and the thickness (mm) of the scalp was measured. Then, the distance (in) through the head was dropped, the deceleration impact velocity (ft/sec) and the energy (in. lbs) applied to the head in a test were measured. After testing the dry weight (lbs) of the head was determined.

Deceleration impacts were applied to (1) the mid-frontal region of 10 heads (9 white, 1 black), (2) the anterior interparietal region of 17 heads (11 white, 4 black, 1 Filipino, 1 unknown), (3) to the occipital region of 9 heads (8 white, 1 black), and (4) to the posterior parietal region of 10 heads (8 white, 1 black, 1 unknown). The impact was applied by dropping the heads unto the steel slab. The heads were caught by hand on the rebound so they struck the slab just once.

Energy required for a single fracture in each of the four regions of the skull varied from 400 to 900 inch pounds. In one skull no fracture was produced with 1000 inch lbs. Regional differences in energy required for fracture are probably unimportant because of the large energy variations found for impacts to a single region. Average energy required for a single linear fracture was 571 in. lbs. in the frontal midline region; 517 in. lbs. in the back midline region; 710 in. lbs. in the top midline region, and 615 in. lbs. above each ear. Energy necessary for fracture in the midfrontal midline regions varied from 425 in. lbs. to 803 in. lbs. Average energy required for a single fracture and for complete destruction were close. The average energy for fracture in each region because of the small number of specimens, is unimportant but the fact that about 400 in. lbs. of energy was the least required for fracture in 6 of the heads is important.

One of the most interesting results of these experiments was the discovery that after enough energy was absorbed to produce one linear fracture very little additional energy was required to produce multiple fractures or complete skull destruction.

Differences in skull and scalp thickness, skull shape, and slight variations in point of impact account for the large variations in energy required for fracture. In a dry skull fractures were produced with only 40 in. lbs. of energy while more than 400 in. lbs. were necessary for fracture in an intact head. This difference is probably due to the great energy absorbing capacity of the scalp.

In order to compare their cadaver head results with those of a living head the authors estimated the energy applied to a batter's head when hit by a fast ball. Assuming a speed of 100 ft/sec in a fast pitch of a 5 ounce baseball its kinetic energy will be 580 in. lbs. Some of this energy will be dissipated in deforming the ball and accelerating the head. However, since the ball is a curved surface, less energy is needed to produce fracture because of the localized contact area. Linear skull fractures and severity of cerebral damage have no direct correlation. Fatal concussion may occur without skull fracture.

Data from stresscoat tests, as well as from experimental and clinical fractures, show some variations in fracture site produced by a blow in a particular region. This is due to individual skull variations and position of the blow.

The degree of this variation was investigated in 62 skulls in which the top of the skull was divided into 12 areas. A stresscoat pattern was obtained in each of the 62 skulls by applying blows in each of the 12 areas. The left parietal region showed the least variation is the predicted position of the fracture line.

This study showed that it should be possible to predict the location of a fracture fairly accurately if the site of the blow is known. Conversely, if a fracture is visible in an x-ray film the site of the blow may be determined.

Time required for absorption of the energy of a blow was determined with an SR-4 strain guage, mounted in the temporal region across the line of the expected fracture. Blows were applied to the head above the ear. As the skull was dropped it struck a switch on the steel slab which was connected to one beam of a two-gun cathode ray tube so the initial contact of the skull with the slab was indicated. The elapsed time from contact of the skull with the slab was indicated. The elapsed time from contact of the head with the slab and the beginning of skull deformation was 0.0006 sec. The skull bone itself required an additional 0.0006 sec until it fractures.

In a followup paper of the mechanism of skull fracture Gurdjian, Webster and Lissner (20) determined areas of different stress level in the skull. Thus, the first set of stresscoat cracks from a posterior parietal blow were in the temporal region; with a more severe blow a second set of cracks appeared in the parietal region superiorly and with an even stronger blow a third set of cracks occurred in the mastoid-parietal region posteriorly and inferiorly. The first set of cracks (temporal region) indicated the area of highest tensile strain where the first single linear fracture would occur. With increasing energy the second linear fracture would be in the region of the second set of cracks, the third fracture in the region of the third set of cracks, etc. This process could continue until a stellate fracture, consisting of 5 or 6 linear fractures radiating around the site of impact was obtained. After this had occurred a slight increase in the energy of a blow would completely destroy the skull.

With increasingly higher velocity impacts depressed fractures, rather than linear ones, are produced. The higher the velocity of the impact, the greater is the possibility of depression fractures or perforation. Six varieties of depression fractures were identified on the basis of the velocity, kinetic energy, and, to a lesser extent, the shape of the injurying object.

Gurdjian, Webster and Lissner (21) extended their study of stresscoat patterns and fractures produced by blows in the (1) midfrontal, (2) anterior interparietal, (3) posterior interparietal, (4) mid-occipital, (5) frontal, lateral to the midline, (6) anterior parietal, (7) posterior parietal, and (8) parieto-occipital areas of the human skull. Each area, left side of skull only, was approximately 2 x 3 in. The mechanism of linear, stellate, communuted, and depressed fractures was described. A total of 100 skulls were used in their studies (17, 18). Clinical fractures, seen in Dr. Gurdjian's patients, were compared with the experimentally produced ones.

Blows to the anterior and the posterior areas of the skull produced side to side tearing apart forces at the skull base and fractures extending throughout the skull base in an anterioposterior or opposite direction. Parietal or frontal blows produce similar forces, anterioposteriorly, in the skull base. Parietal and lateral frontal blows cause side to side fractures of the base. From data obtained in these experiments the clinician, provided the position of the blow is known, should know the location and direction of a possible fracture. Conversely, from the position of the fracture the site of the blow can be determined.

Later Gurdjian, Webster and Lissner (22) discussed the type, location and percentage of fractures that can be predicted from blows to each of the 8 regions of the skull described in previous studies. They pointed out again that clinical and experimental evidence show that linear skull fractures occur at right angles to the maximum tensile stress produced by outbending of the skull at a distance from the site of fracture.

From studies on skull deformation and fracture with the Stresscoat method, attention was turned to an investigation of other aspects of head injury.

Acceleration and intracranial pressure in experimental head injury were determined in living anethesized dogs by Gurdjian, Lissner et. al. (25). Acceleration was measured with a Statham $\pm$ 500 G accelerometer with a natural frequency of 1600 cycles/sec, attached to the skull on the side opposite to that where the blow was applied. Intracranial pressure changes were measured with a specially designed pressure plug mounted in a tapped hole in the skull on the same side as the accelerometer. Accelerometer and pressure plug signals were amplified and passed into a dual beam oscilloscope where they were recorded with a high speed oscilloscope recording camera. Blows were applied to the skull, opposite the side where the instruments were attached, with ball preen hammers of various weights. A 15 lb. hammer was most commonly used. Sometimes the acceleration given the head was changed by putting rubber pads over the head of the hammer.

Results were reported for 24 experiments in which intracranial pressure, acceleration, and the time duration were simultaneously measured. Accelerations varying from 250 G to over 500 G produced minimal, moderate and severe concussive effects. At the same time intracranial pressure rise varied from 25 psi to 95 psi.

The significant factors explaining clinical effects after impact appeared to be the time duration of the pressure and the acceleration. At lower pressure and acceleration values concussion occurred only with appreciable pressure and acceleration duration. With higher pressure and acceleration concussion was produced in a shorter time.

The concussion study were continued in 72 experiments on dogs (Gurdjian, Lissner et. al. (26). Air pressure, for a controlled duration, was applied to the dural sac by a special device attached to a trephine hole in the parietal region of the skull. The device consisted of a cylinder with a spring operated piston. Releasing the spring moved the piston through the cylinder. First, there was a given magnitude of air pressure in contact with the dural sac. Then, as the piston continues to move, the pressure escapes simultaneously with a stop in the air pressure flow. Various speeds were attained by changing the piston mass and the spring size. The pressure and its duration were determined with a small diaphragm type strain guage pressure pickup and photographic records made with a dual beam oscilloscope. Pressure, from a tank of compressed air, was obtained through an adjustable pressure-reducing valve. The tank had a maximum pressure of 120 psi. However, this amount of pressure could not be applied to the dural sac and transmitted through the brain in a very short time because of the size of the hole for the air and the volume to be filled. There was, however, a definite trend between physiological effects and pressure duration.

Results of the experiments were classified as (1) no concussion, (2) threshold concussion, (3) moderate concussion, and (4) severe concussion. The pressure (psi) and time (sec) required for each type of concussion were recorded for each experiment.

In the "Compressed Air" concussion experiments no concussion occurred in 23 dogs with a pressure varying from 4 psi to 33 psi and time from 0.0008 sec to 0.027 sec. Threshold concussion was produced in 25 dogs with 5 psi to 36 psi and a time from 0.00064 sec to 0.045 sec. Moderate concussion was found in 11 dogs with a pressure from 20 psi to 40 psi and a time range from 0.0010 sec to 0.016 sec. Severe concussion was seen in 12 dogs with a pressure variation from 11 psi to 74 psi and a time range from 0.0012 sec to 0.12 sec.

In additional experiments the same four degrees of concussion were studied by dropping weights on a water column in contact with the dura of dogs. In three dogs no concussion occurred with a pressure variation from 19 psi to 22 psi and a time range from 0.0076 sec to 0.013 sec. Threshold concussion was produced in 6 dogs with a pressure from 12 psi to 28 psi acting for a time range of 0.01 sec to 0.019 sec. In 3 dogs moderate concussion occurred with a pressure range of 21 psi to 23 psi and a time variation from 0.0093 sec to 0.016 sec. Severe concussion occurred in 7 dogs with a pressure variation of 23 psi to 50 psi and a time from 0.004 sec to 0.028 sec.

Conclusions drawn from the experiments were (1) the shorter the time duration the higher the pressure required for concussion; (2) the longer the time duration the lower the pressure necessary for concussion; (3)concussion from acceleration, deceleration, or compression is caused by intracranial pressure increase at the time of impact; (4) with compression intracranial pressure increase is more uniform throughout; and (5) acceleration or deceleration injuries produce increased intracranial pressure with pressure gradients in the cranial cavity.

The relation of acceleration to physiologic effect in experimental concussion was investigated by Haddad, Lissner et. al. (27) in 34 dogs. Initially a 4.3 oz. unbonded wire type accelerometer with a range of $\pm$ 500 G and a natural frequency of 1,600 cycles per sec was used. The duration and magntidue of the acceleration were measured. Later, in order to eliminate any error caused by instrumental inertia and to measure value larger than $\pm$ 500 G, a 0.8 oz. barium titanate type accelerometer was used to measure acceleration linearly within and beyond the limits of the experiments. Only peak acceleration, not duration, was measured with the second accelerometer. The two accelerometers were fastened to the same base, so the results could be compared, which was then given an accelerating blow. Acceleration linearly, with respect to accuracy, was measured by both accelerometers. The accelerometer was attached to the skull on the side opposite to the blow which was given with ball preen hammers. Acceleration was varied by padding the hammers to prevent crushing of the skull. Electrical signals from the accelerometers were amplified, led into a dual beam cathode ray oscilloscope and records obtained with a high speed oscilloscope camera. Blood pressure and respiration were recorded.

A concussive effect was considered to be present if definite vasomotor and respiratory effects were present. Peak acceleration (G) and total acceleration duration (sec) were measured. The degree of concussion was classified as (1) none, (2) minimal, (3) moderate, and (4) severe.

No concussion occurred in 9 dogs with peak acceleration varying from 113 G to 750 G. Minimal concussion was found in 12 dogs are a variation in peak acceleration from 130 G to 780 G. Moderate concussion was produced in 3 dogs with a range from 250 G to 650 G in peak acceleration. In 10 dogs severe concussion occurred with peak accelerations from 72 G to 650 G.

The authors concluded that there was little relation between peak acceleration and the degree of concussion. However, concussive effects as the result of intracranial pressure gradients may be produced by acceleration. The duration of the resultant pressures are responsible for concussive effects. High accelerations are necessary to produce high intracranial pressures of sufficient duration to produce concussive effects.

Experimental data were combined with clinical observations by Gurdjian, Webster and Lissner (23) in a study of the mechanism of brain concussion, contusion, and laceration. Plots of degree of concussion against duration of acceleration showed a linear relation between them. Thus, higher pressures for shorter times produce effects similar to those occurring at lower pressures for longer times. A pressure-time plot showed that the greater the pressure the shorter the time required for severe concussion.

The authors pointed out that cerebral concussion, contusion, and laceration do not denote varying degrees of neurological damage, concussion being the mildest form. Concussion appears to be due to brain stem involvement. A brain stem contusion or laceration that may be fatal can occur in other parts of the nervous system with no concussive effect.

Later Gurdjian and Lissner (16) studied the position and motion of the head impact in anesthetised dogs and human cadavers. The head of dogs and of human cadavers were impacted with 2 lb. and 15 lb. rotary hanners with the head relatively fixed or free to move. In many tests the position of the head at and during impact were studied by standard (16 or 24 frames/sec) and high speed cinematography (6,500 frames/sec). Acceleration and deceleration studies of cadaver heads were made with a vertical accelerator in an elevator shaft and flexion-extension injuries of cadavers with a horizontal accelerator.

High speed movies revealed that the head of an experimental animal moves several millimeters when the head is relatively fixed. Such movements impart high accelerative forces to the head. Flexion and extension at the head-spinal junction appear to be a minor factor in concussion which can occur without any apparent movements of the head and neck.

High speed movies show the potential for multiple injuries of the head and body in an accident. Several head impacts may follow one another during a few milliseconds after impact.

The relation of energy, velocity, and acceleration, under controlled experimental conditions, to skull deformation and fracture of embalmed adult human cadaver heads was studied by Evans, Lissner and Lebow (7). The heads were dropped through various measured distances, onto Ford automobile instrument panels placed in the bottom of a vacant 8 story elevator shaft. The impact was applied to the forehead, above the glove compartment, as often happens in an automobile accident. Roentgenograms were taken of each head, before and after testing, to determine if a fracture had occurred during a test. Velocity of the head was determined from high speed motion pictures (1,000 frames/sec) taken of the head at impact. A total of 27 tests, divided into 10 free fall, and 17 guided fall tests was made.

In the free fall tests the head was simply dropped, through a measured distance, onto the instrument panel by cutting the supporting cord. In these tests the total available kinetic energy and the velocity, computed from theory, of the heads at impact varied from 134 to 426.7 ft. lbs. of energy and from 32.5 to 52.5 ft/sec of the velocity. Head weight ranged from 6.88 to 10.0 lbs. and the distance dropped from 16.25 to 42.67 ft. In this series of tests the head of a Negro man was tested 5 times but no fractures were produced in three tests.

The first fracture, in the head of a 59 year old white man, was considered "no test" because the head bounced off the instrument panel onto the concrete floor.

A fracture, in the frontal and right temporal bones of the head of a white man of unknown age was produced by an impact velocity of 52.4 ft/sec (36 mph) and a total available kinetic energy of 318.1 ft. lbs.

An extensive stellate type fracture of the frontal and parietal areas, extending bilaterally to the anterior and middle cranial fossea, was produced in the head of a 57 year old white man at an impact velocity of 52.3 ft/sec (36 mph) and a total available kinetic energy of 348.6 ft. lbs. There was also a large depressed fragment in the frontal area.

In the 17 guided fall tests the head was suspended from a steel triangular frame which slid on 3 guide wires. A circular area of the scalp was removed from the occiput of each head so that a Statham 500 G capacity accelerometer could be screwed onto the occipital bone. The accelerometer was connected to the one channel of a two channel oscilloscope and the deceleration recorded with an oscilloscopic camera. The magnitude and duration of the acceleration were determined from the oscilloscopic record.

During the first 11 of the guided fall tests the triangular frame tended to bind so that the actual velocity and energy values were lower than the theoretical ones. Therefore, 6 additional tests were made with a modified frame that did not bind appreciably. In these tests an electronic switch, fastened to one of the guide wires 2.5 ft. above the impact point on the instrument panel, was opened by the passing frame. The switch was connected to the second channel of the oscilloscope and the actual velocity determined from the oscilloscopic record.

In the first 11 guided fall tests, made with the original triangular frame, the heads weighed from 6.63 to 9.50 lb. and were dropped through a distance of 55.88 ft. (5tests), 67.88 ft. (3tests) and 79.92 ft. (3tests). Impact velocity, measured from the high speed (1,000 frames/sec) motion picture film, varied from 43.5 to 59.2 ft./sec. (29 to 40 mph) in the 6 tests in which it was determined. Available kinetic energy, in the same tests ranged from 268 to 473 ft. lbs. Peak acceleration varied from 180 G to 555 G and the total time duration of the acceleration ranged from 0.00625 sec to 0.0117 sec in the 8 tests for which these data were determined.

The two fractures produced in these tests were in the heads of two Negro men of unknown age. In the first head there were 2 linear fractures in the middle part of the left occipital bone that extended into the parietal bone and the lateral margin of the foramen magnum. At impact the velocity was 60.0 ft/sec (41 mph) and the available kinetic energy 481.9 ft. lbs.

In the second head there was an extensive comminuted fracture of the frontal bone. At impact the velocity was 48.0 ft/sec (32 mph), and the available kinetic energy 268 ft. lbs. and the peak acceleration 555 G for a duration of 0.00903 sec.

In the 6 additional tests, using the modified non-binding triangular frame, the heads varied from 7.19 to 9.0 lbs. in weight and the distances dropped were 67.88 ft. (5 tests) and 79.92 ft. (1 test). The measured velocity ranged from 57.4 to 65.6 ft/sec (39 to 45 mph). The available kinetic impact energy, for the 5 tests in which it was measured, was from 455 to 581 ft. lbs. Peak acceleration in the 6 tests varied from 162 to 724 G with a time duration (5 tests) from 0.0019 to 0.01125 sec.

Two valid fractures were produced in these tests. The first one was an extensive comminuted and egg-shaped fracture on the parietal and right and left sides of the frontal bone of a 60 year old white man. The impact velocity was 64.6 ft/sec (44 mph), the available kinetic energy was 581 ft. lbs., and a peak acceleration of 724 G with a duration of 0.00338 sec.

A second extensive comminuted and depressed fracture of the left parietal bone, with a smaller fracture line between the left frontal bone, occurred in the head of a white man of unknown age. This head had an impact available kinetic energy of 561 ft. lbs. and a velocity of 63.3 ft/sec (43 mph). There was a peak acceleration of 344 G which lasted for 0.0488 sec.

These tests, free fall and guided fall, showed that the embalmed adult human head can tolerate without fracture peak impact accelerations as high as 686 G and a kinetic energy as great as 577 ft. lbs. Most of the available kinetic energy of the tests was expended in denting and deforming the instrument panel rather than acting on the head. The magnitude of the energy producing a fracture is approximately 33 to 75 ft. lbs.

One of the most important results of the study was revealed by the oscilloscopic records of deceleration at impact obtained in 14 tests. Comparison of the oscilloscopic record in 2 typical tests, one with and one without fracture, showed a fundamental difference between the two cases. In both tests the total time duration of the deceleration was 0.018 sec. In the non-fractured skull most of the deceleration occurred in slightly less than 0.010 sec. with the peak deceleration (368 G) at approximately 0.002 sec. In the fractured skull most of the deceleration was completed in less than 0.008 sec while the peak deceleration (555 G) was reached in a little less than 0.002 sec. Thus, in the head with fracture deceleration occurred in less time than in the head without fracture.

Although the peak deceleration occurred at approximately 0.002 sec in both heads the magnitude of the peak (555 G) in the head with fracture was considerably greater than the peak (368 G) in the head without fracture. This difference indicated that the rate of deceleration was greater in the fractured head than in the one without fracture. The oscilloscopic deceleration records clearly showed that both the magnitude of the kinetic energy of a blow as well as its duration or rate of absorption are factors to be considered in skull fracture. Thus, the longer is the time for energy absorption the greater the magnitude of the energy that can be safely tolerated.

Relations between acceleration and intracranial pressure changes in man were also investigated by Lissner, Lebow, and Evans (35). The problems involved in this study were (1) whether or not impact of the head against an automobile instrument panel was augmented by the body weight, (2) the relation between acceleration and intracranial pressure when the head struck a thin steel plate, an automobile instrument panel and a large steel block, (3) the time duration of pressure and acceleration during head impact, and (4) the probable intracranial pressure and its duration accompanying concussion.

Twenty-three tests, divided into 5 series of different types, were made with embalmed cadavers of Caucasian men of known weight. The cadaver was supported, at the armpits, by a frame work of aluminum channels fastened to a massive steel base by 2 ball-bearing-pillow blocks which allowed the cadaver and frame work to rotate freely in a vertical plane. At the beginning of a test the cadaver in an erect position within the frame which was fastened in position by web belts. The frame and cadaver were then allowed to fall forward, by free rotation about the base, so that the cadaver's forehead struck the test object.

Intracranial pressure changes were measured with 300 psi pressure guages mounted in holes in the right temporal and the left parietal regions of the head. A thin rubber gasket was slipped over the threaded end of the pressure guages and thickly coated with F-88 dental cement before they were screwed into the holes in the skull. Before the guages were mounted the brain was sucked out of the head, after all meningeal structures were ruptured, and replaced with fluid gelatin. The head was then packed in ice to set the gelatin. Just before testing the ice was removed. It was assumed that during a test the gelatin would transmit intracranial pressures in a manner approximately similar to that of the brain and other skull contents.

A circular piece of the scalp was removed from the occiput and a 1000 G capacity Statham accelerometer was screwed into the occipital bone to measure acceleration during a test. Straps around the head held the accelerometer in place. The edges of the incisions through the scalp were tightly screwed with overlapping sutures and the seams coated with celloidin to prevent embalming fluid from leaking into the areas where pressure guages and the accelerometer were mounted.

The recording instrumentation consisted of an oscilloscope equipped with a camera, for a permanent record, to which the pressure guages and the accelerometer were connected. Motion pictures, at a speed of 1,500 or 2,000 frames/sec were taken at the time of impact.

The cadavers were impacted at various accelerations and velocities, against (1) a flat steel plate with clamped edges, (2) a 3 inch thick pad on a steel plate, (3) steel block after a 3 ft drop, and (4) 1954 and a 1956 automobile instrument panel. Results of these tests were compared with those from impact tests with intact cadaver head dropped upon a 3 inch thick 160 lb. steel block. In the first 3 series of tests (drops 1-12) the cadaver's legs were amputated in the proximal one-third of the thigh but intact cadavers were used in the other tests. Test velocities were 10.9 mph and 15 mph.

A 3 foot drop on the steel slab had an impact velocity of 10.9 mph with an acceleration from 136 to

237 G. Parietal pressure was -15 to -25 psi and temporal pressure from 32 to 42 psi. The average duration of the pressure was 0.0006 sec. The reason for a negative pressure, greater than the standard atmosphere guage pressure at absolute zero, was the adhesions between the cranial contents and metallic diaphragm of the pressure guages.

Dropping the body on a flat steel plate with clamped edges at an impact velocity of 15 mph produced an acceleration from 104 to 159 G, a parietal pressure of -2 to -15 psi, a temporal pressure of 10 to 25 psi, and a pressure duration of 0.0005 to 0.010 sec.

Impacting the head against a 1954 automobile instrument panel at a velocity of 15 mph produced a variation of 65 to 86 G in acceleration, a parietal pressure of -6 psi, a temporal pressure of 6 to 16 psi, and a pressure duration from 0.008 to 0.016 sec.

A similar test with a 1956 automobile instrument panel reduced acceleration from 43 to 63 G, parietal pressure from -3 to -2 psi, temporal pressure from 5 to 15 psi, and the pressure duration from 0.0005 to 0.022 sec.

A 3 inch thick polyurethane pad on a flat plate, at an impact velocity of 15 mph, produced an acceleration of 84 G, and a temporal pressure of 12 psi for 0.016 sec. Striking the head against another polyurethane pad on a steel block produced an acceleration of 70 G, a parietal pressure of 6 psi, a temporal pressure of 16 psi, and a pressure duration of 0.012 sec.

Impacting a padded 1956 instrument panel at 15 mph produced an acceleration of 53 G, a parietal pressure of 3 psi, a temporal pressure of 9 psi and a pressure duration of 0.022 sec. Another test on the same panel, unpadded, produced an acceleration of 48 G and a temporal pressure of 14 psi lasting for 0.0016 sec.

A slightly depressed comminuted fracture was produced by dropping a head, through a distance of 3 feet, onto a solid steel block. The impact velocity was 10.9 mph, the acceleration -237 G, the temporal pressure 32 psi, the parietal pressure -15 psi, lasting for 0.006 sec. A linear fracture was produced in another head dropped on the same block at the same impact velocity.

In another study by Gurdjian, Lissner, Evans et. al. (24) cadavers were impacted against laminated and tempered automobile safety glass. Preparation of the cadavers for testing was essentially similar to that used in the previous study except that in one series of tests the accelerometer was fastened to the occipital bone with epoxy resin cement while in another series of tests it was held in place by a strap around the head. A total of 95 tests were made on 10 cadavers. In 6 cadavers the legs were amputated at the knees while in the other 4 the legs were doubled back and tied.

During a test the cadaver was tied, face down, on an aluminum carriage mounted on two vertical rails in a vacant elevator shaft. The carriage consisted of very light aluminum frame with web belts to support the cadaver which was tilted 30° to the horizontal so the head was down. An automobile door with the glass to be tested was mounted solidly below the carriage. The cadaver head was positioned to strike the center of the glass.

A cable, operated by an electric winch, was used to raise the cadaver to the necessary height to produce the desired impact velocity. Actual impact velocity was determined by the interruption of two electric circuits just before impact. A quick release mechanism was used to release the carriage which dropped down the elevator shaft guided by the rails. The head, as it was lower than the rest of the body, struck the glass before any part of the body contacted the door frame or any other structure. After the head hit the glass the body was brought to rest and its energy absorbed by a large block of styrofoam.

In the first series of tests 1/4 in thick laminated glass with a 15 mil plastic interlayer and 7/32 in thick tempered glass panels were used. In a second series of tests tempered glass 3/16 in thick, more representative of the then current automobiles, was used.

High speed motion pictures (1,000 to 3,000 frames per sec) were taken in front, bottom, and side views of the head at impact. In some tests a crystal accelerometer was mounted on the glass opposite to the point of impact.

During the tests the cadaver heads struck the glass panels at impact velocities of 2.5 to 34.8 mph with average occipital accelerations as high as 125 to 90 G for forehead midline blows to tempered and laminated safety. The duration of the accelerations varied from about 0.006 to 0.012 sec.

Average temporal pressure was a maximum of 20.9 psi, when the glass was unbroken, with the pressure lasting 0.006 to 0.012 sec. If the glass broke the pressures and their duration were very low. Parietal pressure was always lower than temporal pressure.

The high speed movies showed that the head oscillated 3 to 4 times after impact. Oscillations also occurred when the glass broke but they reduced to about an average of 2-1/2 oscillations. No skull or neck fractures or dislocations were seen when thinner glass panels were used.

Information obtained in the preceding three investigations was reviewed and summarized by Haynes and Lissner (29).

Protection of the head and neck in sports were discussed, from the biomechanical viewpoint, by Gurdjian, Lissner and Patrick (17). Data on injurying factors in athletics were reviewed and the mechanism of injury and the magnitude of the energy causing damage were discussed. They pointed out, on the basis of their previous work, that linear skull fractures in cadavers can be produced by impact energies of 4.6 to 6.9 kg.m which give an average acceleration of 112 G with an increase of about 1,450 mm. Hg. in pressure. These data can be the basis for safe helmet construction. The thickness of the necessary padding can be calculated from the weight and velocity of the injurying object. Doubling the velocity means the padding must be 4 times thicker.

The tensile and compressive strength of adult human parietal bone were determined by Evans and Lissner (4). Tensile and compressive test specimens consisting of outer and inner tables of compact bone separated by diploe, were machined to a standard size and shape from the parietal bone of adult embalmed human cadavers. Smaller compression test specimens were cut from the original larger tensile specimens. All tests were made with a 5000 lb. capacity materials testing machine calibrated to an accuracy of ±%.

The average ultimate tensile strength of 15 tensile speciments, loaded in their long axis, was 10,230 psi (6,030 - 15,800 psi). Average ultimate compressive of the compact bone of 69 specimens, loaded in the long axis of the original tensile specimens, was 22,080 psi (12,400 to 47,800 psi). However, the 56 specimens loaded perpendicular to the original long axis had an average compressive strength of 24,280 psi (4,500 to 46,900 psi).

Specimens from the right parietal bone had a 12% higher tensile strength than those from the left bone. The compressive strength, loaded in the long axis of the specimen, was practically identical for right and left bone specimens but, when loaded normal to the long axis, specimens from the left parietal bone had an average compressive strength of 16% higher than that of specimens from the right side. The average compressive strength of 23 specimens of spongy bone (diploe) was 3,640 psi (1799 to 5770 psi).

Parietal compact bone has a lower average tensile strength than long bone compacta but its compressive strength, both longitudinally and normal to the long axis of the specimen, is similar to that of other bones. However, the compressive strength of parietal bone exceeds its tensile strength by greater percentage than in long bone compacta. The compressive strength of diploe and femoral spongy bone is about equal.

This investigation appears to be the first determination of the mechanical properties of an individual skull bone.

LONG BONES

Professor Evans and Lissner (2) made a study, with the Stresscoat method, of the biomechanical behavior of adult human femurs under a static vertical loading. Ten femurs from embalmed adult human cadavers were tested in a Baldwin-Southwark materials testing machine calibrated to an accuracy of ± 1/2%. A special apparatus was fabricated to orient the femur with respect to the vertical (X) axis, the anteroposterior (Y), and the transverse (Z) axis. Each femur was weighed, and the diameter (in 2 planes) of the femoral neck and shaft was measured. In addition, measurements were taken of the vertical-neck angle, the neck-shaft angle, and the staft-vertical angle.

During a test the load was gradually applied by one of us while the other continually examined the bone for the first appearance of cracks in the Stresscoat lacquer. As soon as the first cracks appeared the machine was stopped and the magnitude of the load and the general distribution of the cracks were recorded. A total of 16 stresscoat tests was made.

After the test the cracks were covered with a red dye solution which after about a minute, was removed with an emulsifying solution. The dye remained in the stresscoat cracks, making them more visible. For photographic purposes the cracks were traced with India ink.

During a test the femur was seen to bend with gradually increasing load. However, after removal of the load the bone returned to its pretest condition. One bone supported a load of 650 lbs. without failure and returned to its original condition after removal of the load. This behavior indicates that the adult human femur under static vertical load behaves like an elastic body.

The first stresscoat cracks appeared on the upper aspect of the femoral neck, immediately distal to the head. With increasing load the pattern gradually spread distally along the upper surface of the neck.

Almost at the same time a stresscoat pattern was developing along the lateral aspect of the femoral shaft, starting a few centimeters distal to the greater trochanter, and then extending down the shaft. In some bones, with an noticable anterior bowing, the stresscoat pattern extended from the lateral side onto the anterior side of the shaft in about the middle third of the bone.

The location of the Stresscoat pattern indicated that, under static vertical loading, the superior (upper) aspect of the femoral neck and the lateral aspect of the shaft were subjected to tensile stress and strain as a result of bending of the bone by the load applied to it. The area where the stresscoat cracks first appeared indicated the region of highest tensile strain where failure could be predicted to occur with sufficient load.

A vertical fracture of the femoral neck, produced by a load of 1280 lbs., was initiated on the upper aspect of the neck, just distal to the head, where the first Stresscoat cracks appeared. The fracture was also parallel with the cracks constituting the stresscoat pattern on the neck of the bone. The point of initiation of the fracture and its directional correspondence with the stresscoat cracks indicated that it arose from failure of the bone under the tensile stress created by bending of the bone.

The average load before the first stresscoat cracks appeared was 720 lbs in the first series of 6 tests and 646 lbs in the second series of 10 tests. Femurs from people over 60 years of age supported a smaller load before appearance of Stresscoat cracks than did femurs of younger people.

In another study Evans, Lissner and Pedersen (9) tested 14 femurs from adult male cadavers under dynamic vertical loading of 15.8 in lbs. of energy. They found that the location of the stresscoat patterns was essentially the same as those found under static vertical loading. Thus, 15.8 in. lbs. of energy produced tensile deformation in the same regions of the femur as do statically applied loads of 400 lbs. to 715 lbs.

Deformation studies of the adult human femur in various types of loadings and orientations were reported by Pedersen, Evans and Lissner (38). Fifteen bones were tested under a vertical dynamic load of 23.7 in. lbs. of energy when the infracondylar plane made a laterally opening angle of 3° with the horizontal plane. Twelve bones were tested in the "abduction" position and 12 in static torsion loading.

Minor variations in bone orientation had little effect on the deformation patterns. If all other factors are equal the degree of deformation increases with increasing load but varies inversely with the mass of the bone.

Additional evidence shows that fracture occurs under tensile stress, the fracture originating at the site of greatest stress concentration. This was true in all methods of loading.

Torsion fractures are failures under tensile stress, not shearing as often stated.

The role of tensile stress in the mechanism of femoral fractures was expanded and further emphasized by Evans, Lissner and Pedersen (10).

A total of 50 femora, most of them stresscoated, from embalmed adult human cadavers were subjected to static and dynamic loading in different orientations of the bone. The controlled variables in each test were the magnitude, point of application, direction, type of force and bone orientation.

Vertical transverse fractures of the femoral neck were produced with static vertical loading of the femoral head without involvement of a torsional force.

Static and dynamic loading of the greater trochanter, with the femur in slightly different positions, resulted in subcapital, intertrochanteric, abduction and oblique fractures of the femoral neck. No torsional force was involved in the mechanism of these fractures. Spiral fractures of the femoral shaft were produced by static torsion loading and transverse frac-