Frequency response analysis of the in-vivo human skull
by Gerald Martin Grammens

A thesis submitted in partial fulfillment of the requirements for the degree of MASTER OF SCIENCE in Mechanical Engineering
Montana State University

Abstract:
In this study, a finite element model of the human skull is presented. The intent is to explore the feasibility of determining in-vivo material properties of the human skull by frequency response analysis.

Qualitative results are obtained by approximating material and geometrical parameters for the skull. Initially, boundary conditions were chosen to prevent rigid body translation by securing nodes about the mid-plane. The axisymmetric mode shapes obtained reveal resonance of the fundamental bending mode at 3700 Hz.

To account for this disparity of results compared with those obtained by previous investigators, an end-mounted support at a single node was imposed to replace the mid-mount boundary condition. This forced boundary condition was of the type employed in these previous studies. For the same skull with the end-mount boundary condition, the first resonance occurred at 440 Hz. The corresponding fundamental bending mode shape can best be described as a rigid body motion of the bulk of the skull with a local deformation at the fixed node. Recognition of the nature of this mode shape raises serious implications concerning past interpretations of such results.

The qualitative results of this study suggest that either of the two mode shapes may prove helpful in determination of skull material properties, but not without major experimental and analytical difficulties.

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by

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A thesis submitted in partial fulfillment
of the requirements for the degree

of

MASTER OF SCIENCE

in

Mechanical Engineering

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March, 1977

ACKNOWLEDGMENT

The author wishes to express his appreciation to Dr. Dennis Blackketter for the suggestion of this thesis topic and for his assistance throughout its evolution.

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ABSTRACT

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INTRODUCTION

A host of investigations have been carried out to determine the response of the human head to external excitation. Most of these investigations have as their objective advancing the understanding of the events which take place during the short period subsequent to impact of the skull. The impetus frequently cited for these studies is the statistically impressive rate of occurrence of death or injury due to impact sustained by the head. This is a result, primarily, from automotive and related accidents.

Such studies assume an element of uncertainty when a particular set of rheological parameters is assigned to the analytical models of in-vivo (live) human bone and associated tissues. The non-availability of this in-vivo information constitutes an obstacle to obtaining meaningful analytical results. In previous studies this impasse has been readily circumvented by taking as in-vivo those properties determined from, or extrapolated from, in-vitro (or ex-vivo) studies. The importance of these parameters in quantitative prediction of in-vivo behavior of the human head, along with the emergence of a wide range of values in recent literature, combine to substantiate the need for further refinement of in-vivo rheological properties.

A method to determine in-vivo material properties of human bone has been advocated by Garner and Blackketter [13]. This calls for an experimental procedure to measure the response of the system to an externally applied harmonic force over a range of frequencies. Similarly, a frequency response analysis is performed by utilizing an analytical model of the physiological structure under investigation. The input material parameters for the analytical model are altered so that the experimental and analytical responses most closely correspond. Those optimized material parameters are then said to be the in-vivo material properties.

The long bones, such as the tibia and ulna, are typically chosen for such studies because they can be characterized as having relatively simple geometry. However, experimental results and analytical modeling become complicated with the presence of attached and surrounding soft tissues. Also, slight changes in orientation and support conditions can significantly effect experimental response characteristics, which in turn adversely effects reproducibility.

Because the properties differ significantly between the long bones and those of the skull, it is imperative that similar determinations be attempted for the skull itself. In addition to the desirability of determining the in-vivo properties of the skull, such a study will provide further indication of the usefulness of frequency response analyses for these purposes.

One can surmise that the skull is less subject to the difficulties encountered in obtaining reproducible experimental results. Damping due to the brain and its surrounding tissue can be considered constant over periods of time such as might elapse between one experimental sampling and the next. In addition, one might expect little variation in brain properties from one individual to another. Positioning of the skull and accelerometer transducers can be easily reproduced, and, skull support conditions have minimal effects for properly chosen modes. Inherent with the skull is a greater complexity of geometry, which tends to detract from its desirability as the subject of dynamic analysis.

Because cranial bone properties cannot be equated to those of bone serving other functions [31], a method is needed to determine in-vivo bone properties specifically for carnial bone. The question at hand, then, addresses the feasibility of using the human skull to determine the material properties (or changes in material properties) of in-vivo bone. Success in this venture is contingent upon the ability of both the experimental and analytical investigations to accurately detect and predict at least one bending mode of deformation excited by a specific driving force. It is also essential to convincingly demonstrate that indeed the <u>same</u> modes are being described by both segments of the procedure. The emphasis in this study is placed on an analytical determination of the frequency response of the human skull, and, an attempt to determine which mode is likely to aid in the establishment of in-vivo skull (bone) properties.

Chapter 1

LITERATURE REVIEW

The bulk of work done in determining the mechanical properties of bone has been carried out in the last ten years. For earlier work the emphasis had been placed on determining parameters which describe bone as an elastic material. As early as 1965 Sedlin [27] published the results of experimental work in which he demonstrated that, for small deformations, bone behaves as a linear viscoelastic material. Furthermore, Sedlin's experimental results were best modeled by the 3-parameter solid model, also referred to as the standard linear solid model. Sedlin's work was conducted using in-vitro compact femoral bone.

Curry [5] conducted experimental investigations on the microscopic level dealing with the effect of mineral content of bone and its relationship to measured engineering properties. Curry studied the effect of mineralization or calcification on the elastic parameter--Young's Modulus. As will be seen in the following chapter on bone structure, one might have expected that Curry's work would verify the high causal-effect relationship between mineral content and Modulus of Elasticity.

Because of requirements for non-destructive testing, most information on bone, particularly human, is the direct result of in-vitro experimentation. Values used in studies for which in-vivo bone properties are necessary input parameters are usually taken directly from or, are extrapolated from, available in-vitro data. The validity of using in-vitro properties to describe in-vivo behavior is dependent upon the changes incurred by bone following death.

Past studies apparently are not in concurrence as to the effect of post-mortem age on bone properties. The following studies demonstrate this point.

McElhaney [22] indicates that the mechanical properties of embalmed bone do not differ "significantly" from immediate post-mortem properties. McElhaney's properties refer to the elastic properties--Young's Modulus and Poisson's Ratio.

Evans [8] modeled bone as a viscoelastic material and states that the elastic parameter is not effected by embalming.

Tennyson [28] investigated the stress-strain characteristics of beef femur bone as a function of post-mortem age (PMA). Tennyson initially modeled bone using the 3-parameter linear viscoelastic model. His results indicated that one of the stiffness parameters was small enough to allow him to neglect its effect. This reduced the model to the 2-parameter or Voight model. Further results based on the Voight model produced little change in the stiffness parameter with increased PMA, however, the viscous parameter showed radical change during early PMA. Here, early PMA refers to the first 24 hours of post-mortem time. It is necessary to extrapolate Tennyson's data to speculate on immediate post-mortem effect on bone properties. With decreasing PMA, approaching zero, the viscous parameter appears to increase exponentially for at least as far as Tennyson's data is shown (beginning approximately at .5 days PMA). Tennyson obtained his experimental data from the split-Hopkinson-bar technique.

Garner [13] modeled the human forearm using the Sedlin or 3-parameter linear viscoelastic solid. Comparison to in-vivo experimental data indicates large differences in in-vivo and in-vitro rheological bone properties.

At this point the literature seems to indicate that bone's transition from the in-vivo state to an in-vitro state involves small changes in the elastic characteristics and marked changes in the viscous or damping characteristics. The inconclusiveness of previous work establishes the need for meaningful in-vivo results. Furthermore, in the literature reviewed here, values taken for Young's modulus for the skull range from 6.0×10^5 psi to 2.29×10^6 psi, while the majority of investigators use an intermediate value of 2.0×10^6 psi. The following discussion can account for a portion of the variation in this one parameter.

Bone properties vary from species to species, from one individual to another, and from one functional physiological location to another within the same specimen. These facts are borne out by the following studies.

Wood [31] tested in-vitro cranial bone samples from 30 subjects ranging in age from 25 to 95 years, and performed tensile testing vs. strain rate. For cranial bone, no regional variation in the properties measured was noted, and no variation of properties with respect to tangential direction was found. Wood also states that cranial bones have properties intermediate of those for the longitudinal and transverse properties of long bones.

Hubbard [18] used beam testing techniques to determine the flexural stiffness and strength of layered cranial bone. Test specimens were obtained from embalmed calvaria (skull less jaw and facial bones). From these, the elastic properties were determined using a three-point flexure test. An equivalent modulus of elasticity was determined for the actual cross-section. Hubbard noted no significant effect resulting from the test sample extraction site, nor due to orientation, normal or inverted (i.e. convex or concave) of the test specimen in the test apparatus.

McElhaney [21] set out to determine the mechanical properties of cranial bone which are relevant to the study of biomechanics of head injury. Bone specimens were obtained from embalmed cadavers, craniotomies, and autopsy. In addition to the human cranial bone tested, specimens were also obtained from monkeys and tested. It is interesting to note that there were significant differences in resulting elastic parameters between the two species. Testing was done on the Tinius Olsen Electromatic Testing Machine with the bone samples maintained in "wet" condition. McElhaney noted that variations in the thickness of the porous middle layer of bone (diploë) resulted in a high standard deviation for material properties such as energy absorption and gross stiffness. It was also shown that the tensile properties for the inner and outer layers (tables) of compact cortical bone display no significant differences. Again, McElhaney is modeling cranial bone as an elastic material.

Research inspired by a high occurance rate of head injuries largely due to automotive accidents has been conducted in which various approaches have been undertaken to attempt to increase understanding of the biomechanics of the human skull.

Gordon et. al. [14] used a 2-dimensional model of the human head to determine the response to impact. The results were in terms of a pressure distribution through the brain. The work employed a somewhat more sophisticated model of the head than that used in previous analyses, in that a finite wave speed was used for the brain and skull.

Engin [6] determined the response for a thin elastic spherical shell filled with an inviscid compressible fluid. In a similar analysis Kenner and Goldsmith [20] worked with wave propagation through a liquid filled sphere for impacts of greater duration than those which can be represented by Dirac's delta function.

The above three analyses represent an approach to modeling the human skull and brain which involves the use of closed form solutions. These methods are useful in demonstrating qualitatively phenomena such as wave propagation, however, the closed form solutions require simplifying assumptions such as a spherical geometry for the skull, constant thickness of the cranium, and an inviscid irrotational fluid for the brain. An alternate approach incorporates the use of finite element techniques, which allows consideration of irregular geometries and variation in skull thickness.

Hardy and Marcal [16] hypothesized in their investigation of brain damage (particularly that resulting from automotive accidents) that, even though it is the brain itself which limits the impact absorption capacity of the human head, this limit may be best measured in terms of skull response. This prompted Hardy and Marcal to model the skull using doubly curved triangular shell finite elements. The skull was modeled as an elastic material, and, only static deflections were studied.

Nickell and Marcal [24] carried the preceding analysis one step further. The same finite element model was used, and the essential aspect for any impact study--the dynamic response of the skull--was considered. Three different types of support boundary conditions (frontal, rear, and base mounted) were used to determine natural frequencies and associated mode shapes which (it was hoped) might be relatively independent of the arbitrary support conditions. The lower mode shapes are the result of the particular imposed boundary conditions. In general, these can be described as rotation (mostly rigid body) about the prescribed support. The third mode in each case appears to exhibit relative motion of points at the support and those diametrically opposite, and occurs in the frequency range of 400-700 Hz. Nickell and Marcal hypothesize that this mode shape is a result of deformations due to the variations in thickness and curvatures of the particular skull modeled.

The above study used a modulus of elasticity approximately one third of that found in literature of similar investigations. This would cause the observance of lower frequencies for given mode shapes. No reference is given to support the value taken for this parameter.

Hickling [17], as was the case with most other investigators, had as his goal the determination of the mechanics of brain and skull damage during and after impact, particularly that associated with automotive accidents. He modeled the skull and brain as linear viscoelastic and isotropic materials and used a spherical geometry for the head. For his study and indeed applicable to other such studies, Hickling laments,

> "Unfortunately, the ability of the present model to predict tolerances is greatly hampered because adequate data on the material properties of the skull and brain are lacking and because appropriate damage criteria are not known".

Having examined some of the models of the skull and noting inherent difficulties and inadequacies, our attention is now directed to some of the experimental work which has been performed on the skull.

Early experimental work on the dynamic response of the skull was performed by Franke [10] and updated by Gurdjian et. al. [15].

Franke frontally loaded a dry preparation (skull) with a small support at the occiput (rear of skull) and found resonance at 820 Hz.

The same loading, when applied to a live human skull, produced no measurable response. This failure was described as the result of insufficient coupling between the force piston and the skull due to the layer of skin. Franke then performed his experiment on cadavers with the skin removed at the forcing location. The first flexural mode was found at 610 Hz. Without substantiation, this mode was taken as the dampened version of that found in the dry preparation at 820 Hz.

Gurdjian frontally mounted cadaver calvaria to a source of a harmonic force and measured the response at the left side, vertex, and occiput of the skull. The calvaria were tested both empty and filled with silicon gel. Resonant frequencies were detected at approximately 300 and 900 Hz. The first, an anti-resonant mode, resulted in an occiput acceleration amplitude three to four times greater than that of the frontal location where the harmonic load was applied. The second resonance exhibited relatively large frontal excursion as opposed to that of the occiput. With response monitored at only three points, little information is presented to give indication of actual mode shape.

In addition to the cadaver studies, human volunteers were used in a similar experimental set up, and response to a frontal force was measured. Gurdjian states that for frequencies below 150 Hz. only local deflections at the point of application of the force are detectible.

Prior and subsequent work has been done to determine resonance of biological structures. Those most applicable to this study are work done by Jurist [19] and Garner and Blackketter [13]. Jurist's work is mostly clinical in nature, and concentrates on the long bones of the outer extremities (ulna, radius, tibia, and fibula). Jurist modeled these as hollow cylinders of constant cross section and proceeded to determine the first bending frequencies.

Garner's work, which is the antecedent to this study, represents a more sophisticated analytical-experimental approach to correlate low frequency response to parameter determination. Garner used a finite element model of the forearm in which the bones (radius and ulna) along with the soft tissues were modeled as viscoelastic materials. Garner varied the rheological parameters in his analytical model until the response most closely coincided with that obtained from his experimental work, and thus defined the rheological properties.

Chapter 2

THE HUMAN SKULL

In order to more fully justify the use of simplifying assumptions necessitated by our analytical model of the skull, the present chapter is included. We will first examine the material of which the skull, the subject of our investigation, is composed--bone. Our next concern will be with the biological structure, the skull itself.

Bone

Bone is a highly specialized form of connective tissue. On a microscopic level, bone is composed of a soft branching matrix of cells arround which are deposited mineral salts. These salts are composed primarily of calcium and phosphate, which results in bone's characteristic hardness. It is this hardness which most differentiates bone from other connective tissues such as cartilage and ligaments. In addition to the inorganic bone salts and water, the interstitial substance includes a fibrous structure comprised of collagen. These collagenous fibers tend to orient themselves in a longitudinal fashion in long bones so as to maximize strength in directions of greatest tensile stress.

Most bones (particularly the long bones) are modeled during early development as cartilage; and only in later development are they remodeled as cortical bone. It is this remodeling process which results in the presence of haversian systems (or osteons). Consider a hollow core (many times greater in length than diameter) called the haversian canal. Around this haversian canal, the bone matrix and interstitial materials are deposited in concentric layers or lamellae. Lamellae of a haversian system and the outer lamellae of adjoining haversian systems become connected along what are termed cement lines. Thus the thickness of say a long bone is composed of many haversian systems.

Macroscopic Properties

In macroscopic perspective there are two basic types of bone: 1) hard compact cortical bone such as that found in the shafts of long bones, and 2) spongy or cancellous bone which consists of many fine partitions called trabeculae, which form cavities containing red or fatty marrow. Cancellous bone is found in the vertebrae, majority of flat bones, and at the end of long bones. Thus cancellous bone is porous and is characterized by intricate architecture.

The inorganic mineral is the major component of bone. McLean and Urist [23] cite for Bovine bone the following weight percentages

Mineral	72%
Organic Matter	24%
Water	4%

When the fatty material is removed, 90-96% of the remaining organic matter of bone is collagen.

Bone Dynamics

Changes occur in bone by way of the normal physiological mechanisms of growth, aging, maintenance and repair, by pathological mechanisms such as osteomalacia, and by trauma such as impact with a foreign object culminating in fracture.

Three types of specialized cells account for the physiological dynamics of bone. The first of these is the osteoblast. Osteoblasts appear on the surface of growing or developing bone and provide the biological mechanism for the formation of new bone. Osteocytes are osteoblasts which have become surrounded by the matrix, which subsequently calcifies, and provide for the maintenance of surrounding bone. The last of the three types of cells is the osteoclast. Osteoclasts are usually found on the bone surface near areas of active remodeling and perform the physiological function of resorption of existing bone.

Changes in bone are apparent during growth, but even when the geometric parameters are constant, as in periods during adulthood, the process of remodeling (replacement of existing bone with new) is active. By way of example, McLean and Urist [23] state that for a typical 57 year old human, the bone mass turnover in one day is 0.036% for the femur and fibula and 0.012% for the tibia.

The physiological mechanism of aging and the pathological mechanisms can cause an upset in the balance of osteogenic activity, resulting in changes in bone's microscopic structure and therefore, the macroscopic properties. McLean and Urist discuss the condition of atrophy which refers to a loss of substance of bone without a corresponding loss in the external dimensions (or gross volume) of the bone. This may advance to a state where as much as 75% of the original bone mass has been lost.

Little is known of the effect of early post-mortem time on the macroscopic material properties of bone. Some researchers, such as McElhaney [21] indicate that there is little effect on bone properties due to post-mortem age. Others, such as Garner and Blackketter [13], cite indications of marked changes. Tennyson [28], using Split-Hopkinson-bar testing techniques, plots changes in an elastic and viscous parameter for bone with respect to post-mortem time. His results indicate little change in the elastic parameter but great effect on the viscous parameter immediately after death and through the first few days post-mortem.

The Skull

The human skull consists of the calvarium, the jaw (mandible), and the facial bones. The calvarium is an ellipsoidal shaped enclosure which serves to protect the brain. The one major opening in the calvarium, through which the medula oblongata (anterior portion of the brain which connects to the spinal cord) passes, is the foramen magnum located at the base of the skull. The calvarium is composed of eight different bones which are joined by cartilage and a fibrous tissue in early development. By adulthood these bones are joined in a much more rigid manner at the boundaries along connective formations referred to as sutures. Examination of cadaver skulls reveals great variations in the degree to which the cranial bones have fused at the sutures. The range is from interlocking, but still separate, bones to cases in which fusion is so complete that the suture lines are hardly discernible.

The bones which comprise the calvarium are sandwich structured through the thickness. The bones are composed of inner and outer tables of compact cortical bone separated by a porous inner layer of bone called the diploë. The thickness of the diploë layer varies within the constituent bones from as much as half of the total skull thickness to no thickness at the sutures. The thickness of the calvarium varies with location. Typical variations might be from 0.12 inches in the temporal region to 0.25 inches in the frontal region.

The regional nomenclature for a typical skull is shown in Figure 2-1.

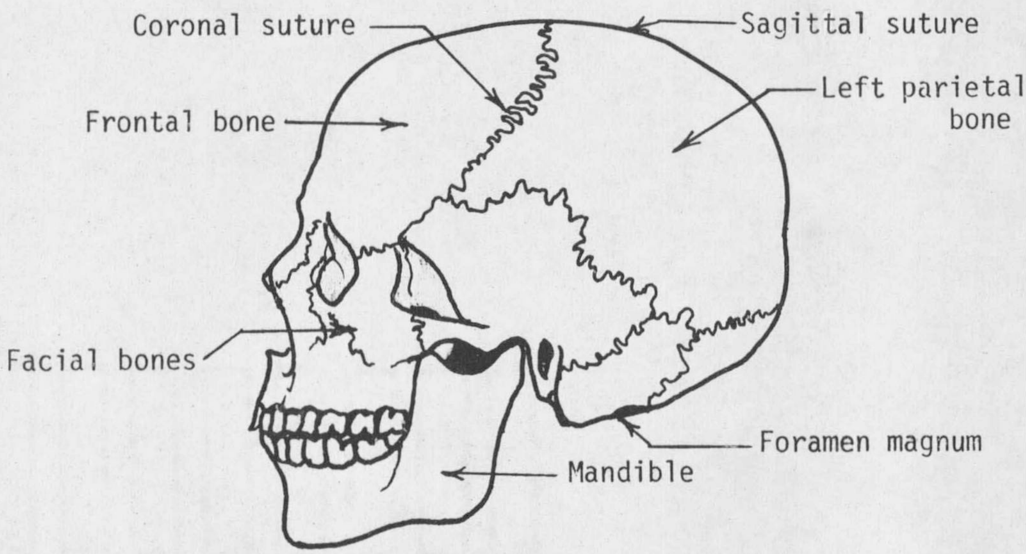

FIGURE 2-1. Typical Skull.

Having discussed briefly the physiological nature of bone and the skull itself, we have the basis on which evaluation of the validity of simplifying assumptions can be made. We must also consider the cases for which specific conditions are requisite to the acceptance of a particular simplifying assumption.

Earlier it was noted that, with the exception of the collagenous fibers, bone appears to be homogeneous on a macroscopic level. The effect of the collagenous fibers is minimal in the flat bones such as those which comprise the skull. Along these same lines, it was previously reported from an experimental study by Hubbard [18] that no significant variation in bone properties was measured in directions tangential to the skull surface. Thus, we have the basis for assuming bone to be homogeneous in in-plane directions.

Variations of thickness of the inner and outer tables, and the diploë layer, along with slightly altered properties in a normal (to the surface) direction suggest that inclusion of these characteristics is necessary for quantitative results. This is of even greater significance for bending deformation, where stiffness increases as the cube of the distance from the central fiber. A constant thickness and the use of gross properties to compensate for diploë thickness can only be expected to produce qualitative results.

The homogeneous shell model will suffice to predict bending response for a skull where sutures are sufficiently fused so as to provide continuity between component bones. It must be cautioned that in-vivo skulls with merely interlocking or cartilageneous sutures present a physical situation which is not likely to be adequately modeled quantitatively by the homogeneous shell.

The calvarium, being nearly a complete enclosure, can be modeled in simplest form as a spherical shell of constant thickness. Though the sphere is a first generation model of the skull, its response is indicative of the resonant frequencies and mode shapes of the skull. A more accurate geometrical representation is the ellipsoid. The typical calvarium, however, deviates somewhat from this geometry. In some specimens irregularities even result in a lack of symmetry about the sagittal plane (i.e. non-symmetry between the left and right sides of the calvarium). A truly quantitative model would take into account even these irregularities.

Other difficulties in modeling the skull which must be considered or for which proper assumptions must be made are: the effects of attached or surrounding facial bones, jaw bone (mandible), brain and fleshy materials, and the physiological support system.

Chapter 3

MODELING CONCEPTS

The analytical model of the skull presented in this study was developed so that, if necessary, consideration can be given to the viscoelastic nature of bone, irregularities in geometry, and regional variations in thickness and material properties. Initially, however, the skull is modeled as a thin wall homogeneous shell of constant thickness and of axisymmetric geometry. The intent here is to obtain qualitative insight into skull response characteristics.

The use of an axisymmetric geometry permits the generation of nodal coordinates and assumed modes to be completed with minimal computational effort. Of greater significance is the capacity to effect a coordinate reduction of the original 336 degree of freedom system to one described by 28 generalized coordinates, while maintaining the capability of identifying any axisymmetric mode.

Geometry Generation

The skull was modeled in this study as an ellipsoidal shell for ease in generating nodal locations and to fill the requirements of axial symmetry. The equation for an ellipsoid in cartesian coordinates is

$$1 = \frac{\overline{X}^2}{A^2} + \frac{\overline{Y}^2}{B^2} + \frac{\overline{Z}^2}{C^2} \tag{3-1}$$