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Photoelasticity applied to the study of human bones

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Abstract

The photoelastic technique is applied to obtain the distribution of stresses in some critical points of the human body, specifically in the femur and the mandibular bones. The study is done based on simplified hypothesis, using two and three-dimensional photoelastic models, and the results are correlated with real bones and also compared with some results from the literature.

1 Introduction

Two distinct ways exist in the development of the stress analysis knowledge. The first one is that supported by mathematical concepts in which the *Theory of Elasticity* was outlined and developed. This stage was important because the general principles developed provide the foundations for complete theoretical or experimental solutions of all stress analysis problems. The second way, known as *Experimental Stress Analysis*, provides the means by which conclusions reached by theoretical reasoning are checked and related to the physical world. Experimental Stress Analysis has been regarded for some time as a distinct activity within the field of mechanical and civil engineering, which objective is the determination and improvement of the mechanical strength of machines and structures. Today, the experimental techniques of stress analysis have a wide application, including studies in Biomechanics.

Experimental Stress Analysis refers to the process by which stress distribution in structural elements or machine components is determined by making measurements of the physical changes produced by the applied loads on either the actual structure or a suitable model. In particular, *Photoelasticity* is an experimental technique for stress and strain analysis that is very useful for members having complicated geometries, complicated load conditions, or both.

Photoelastic analysis is used for problems involving two and three-dimensional geometries in which stress or strain information is required for extended regions of the structure rather than point-by-point information. Recently, ready access to large digital computers together with the development of the Finite Element method of analysis have provided a completely new and complementary way of approaching stress analysis problems.

Several attempts have been made to apply experimental techniques of stress analysis in problems related with the area of Bioengineering. Measurements of shock absorption characteristics of athletic shoes were studied using strain gages [1, 2], Moiré fringes [3], piezoelectric transducer [4], in order to improve sole designs. Nusholtz [5], using electromechanical transducers, measured the impact response of the human thorax, providing basic information for understanding the kinematic response of the human body. In a theoretical analysis supported by experiments using pressure transducers, Sonnerup [6] studies the intervertebral disk in compression. Rubayi [7] investigated the strains in a small bone of the human wrist joint employing electrical resistance strain gages. Using photoelastic coatings, Andonian and Tudor [8] presented the stress distribution in lion and wolf skulls. A large number of fractures of the human bones and the necessity of prosthesis are motivating researchers to make studies in this direction. Paul [9], using experimental techniques found the load actions on the human femur when walking and some resultant stresses. The pressure distribution on the cartilage surfaces in the human hip joint was measured experimentally by Carlson [10], replacing the ball part of the joint by a specially instrumented prosthesis. In a similar study, Singerman et al. [11] measure the contact stress distribution using pressensor films. Related with the mandible bone, Smith [12] presented a numerical model of the mandible, its articulating surface and the forces exerted by the primary masticatory muscles. Osborn and Barager [13] measured the direction of the loads related with the occlusal plane. Other studies have been involved with the determination of the force actions and mechanical capabilities of the human jaw muscles and tempo-romandibular joint, as reported by Barbenel [14], Van Euden et al. [15], Throcknorton [16] and Sakaki et al. [17]. Using photoelastic stress analysis, Standlee et al. [18] reported an important study to visualize the intensity and direction of stress within the mandibular condyle generated by various occlusal forces. The static tensile strength and physical properties of compact bone have been reported by many authors [19, 20, 21]. On the other hand, Tennyson [22] discussed the experimental results obtained when dynamic viscoelastic responses of bones are obtained. Saha and Hayes [23] presented results for fresh beef bone samples tested against the tensile impact strength and static tensile.

The objective of this investigation was to apply an experimental technique to study the stress distribution in bones of the human body. Photoelasticity was used to obtain the distribution of the stress in some critical points of the femur and the mandibular bones. The study is done based on simplified hypothesis, using two and three-dimensional models, and the results are correlated with the real bones and also compared with some results from the literature.

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2 Photoelasticity

Photoelasticity is an optical technique for stress and strain analysis that makes use of special types of plastic models, a very powerful tool to study complicated structures. In such cases, analytical methods become unwieldy or impossible and analysis by experimental means often provides the only realistic solution. Photoelasticity is a full-field method which yields an overall picture of the stress distribution, quickly giving qualitative information about the stress distribution, and further on giving quantitative data about the stress after proper interpretation of the optical parameters [24].

2.1 Two-dimensional Photoelasticity

As stated previously, photoelasticity methods can be applied to any state of stress; however, they can be more easily applied in studies of two-dimensional problems. In general, the method requires the manufacturing of a model of the specimen out of some transparent, homogeneous, isotropic, linear, elastic material possessing the desired optical properties. The fundamental optical property required is that of temporary double refraction which certain transparent materials exhibit when subjected to stress or strain. When the study is conducted, loads which closely simulate those of the prototype being evaluated are applied to the model. The optical effects in the model resulting from these loads are then viewed in a field of polarized light which is provided by an instrument known as a polariscope. With a white light source, optical effects are manifested as colored bands covering the range of the visual spectrum. With a monochromatic light source, optical effects are manifested as a series alternate dark and light bands, referred to as isochromatic fringes, which are labeled according to the number of darkness-brightness cycles that occur at a point in the model as the load is increased from zero to its final value. The fringe orders are commonly related to the stress in the model by means of a relationship known as the *stress-optic law*.

In the case of two-dimensional stress fields, the isochromatic fringes are loci of points of constant maximum shear stress in the plane of the specimen. By the simple process of counting fringes and multiplying their number of order by a calibration constant, the maximum shear stress distribution, or principal stress difference, throughout the specimen can be determined. On free boundaries and at any other point where the stress field is uniaxial, the maximum shear stress is equal to one-half of the nonzero principal stress. Since critical stresses in a two-dimensional body frequently occur on a free boundary, the photoelasticity method provides an excellent means for determining stress distributions. At points in the interior of two-dimensional specimens, individual values for the two principal stresses cannot be obtained directly from the optical patterns without using supplementary data or employing numerical methods.

2.2 Three-dimensional Photoelasticity

Historically, photoelasticity methods were, for many years limited to two-dimensional stress problems. In the latest years, the stress-freezing technique was developed, what extended the use of the method to three-dimensional problems. The stress-freezing method of "locking in" model deformations due to an applied load, is based on the diphase behavior of many polymeric materials when they are heated. Such materials are composed of long-chain molecules into a three-dimensional network of primary bonds, but a larger number of the molecules are less solidly bonded together into shorter secondary chains. When the polymer is at ambient temperature, both sets of molecular bonds act to resist deformation due to applied loads. As the temperature of the polymer is increased, however, the secondary bonds break down and sole the primary bonds effectively support the applied loads. Since the secondary bonds constitute large part of the polymer, the deformations which the primary bonds suffer are quite large but elastic. If the temperature of the polymer is lowered to ambient temperature while the load is maintained on the model, secondary bonds will be formed again among the highly elongated primary bonds, and lock them in their extended positions. When the load is removed, the primary bonds relax to a moderate degree, but the main portion of their deformation is not recovered. Since the deformations are "locked-in" in a molecular scale, the deformation and accompanying optical response are maintained in any small section cut from the three-dimensional model. In practice, slices and subslices are cut from the model in regions of interest, viewed and photographed in a polariscope, and analyzed in a manner similar to that used for two-dimensional models.

2.3 Complementary Background

Two and three-dimensional photoelasticity studies require manufacturing models of the prototype and, for that reason, a relation for the transition model/prototype must be used. In general, the stress obtained from the similar photoelastic model (m) can be related to the prototype (p) through a simple mathematical equation given by the theory of models [25],

$$\sigma_p = \sigma_m \frac{P_p}{P_m} \cdot \left(\frac{l_p}{l_m}\right)^2 \cdot \frac{I_m}{I_p} \tag{1}$$

where "σ" is the stress in a particular point, "P" is the applied load, "l" is the main dimension and "I" is a property of area (moment of inertia).

On the other hand photoelasticity, like all experimental techniques, needs special types of equipment and materials for its application [24, 26, 27, 28, 29].

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3 Application and Results

The theory presented was used to study the stress distribution in two different parts of the human body, the femur and the mandibular bones. In those cases, the necessity of corrective prosthesis or tooth implantation still constitute an exciting subject of research to better understanding the behavior of such systems. An analysis of the mechanical factors pertinent to this in order to reach the real stress distribution is, however, particularly complicated in view of the complex dynamic force system to which they are subjected, the complex anatomical configuration of the members and the variable, anisotropic, rate-dependent mechanical properties of the material of the bone. The study based itself on simplified hypothesis, in which isotropic materials are used and a concentrated load is applied.

3.1 Plane Simulation

Plane models using photoelasticity material were made with the geometry of the meridional plane of the femur and with the shape of the lateral projection of the mandible. Figures 1 and 2 show one example of the samples studied, using a special load system. From this simple plane analysis, qualitative information from the models are found to be compatible with those observed on the real bones. Figure 1a shows the internal structure of one dry real bone of femur, having area with great porosity in contrast with a no-porosity area with high strength bone material on the boundary, down of point A. The photoelastic model with the typical fringe distribution, shown on Figure 1b, reveals similar features, where the fringe order number increases down of point B, similar position of point A. Also, low fringe order can be observed in the area corresponding to the high porosity region in the bone. It should be pointed out that the position of zero fringes (zero stress) on the photoelastic model are located on the region of the main arteries, point C on Figure 1b. Figure 1c shows the tension stress distribution in the area covering the position of the neck of the femur, where normally fractures occur.

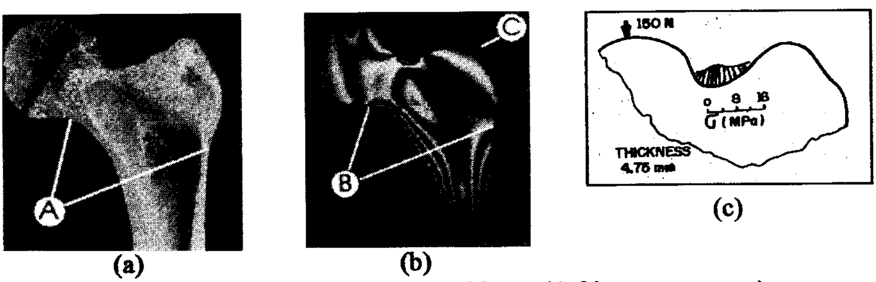
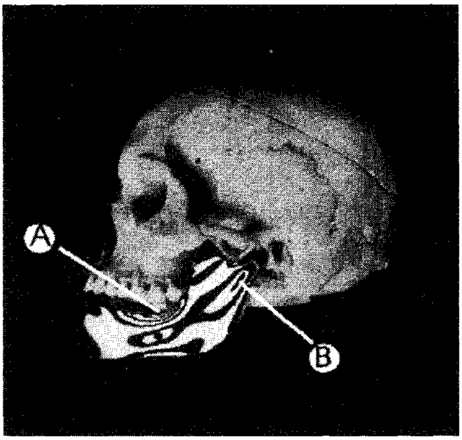
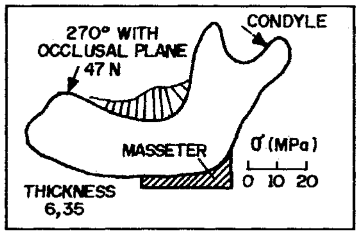


Figure 1. Femur bone: a) section of the real bone; b) fringe patterns; c) stress distribution on the boundary of the neck.

The analysis on the mandible plane photoelastic models also gives significant information and is compatible with the bone structure. The highest level of fringes (stresses) observed in the position of the molar tooth, point A shown on Figure 2a, coincide with the position of the mandible which has the greatest thickness. Point B on the photoelastic model (Fig. 2a), which represents zero stresses (or isotropic point), is located almost in the same position of the mandibular foramen, a hole in the mandible through which nerves and the inferior alveolar artery pass. Those nerves and arteries follow a channel inside the mandible which has the same position in the photoelastic model where low stress (firnge) values are found. In order to evaluate the maximum stresses on the shape of mandible studied, the tension stress distribution was calculated and is shown in Figure 2b.



(a)



(b)

Figure 2. Mandible bone: a) fringe patterns; b) stress distribution.

In order to make the transition model/prototype, a shape for the transverse section of the bones was assumed, so that it was possible to estimate the maximum stresses acting on the real bones, under the hypothesis made. The calculation of the maximum stress in the neck of the femur was made, using an elliptical section where a = 28,5 mm, b = 20,5 mm and the cortical thickness e = 3,5 mm. With the data obtained experimentally as shown in Figure 1c, assuming the vertical loading found by Paul [9] as $P_p = 2720$ N and making use of Equation (1), the maximum tensile stress was obtained as $\sigma_p = 62$ MPa. Similar calculation was done for the mandibular bone with the load applied on the position of the incisive tooth and directed over the range of 270° to 300° with respect to the occlusal plane [13]. The transition was made using half of the the bite of $P_p = 196$ N suggested by Trocknorton [16], using transversal elliptical section with a = 26 mm, b = 16 mm and the critical thickness assumed to have an average value of e = 2,2 mm. The greatest value for tension stress was $\sigma_p = 23{,}5$ MPa.

3.2 Three-Dimensional Simulation

A more sophisticate three-dimensional model using epoxy resin [26] was made, copy of the real femur bone. Stresses were frozen for a vertical applied load on the head of the femur and slice and subslices of the meridional plane were cut from the model, as shown on Figure 3. The complete state of stress (τ_{ij}) was determined on the boundary of the femoral neck. Figure 4 shows some results on the plane of the meridional slice. From the results, the principal stresses can be obtained and the transition model/prototype made in a similar way as in the plane model. Stresses calculated on a simplified basis are found to have maximum of 54 MPa and 157 MPa in tension and compression, respectively.

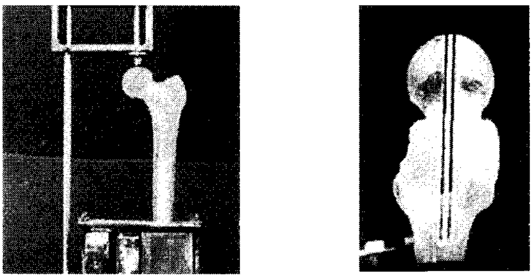
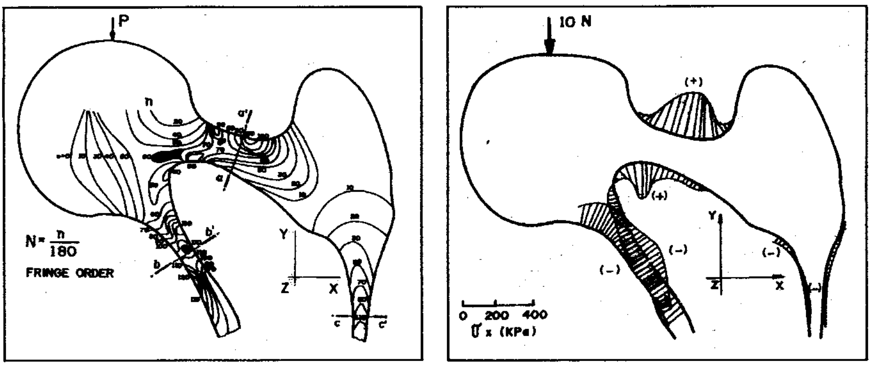


Figure 3. Three-dimensional model of the femur, loaded and sliced.



(a) (b)

Figure 4. Meridional Slice: a) fringe order and position of transversal slices; b) σ_x stress distribution.

As in the case of the femur simulation, a real mandible bone was used to build the three-dimensional epoxy resin model, shown on Figure 5a. The load of $P_m = 10{,}1$ N was applied in a similar manner used for the plane model (Fig. 2b) and the stresses frozen. Due to the symmetry of the mandible, meridional and transversal slices were cut from the model (Fig. 5b) and the complete state of stresses (τ_{ij}) obtained on the boundary, in the position of the molar tooth. Figure 6 shows the stress distribution on the critical area, determined from the model. Making the transition model/prototype with the principal stresses, as performed before, the maximum tension stress on the prototype was found to be $\sigma_p = 22$ Mpa.

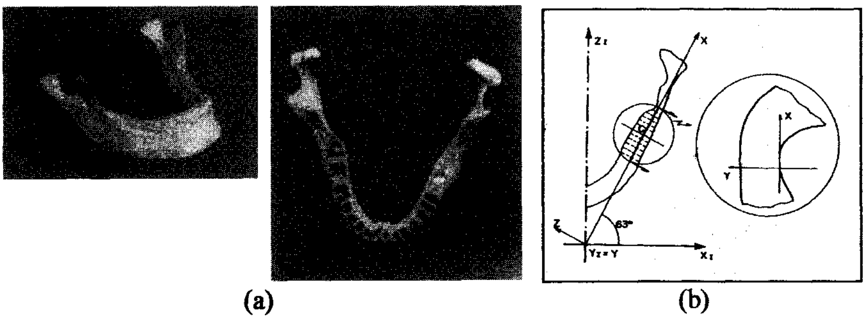


Figure 5. Mandible bone: a) three-dimensional model; b) reference and slices.

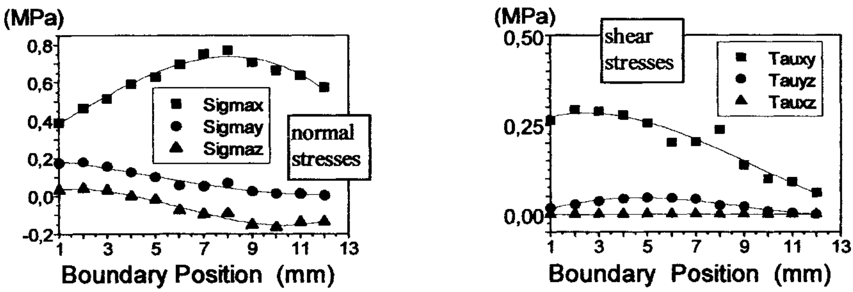


Figure 6. Mandible: State of stress on the critical area.

4 Comparison and Conclusion

The stress values calculated in this study using Photoelasticity under the simplified basis, including no influence from muscles, was a tensile stress of 23,5 MPa for the plane simulation of the mandible and 22 MPa for the three-dimensional model. For the bone of the femur, the plane simulation gave a tensile stress of 62 Mpa and the three-dimensional model the maximum tensile stress of 54 MPa

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(section aa' on Fig. 4a) and 157 MPa in compression (section bb' on Fig. 4a), both in the area of the neck. The literature reviewed gave no indication of possible failure criteria for bone under combined stressing. Evans [20] quotes ultimate stresses in tension of 110 Mpa and Saha and Hayes [23] found 121,3 ± 36,5 MPa in tensile-impact test. This would indicate an average safety factor of 5 for the mandible and only 2 for the femur. Other comparison refers to the calculation of the stress in the neck of the femur made by Paul [9], in which the forces of the muscles and ligaments was considered. Paul found the maximum of 19 MPa and 29 MPa in tension and compression, respectively. From these considerations, it can be concluded that the muscle activities on the femur are much greater than on the mandible in terms of force absorption. Muscles of the mandible are to being into act the mandible bone. Another important feature is that the results of plane stress analysis were considerably close to the three-dimensional approach for both, mandible and femur bones, although the three-dimensional analysis gave small values as expected. Nevertheless, the plane study does not reveals significant conclusion as observed from the three-dimensional results of the mandible (Fig. 6), where the sign of σ_z changes at the position of maximum τ_{yz} (area of the molar tooth), which characterizes a torsion on the mandible.

This investigation demonstrates the utility of photoelastic studies for determining stress distribution in bones of the human body. When cross-fertilization occurs between interdisciplinary areas, it often leads to a breakthrough which in turn spawns an array of ideas [30]. In Biomechanics, as elsewhere, the biggest issue is to establish a dialogue between the engineer and the individual with a need. These examples scratch the surface of a potentially vast and fruitful area of research. Engineers do, indeed, have a contribution to make in Life Sciences, beyond building better tractors, computers, aircraft, and the like.

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6 Key Words

Biomechanics; Photoelasticity; Stresses in Bones.

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