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THE ELASTIC MODULUS OF FETAL CRANIAL BONE: A FIRST STEP TOWARDS AN UNDERSTANDING OF THE BIOMECHANICS OF FETAL HEAD MOLDING*

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Abstract – Fetal head molding is the change in shape of the fetal head due to forces of labor. The biomechanics of this process are poorly understood. To understand it better, classical engineering structural analysis is being applied to analyze the process. A fundamental part of this analysis is to describe the mechanical properties of the constituent materials, a knowledge which has been lacking for fetal cranial bone.

As a first step toward defining the mechanical properties, 86 specimens of fetal cranial bone obtained from specimens ranging in estimated gestational age from 25 to 40 weeks were tested in three-point bending. In addition, 12 specimens from a 6-yr-old calvarium were tested for comparative purposes. The data indicate that the elastic modulus is highly sensitive to gestational age and fiber orientation. Elastic modulus values for specimens with parallel fiber orientation are in the range of 1.65×10^3 MPa for preterm bone to 3.86×10^3 MPa for term bone. The results are discussed with respect to the limitations of the test method. Finally, clinical inferences are made.

INTRODUCTION

Molding is the change in shape of the fetal head due to the forces of labor. The process is a normal part of every labor and as such it allows the fetal head to more easily fit the shape of the maternal birth passage. However, excessive molding with subsequent cerebral trauma has been linked to conditions ranging from subtle psycho-neurological disabilities to mental retardation, cerebral palsy and even death (Willerman, 1970; Glenting, 1970; Fianu, 1976).

Despite the undisputed importance of head molding in both normal and abnormal labor, little research has been done to understand better the biomechanics of the process. Those studies which have been undertaken have been widely spaced in time, using a number of very different methods (Murray, 1888; Stumpf, 1907; Holland, 1922; Moloy, 1942; Baxter, 1946; Borrell and Fernstrom, 1958). The results have been largely qualitative in nature and allow for little quantitative understanding of the molding process.

As a first step towards the quantitative description of fetal head molding biomechanics, classical techniques of structural analysis are used. The fetal head may be viewed as a very complex mechanical structure. Over 50 years ago this complexity was described: "The fetal head consists of a non-rigid-shell, of a peculiar shape, composed of loosely-jointed plates of pliable bone jointed on a rigid base and strengthened by the attachment of the dura mater and its system of septa: the contents of this shell are partly solid and partly fluid." (Holland, 1922).

As in all structural analyses, the mechanical response of the fetal head can be derived from a knowledge of three main areas:

(1) the mechanical properties of the tissues;
(2) the structural configuration, particularly the structural configuration of the cranial bones, dura mater and brain; and
(3) the loads applied during labor and delivery.

Preliminary information is currently available for both the structural configuration of the various head components (Scammon and Calkins, 1929) and the fetal head loads during labor (Lindgren, 1960; Schwartz *et al.*, 1970). To date, however, no investigations of the mechanical properties of fetal cranial tissue have been described.

This study will examine the elastic modulus of fetal cranial bone as determined from three-point bending tests. In addition, the possible influences of gestational age, specimen location and bone fiber orientation on the elastic modulus will be addressed.

DESCRIPTION OF MATERIAL

The fetal cranial bone used in this study was obtained from 6 subjects ranging in estimated ges-

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Table 1. Biographical data for test material used in this investigation

Calvarium no.	Estimated gestational age (weeks)	Sex	Weight (g)	Bone thickness (mm) Mean	S.D.	Cause of death
1	40 ± 2	M	3118	0.76	0.10	Congenital heart failure
2	38 ± 2	F	2800	0.71	0.15	—
3	27 ± 2	M	970	0.63	0.10	Insufficient respiratory effort
4	25 ± 2	M	650	0.41	0.10	Prematurity
5			Not tested due to bone deterioration			
6	28 ± 2	M	1025	0.64	0.05	Systemic candidiasis, bronchopulmonary dysplasia
7	40 ± 2	M	3640	0.86	0.08	Meconium aspiration
8	6-yr-old	M	16600	3.33	0.28	Acute bronchopneumonia (hydrocephalic)

tational age from 25 to 40 weeks. In addition, the calvarium of a 6-yr-old child was tested for comparative purposes. No material was utilized in this study which showed evidence of deterioration due to pathological conditions. Table 1 provides a summary of the biographical data for the test material.

At the time of post-mortem examination (usually within 12–24 hr after death), the cranial bones of the vault with attached septa and dura mater were removed in one piece and placed in a container of cold buffered saline. The material was maintained under refrigeration at between − 10 and − 20°C until the time of specimen preparation and test.

Specimen size and shape

Fetal cranial bone presents a unique set of constraints which influence specimen size and shape. The bones of the fetal cranium are highly curved and very thin structures. Together, these constraints prohibit the fabrication of flat, straight specimens for use in bending tests. In addition, the small size of the fetal cranium severely limits the maximum overall dimensions of any specimen if a reasonable number are to be obtained from each cranial bone.

The theory of flexure is most accurate for long-thin beams, i.e. beams with large length–thickness ratios. Beams having length–thickness ratios from 8 to 24 or more (depending on material and cross-section) are capable of being accurately analyzed using Euler–Bernoulli beam theory (Roark, 1965). Using these values as a guide, and assuming a bone thickness of 1 mm, a span length of 20 mm was selected. In order to allow a suitable amount of material to project beyond the end support, the overall length of the specimen was increased to 25 mm. A beam width of 2 mm was arbitrarily chosen to allow a large number of specimens to be obtained from a bone.

The specimens had a variable thickness due to the varying bone thickness, and they were curved in the plane of the long axis of the specimen due to cranial bone curvature. Because it was not possible to fabricate specimens without these conditions, those variables had to be taken into account in the subsequent data analysis.

Specimen preparation

An anatomical feature which must be taken into account in specimen preparation is the grain structure of the fetal calvarium. Split-line analysis studies have demonstrated that the fetal skull has an ordered grain structure, particularly in the regions of the vault (Burdi, 1978). At adulthood, this well-defined pattern is replaced by a random pattern; the grain structure of the adult is more homogenous than that of the fetus (Dempster, 1967). Specimens were therefore taken with the long axis of the specimens oriented in preferred directions with respect to the ordered grain structure of the bone.

At the time of specimen preparation, the refrigerated cranial vault was allowed to warm to room temperature in its saline-filled storage container. The membranous sutures holding the bones of the vault together were excised. The inner and outer membranes covering the bone were carefully removed. Each bone was placed in a separate saline-filled container. At all times, the bone was kept wet by periodic bathing in a normal saline solution.

A 25 mm square mylar template was used in the initial step of bending specimen fabrication. The outline of the square was transferred to the bone using a No. 2 pencil. Using the outline, a square section of material was removed from the cranial bone using a razor knife and scissors. The section of bone was then positioned in a miter box with the bone fiber oriented so that the finished specimens would have fiber orientation either parallel or perpendicular to the long axis of the specimen, as desired. Successive parallel cuts were made with a razor until all possible specimens had been taken from the section. After each specimen was fabricated, it was placed in a labelled saline-filled container.

Specimen geometry measurement

Because the specimen thickness and initial curvature could not be controlled during specimen fabrication, the degree of influence which these variables had on specimen response was evaluated. Classical flexure theory indicates that the deflection of a beam is not influenced by a small initial curvature. The word "small", however, is only vaguely defined and hence a careful evaluation of the effect of initial curvature for the particular case of the fetal cranial bone specimens was necessary.

Examination of the bending specimens revealed a maximum initial midspan displacement of approximately 2.5 mm with respect to the specimen ends. In order to ascertain the effect of this initial curvature, a beam with a half-sinusoidal initial curvature was modeled using finite element analysis. This beam was assumed to be made of isotropic material possessing a rectangular cross-section. The beam was discretized into 8 beam elements. The initial curvature at midspan was assumed to be one-tenth the span length, a curvature which corresponded to the greatest initial curvature seen in the bone specimens. Beam dimensions were those described above with an assumed depth of 0.65 mm. Comparison of the finite element beam response with that predicted by simple beam theory for an initially straight beam in three-point bending indicated that the initial curvature induced an error in the calculated deflection of less than 1%. Hence, it was concluded the effect of initial curvature was small and could be ignored in the data reduction.

A second finite element analysis was performed to evaluate the effects of varying specimen thickness on beam response. The thickness of a beam influences its response through the area moment of inertia

$$I = bh^3/12, \tag{1}$$

where

I = area moment of inertia (mm^4),
b = beam width (mm),
h = beam thickness (mm).

The width and depth of a randomly selected specimen were measured with dial calipers at 5 equally-spaced locations along the specimen. With these measurements, an average area moment of inertia for the 8 finite elements of the specimen were computed. The inertia data from all subsections were also averaged to give an overall mean value to be used in the equation for the midspan deflection of an uniform beam in three-point bending

$$y = Pl^3/48EI, \tag{2}$$

where

y = midspan deflection (mm),
P = midspan load (N),
l = span length (mm),
E = elastic modulus (MPa),
I = area moment of inertia (mm^4).

Comparison of the results of the finite element model with simple beam theory indicated that averaging sectional data induced an error in the calculated deflection of 18%. Hence, the varying thickness of the specimen could not be ignored. As a result, the width and thickness of each specimen was measured with dial calipers to the nearest 0.01 mm at 5 equally-spaced locations along each specimen. The measurement locations were at the end support points, the midpoint and the points equidistant between the supports and midpoint. These data were recorded in tabular form for use in the data reduction.

PRELIMINARY INVESTIGATIONS

Previous investigators of fetal skull molding indicate that molding is primarily a bending phenomenon (as opposed to a tensile or torsional phenomenon) (Holland, 1922; Borell and Fernstrom, 1958). In addition, the process is slow, as uterine contractions during labor generally average about 1 min in duration. Therefore, an appropriate test of fetal cranial bone is a quasistatic bending test.

Prior to the main test program, however, preliminary investigations were performed in order to determine the effects of variations in several parameters on material response. Two preliminary investigations were performed. These were:

(1) the effect of multiple load–unload cycles on specimen response;
(2) the effect of testing with the specimen immersed in warm saline vs testing in room air.

(a) *Multiple-cycle loading*

Fung (1972), as well as others, has noted a differential response of biological tissue with multiple-cycle loading. In general, tissue needs to be exercised, i.e. preconditioned, through several loading cycles before repeatable material response is obtained.

To test this effect, 4 specimens from the right parietal bone of calvarium No. 2 were tested in three-point bending at a crosshead rate of 0.5 mm/min. Each specimen was cycled through 10 complete load–unload cycles. Two specimens were tested allowing 2 min between each load while the other two specimens were cycled continuously through all 10 load–unload repetitions. Each specimen was kept moist throughout the test procedure by a constant drip of normal saline.

In each case, repeatability was established after the completion of the third cycle. Therefore, all specimens subsequently tested were cycled three times to precondition the material prior to measuring material response.

(b) *Warm saline vs room air*

After preconditioning, 8 bending specimens from the right parietal bone of calvarium No. 2 were loaded in three-point bending to a midspan deflection of 0.25 mm and unloaded. The specimens were divided into 2 equal groups. In the first group, the specimen was initially tested in normal saline maintained at 37 ± 2°C. After allowing 15 min for equilibration with ambient air, the specimen was reloaded to a midspan deflection of 0.25 mm and unloaded. In the second

Table 2. Beam stiffness comparison for specimens tested in warm saline vs room air

Specimen	K_s (N/mm)	K_a (N/mm)
2PR-PD1	1.23 (2)	1.23 (1)
2PR-PD2	1.58 (2)	1.40 (1)
2PR-PD3	1.93 (2)	1.75 (1)
2PR-PD4	2.80 (2)	2.98 (1)
2PR-PD5	2.36 (1)	3.15 (2)
2PR-PD6	3.15 (1)	2.54 (2)
2PR-PD7	2.28 (1)	2.45 (2)
2PR-PD8	1.75 (1)	2.01 (2)
Mean	2.11	2.21
S.D.	± 0.64	± 0.71

K_s, stiffness as tested in saline; K_a, stiffness as tested in air. Number in parentheses indicates order of testing.

group, the order of testing was reversed. In all cases the stiffness of the beam (defined as the load divided by the midspan deflection) was computed. Table 2 summarizes the results.

Although the stiffness of the specimens tested in normal saline was slightly less than that of those tested in room air (4.4%), the difference was not statistically significant. Sedlin (1965) noted a similar difference in the Modulus of Elasticity of femoral specimens tested as cantilevers at 27 and 37°C. While the evidence indicates that a difference in specimen response between the two test conditions exists, the difference was not large enough to warrant the increased difficulty of testing using the warm saline bath. All specimens therefore were tested at ambient room conditions.

BENDING TEST PROCEDURE

Bending tests were performed on 86 specimens obtained from 6 fetal calvariums. In addition, 12 full-section specimens from the parietal bone of a 6-yr-old child were tested for comparative purposes.

All test equipment was allowed to warm up for 1 hr before calibration and testing. The load transducer was calibrated with suspended weights. The LVDT displacement transducer was calibrated in place using a micrometer head rigidly mounted to the side frame of the test machine (Instron Floor Model TT-C). Each specimen (in its container of normal saline) was allowed to equilibrate to room temperature for 1 hr prior to the test. During testing, each specimen was kept moist by a constant drip of normal saline. The test procedure was as follows:

(1) The specimen was centered on the bending supports of the test fixture with the convex surface upwards.
(2) The specimen was cycled 3 times to a midspan deflection of 0.25 mm at a crosshead speed in both loading and unloading of 0.5 mm/min.
(3) The specimen was loaded to a midspan deflection of 1.5 mm at a crosshead speed of 0.5 mm/min and then unloaded at the same speed.
(4) The specimen was removed from the fixture and returned to its storage container.

METHOD OF DATA REDUCTION

The method of Unit Loads was utilized to account for the specimen thickness variability. For a beam in pure bending, the load deflection equation can be expressed as:

$$y = \int_0^l \frac{Mm}{EI}\,dx, \tag{3}$$

where

y = deflection at the desired location (mm),
l = span length (mm),
M = bending moment for the applied load distribution (N-mm),
m = bending moment created by a unit point load acting at the location where the deflection is to be found (N-mm),
E = elastic modulus (MPa),
I = area moment of inertia (mm^4),
x = position along the span (mm).

For a simply-supported beam with a concentrated load at midspan and a constant Modulus of Elasticity, equation (3) can be written as

$$\delta = \frac{P}{4E}\left[\int_0^{l/2} \frac{x}{I}\,dx + \int_{l/2}^{l} \frac{(l-x)^2}{I}\,dx\right], \tag{4}$$

where

δ = deflection at midspan (mm),
P = applied load at midspan (N),
x = linear coordinate along the beam (mm).

Solving equation (4) requires a knowledge of the inertia of each beam section. Since the specimens had a randomly varying thickness, the beam was divided into sections of constant inertia.

Subdividing the beam into 4 equal segments and solving equation (4) yields

$$\delta = \frac{Pl^3}{768E}\left[\frac{1}{I_1} + \frac{7}{I_2} + \frac{7}{I_3} + \frac{1}{I_4}\right], \tag{5}$$

where

I_1 = inertia of section 1 (mm^4),
I_2 = inertia of section 2 (mm^4),
I_3 = inertia of section 3 (mm^4),
I_4 = inertia of section 4 (mm^4).

The accuracy of this method was compared with both an 8 element model based on finite element methods and an 8 section model using the unit load method. An analysis of several randomly selected specimens indicated that the results for the 4 section model deviated from the other more complex models

Table 3. Effect of gestational age and fiber orientation on the elastic modulus as determined from bending tests

Calvarium no.	Estimated gest. age (weeks)	Bone	Fiber orientation	No. of specimens	Elastic modulus ± S.D. $\times 10^{-3}$ (MPa)	Coefficient of variability (%)
1	40 ± 2	PR	PA	3	4.01 ± 1.28	31.9
		PR	PD	3	1.74 ± 0.59	33.9
		FL	PA	2	3.05 ± 0.88	28.9
		FR	PD	2	1.70 ± 0.79	46.5
2	38 ± 2	PR	PA	9	4.24 ± 0.73	17.2
		PR	PD	8	0.84 ± 0.19	22.6
3	27 ± 2	PR	PA	10	0.94 ± 0.41	43.6
		PR	PD	3	0.18 ± 0.03	16.7
4	25 ± 2	PR	PA	8	1.30 ± 0.60	46.2
		PL	PD	3	0.12 ± 0.01	8.3
6	28 ± 2	PR	PA	5	3.62 ± 0.46	12.7
		PR	PD	5	0.14 ± 0.08	57.1
7	40 ± 2	PR	PA	5	3.72 ± 0.35	9.4
		PL	PA	5	3.30 ± 0.64	19.4
		PL	PD	5	0.57 ± 0.14	24.6
		FR	PA	5	2.83 ± 0.96	33.9
		FL	PA	5	3.29 ± 0.71	21.6
8	6-yr-old child	PL	PA*	6	7.38 ± 0.84	11.4
		PL	PD*	6	5.86 ± 0.69	11.8

(1) Bone: PR— parietal, right; PL — parietal, left; FR — frontal, right; FL — frontal, left.
(2) Fiber orientation: PA — parallel to long axis of specimen; PD — perpendicular to long axis of specimen; * — for calvarium No. 8, PA designates specimens parallel to sagittal suture, while PD designates specimens perpendicular to sagittal suture.
(3) Test conditions: three point bending; deflection rate — 0.5 mm/min; tested at room temperature while being kept moist with normal saline drip.

by a maximum of 2%. Hence this simpler model was employed in all subsequent data reduction.

The rationale behind the bending data reduction was to compute a fictitious beam stiffness from the specimen geometry data, assuming an elastic modulus of 6895 MPa. The actual beam stiffness was measured from data obtained from the bending test. The ratio of these two stiffnesses was a direct measure of the elastic modulus of the material.

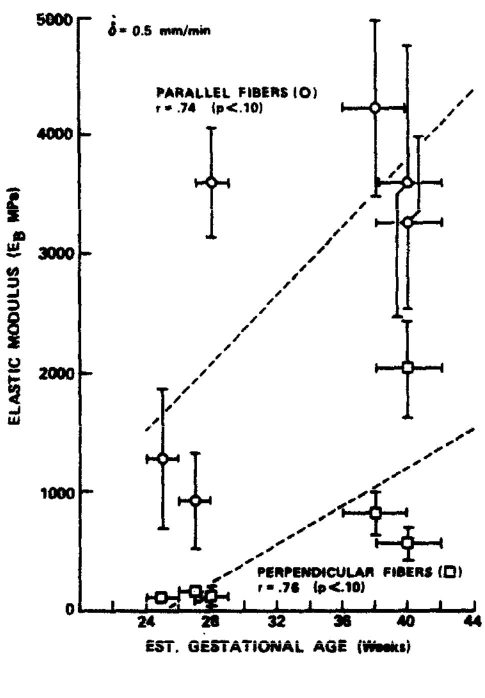

Fig. 1. Effect of gestational age and fiber orientation on the elastic modulus as determined from bending tests. Vertical bars represent ± 1 S.D. of mean value of modulus; horizontal bars represent range of estimated gestational age.

RESULTS

The elastic modulus as determined in bending was calculated for 86 specimens from 6 calvariums. The results are tabulated in Table 3 and illustrated in Fig. 1. This figure depicts the effects of gestational age and fiber orientation on the elastic modulus.

The data were subdivided into two groups according to estimated gestational age (E.G.A.). Data from specimens between 24 and 30 weeks E.G.A. were classified as preterm, while data from specimens between 36 and 40 weeks E.G.A. were classified as term. A Student t-test of the grouped data indicate that highly significant differences ($P < 0.001$) exist for the elastic modulus between term and preterm material, regardless of fiber orientation. Similarly, highly significant differences ($P < 0.001$) in the elastic modulus exist between parallel and perpendicular fiber orientation specimens, regardless of gestational age (Table 4).

A first order least-squares curve was fitted to the experimental data for both parallel and perpendicular fiber orientation specimens. A moderate coefficient of determination exists for both parallel fibers ($r = 0.74$, $P < 0.10$) and perpendicular fibers ($r = 0.76$, $P < 0.10$).

The results of the testing for the 6-yr-old parietal

Table 4. t-testing the effect of gestational age and fiber orientation on the elastic modulus (in bending) of fetal parietal bone (standard deviations in parentheses)

Gestational age group	Elastic modulus in bending E_B (MPa) Parallel fibers	Perpendicular fibers	Statistical significance
Pre-term	1650 (1170) $n = 23$	145 (62) $n = 11$	$P < 0.001$
Term	3880 (780) $n = 22$	951 (572) $n = 16$	$P < 0.001$
Statistical significance	$P < 0.001$	$P < 0.001$	

bone are listed in Table 3. A paired Student t-test of the data from the 6-yr-old indicates that a significant difference ($P < 0.001$) in elastic modulus exists between specimens oriented parallel to the sagittal suture and those oriented perpendicular to the sagittal suture.

DISCUSSION

Fetal cranial bone is a very thin, non-homogeneous and highly curved material with a distinctly oriented fiber pattern. As such, it presents unique problems in determining the mechanical properties of the material. The typical "dumbell" specimens used in uniaxial tension tests cannot be fabricated from the highly curved structure. More indirect test methods are necessary.

Simple (or three-point) bending tests provide a method to alleviate some of the difficulties. Bending tests, however, are primarily a structural test and material properties can only be inferred using assumptions about the material's structure. While Euler–Bernoulli bending theory provides a method for handling a wide variation in structural and material parameters, solutions to the differential equations require constraining simplifications. Typically, the beam is assumed to be of uniform cross-section, and composed of homogeneous isotropic material. None of these criteria are met for the bending specimens fabricated from fetal cranial bone.

In our study, certain of these variables have been taken into account by using approximate methods based on energy conservation. The method of Unit Loads was utilized to allow the varying beam thickness to be approximated in the deflection equation. This requires a simulation of the continuously varying beam thickness by a discrete number of constant thickness elements. Comparative studies have shown that the use of 4 elements results in only a 2% difference from the more refined 8 element model, hence justifying the use of this simpler model. Nevertheless, some error in the final result is inevitable because the actual continuous variation in beam thickness is not modeled.

The fiber pattern of fetal cranial bone is radially oriented within each bone and the focus of the pattern lies at the center of ossification. For example, in the parietal bone, the fibers radiate from the parietal eminence. In fabricating a specimen of finite width, it is therefore impossible to have all of the fibers running either parallel or perpendicular to the long axis of the specimen. Deviations from this ideal condition can be severe, particularly for perpendicular fiber oriented specimens. While every effort was taken to minimize this variable during specimen fabrication, the variations which did exist were ignored in the resulting analysis. Furthermore, it is not possible to easily account for the variations in bone porosity. Therefore, the analysis utilizes the simplifying assumptions of homogeneity and isotropy.

All of these factors, in conjunction with the variability of the material itself, result in the wide range of coefficient of variability seen in Table 3. Nevertheless, when these data are plotted (as shown in Fig. 1), distinct trends become evident. Linear regression analyses indicate that there is a significant increase in the elastic modulus with gestational age, regardless of the fiber orientation of the specimen. Conversely, there is a significant difference in the elastic modulus with fiber orientation, regardless of the gestational age of the specimen. While data are obviously limited, this trend seems to continue at least through the sixth year of life. Wood (1971) was able to discern no such difference in the adult—a finding which is supported by the homogeneous grain structure of adult cranial bone.

No conclusions can be drawn from these data regarding the homogeneity of response of the material between cranial bones of the same skull. Examination of the data for calvarium No. 7 in Table 3 (a well-ossified term skull) indicates a range of mean elastic modulus values as well as coefficients of variability. Conclusions regarding homogeneity of response must await more sensitive test methods.

The results of this study allow several hypotheses to be formulated relating to clinical observations and procedures. First, the significantly lower modulus values exhibited by the preterm specimens may be one factor in explaining why preterm infants are at a higher risk of receiving cerebral trauma during parturition than are term infants. In the preterm fetus, the lower value of the elastic modulus coupled with the thinner structure of the cranial bones allow much greater deformation of its head. This conclusion may be one contraindication for the use of forceps to "protect" the preterm fetal head during delivery. Inadvertent pressure from forceps application may create the situation that the forceps are intended to prevent.

Secondly, the radially oriented grain structure with its associated variation in material properties of the cranial bones and platelike nature are teleologically well-adapted to the deformations required for the safe passage in the birth canal. Abnormal variations in any of these factors, however, can have adverse consequences on the molding process and ultimately on the well-being of the fetus.

It is well-established that extreme molding in the term fetus is retarded by the mechanical interactions of adjoining skull bones. As molding progresses and extremes are reached, the frontal and occipital bones interlock at the temporal and lambdoidal sutures, respectively. In addition, the parietal bones interlock at the sagittal suture. Thus, the head, after optimally adapting to the conformation of the birth canal, begins to act as a more rigid body in protracted labors, preventing further molding and protecting the brain. In the case of the premature fetus, however, the skull bones are not well ossified, especially at the margins, and the sutures are wider. Thus the interlocking mechanism does not create as rigid a structure, and for this reason, too, the premature infant is at greater risk to birth trauma. This reasoning would substantiate the argument of contemporary obstetricians who feel that contraindications to forceps delivery is a protracted labor and a premature labor.

CONCLUSIONS

This study indicates that the development and maturation of the human skull is a continuous process in terms of both structural and material properties. The values for the elastic moduli as determined from three-point bending tests indicate that:

(1) the mean value of the elastic modulus (as determined from bending tests) for specimens with the long axis of the specimen oriented parallel to the grain pattern of the bone ranges from 1.65×10^3 MPa for preterm bone to 3.86×10^3 MPa for term bone;
(2) there is a highly significant difference ($P < 0.001$) in the modulus between term and preterm cranial bone, regardless of specimen orientation with respect to the fiber of the bone;
(3) there is a highly significant difference ($P < 0.001$) in the elastic modulus between parallel and perpendicular fiber oriented specimens, regardless of gestational age;
(4) the difference in elastic modulus with fiber orientation seem to persist at the sixth year of life but is not evident by adulthood;
(5) the significant differences in properties which exist between the preterm and term material could be one factor in the explanation as to why preterm infants are more at risk for cerebral trauma than are term infants.

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