MATERIAL PROPERTIES OF THE INFANT SKULL AND APPLICATION TO NUMERICAL ANALYSIS OF PEDIATRIC HEAD INJURY

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ABSTRACT

The impact response of the infant head depends not only on its unique geometry, but also on the age-dependent mechanical properties of the skull and sutures. In the present study, human tissue samples containing cranial bone and sutures were taken from fresh surgical specimens and prepared as miniature specimens for mechanical testing in bending and tension. The constitutive behavior for experimentally-determined infant skull and suture tissue was implemented in three-dimensional finite element models of the pediatric head to examine the sensitivity of skull and brain strains to variations in the impact direction. The head impact simulations demonstrated the directional dependence of skull fracture and traumatic brain injury risk.

ALTHOUGH IMPACT-RELATED HEAD INJURY in adults has been studied extensively, researchers have only recently begun to quantitatively assess the mechanisms of head injury unique to the pediatric population (Thibault, 1997a,b). The pediatric skull differs markedly from the adult skull in geometry, structure, and material properties (Kraus, 1990; Luerssen, 1991; Walker, 1985). The neonatal skull is a loose aggregate of thin pliable bone plates connected by sutures. The compliant structure is capable of substantial deformation during childbirth, as well as during traumatic impact loading. At birth, the cranial bones are thin, flexible, and contain no diploe. As the child grows, the cranial bone differentiates into inner and outer layers of compact bone that enclose a middle layer of cancellous bone (Hubbard, 1971a; McElhaney, 1970).

Cranial sutures are junctions between the cranial bones. At birth, cranial sutures permit overlap of the calvarial bones allowing passage through the birth canal. As cranial sutures develop during infancy, cranial adjustment to the expanding brain takes place by bone deposition at the sutural margins. Mature sutures are highly interdigitated and capable of absorbing energy during impact loading (Hubbard, 1971b; Jaslow, 1990).

Characterization of the mechanical properties of adult human and mammalian skull (Evans, 1957), suture (Hubbard, 1971b; Jaslow, 1990), and brain (Fallenstein, 1969; Ommaya, 1968; Shuck, 1972) are well documented in the literature. However, the fontanelles, uncalcified non-interdigitated sutures and the thin, flexible dural-cranial bone layer for the undeveloped cranium of the young child allow significant shape change to occur in the skull structure under quasistatic and dynamic loading. Further research is needed to fully characterize the mechanical properties of cranial bone, sutures, and brain in the young child for a wide range of age and loading conditions (McPherson, et al., 1980, Thibault, 1997a). The application of numerical modeling techniques to study pediatric head injury has been limited by the availability of material property data, as well as by the complexity of the structural features characteristic of the developing pediatric braincase.

In this study, experiments were first conducted on human skull and suture tissue to characterize their age-dependent material properties. The biofidelic constitutive behavior was then incorporated into three-dimensional finite element models of the pediatric skull-brain structure. The head models were subjected to lateral and posterior impacts to investigate the directional dependence of skull and brain injury risk under traumatic loading conditions. The long-term goal of this research is to develop infant skull finite element models for the investigation of injury thresholds in the pediatric population.

MATERIALS AND METHODS

EXPERIMENTS. Normal human infant cranial bone and suture tissue was obtained from surgical procedures performed at St. Christopher's Hospital for Children, Philadelphia, PA. Specimens were donated for this study and authorized during the general consent for the initial surgical procedure. The protocol for this research was approved by the IRB at Allegheny University of the Health Sciences. Subjects ranged from three months to seven years of age. For each subject, age, sex, primary diagnosis, and nature of the surgical procedure and site of the specimen were recorded (Table 1). Subjects with a history of trauma or metabolic disease were excluded.

Bone samples were excised using scissors or craniotome (Midus Rex) and placed in a sealed container wrapped in gauze pads soaked with phosphate buffered saline. The samples were then refrigerated at 4°C until preparation for mechanical testing. Three-point bending specimens and tension specimens were prepared from the excised bone samples using a high speed Dremel Multi-Pro cutting wheel under a constant drip of saline. Test specimens were obtained from normal parietal bones, occipital bones, coronal sutures, sagittal sutures, and metopic sutures.

Once obtained, specimens were tested in either three-point bending or tension, using custom-designed fixtures, a Model 858 Bionix Materials Testing

System (MTS Systems Corp., Eden Prairie, MN) with a 25 pound load cell (Lebow-Eaton, Troy, MI), capable of up to 100 Hz oscillation for small displacements. Force and centerline deflection were recorded with a digital data acquisition system. Before testing, bone samples were warmed to 25°C in their sealed storage containers. Typical specimen dimensions were approximately 20 mm long by 7 mm wide by 1-2 mm thick.

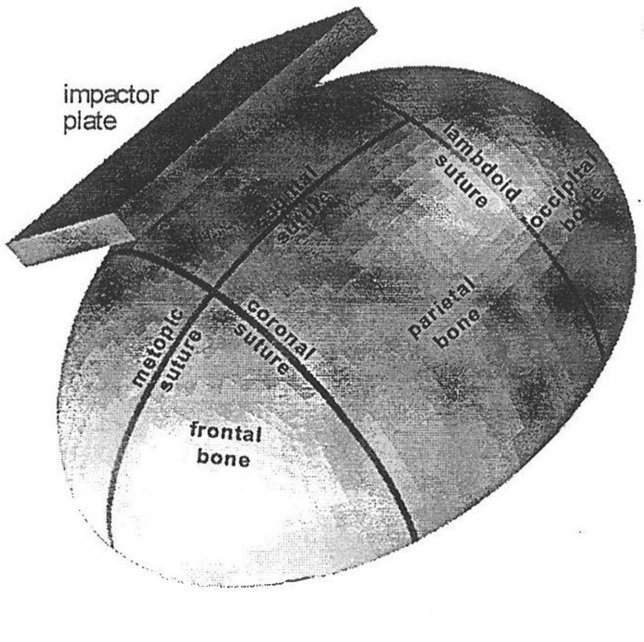
Table 1. Summary of Age and Diagnosis of the Bone Sample Donors (B=bone; S=suture)

Donor	Age (mo)	Diagnosis	Procedure	N
S2	15	Sagittal Cranio-synostosis	Pi Procedure	3 (S)
S3	7	Sagittal Cranio-synostosis	Strip-Craniectomy	20 (S)
S4	5	Sagittal Cranio-synostosis	Strip-Craniectomy	2 (B) 19 (S)
S5	60	Hydro-cephalus	Decompression / Craniectomy	6 (B)
S6	3	Sagittal Cranio-synostosis	Strip-Craniectomy	4 (B) 22 (S)
S7	84	Chiari-I malformation	Decompression / Craniectomy	3 (B)

Cranial bone was tested in three-point bending at loading rates between 0.3 and 30 mm/sec deflection. Tensile tests were conducted on sutured specimens at axial loading rates between 0.03 and 30 mm/sec. Only data from the 30 mm/sec experiments are presented here. Cranial bone specimens were analyzed using simple beam theory to determine elastic modulus, rupture modulus and energy density absorbed up to failure. Analysis of the suture specimens was limited to reporting of the force-deflection characteristics.

The three-point bending data were further analyzed through the use of the inverse finite element method (Hou, 1996; Kurtz, 1997) to infer the elasto-plastic behavior of infant cranial bone. A three-dimensional finite element model consisting of 486 nodes and 186 solid hexahedral elements was constructed based on the geometry of the beam bending experiment. Simulations were performed using LS-NIKE3D V980 (LSTC, Livermore, CA). Frictionless contact was included between the cylindrical indentor and the bone specimen. Initial parameter studies demonstrated that predicted load-displacement behavior of the virtual beam bending experiment was not sensitive to friction at the indentor/specimen interface. Convergence studies were performed to verify the spatial and temporal stability of the contact solutions, and the linear mechanical response of the model was validated using beam theory.

The elasto-plastic behavior of infant cranial bone was modeled using isotropic, rate independent plasticity. The true (effective) stress-strain behavior was idealized as bilinear elastic-plastic model, with strain hardening in the plastic regime. The predicted force–deflection behavior of the simulation was compared to the experimental data to determine the elastic modulus (E), yield stress (σ_y), and hardening modulus (E_h) of each bone sample. The three material model parameters (E, σ_y and E_h) were parametrically varied until optimal matching was obtained (in a least squares sense) for the experimental force–deflection data up to failure.



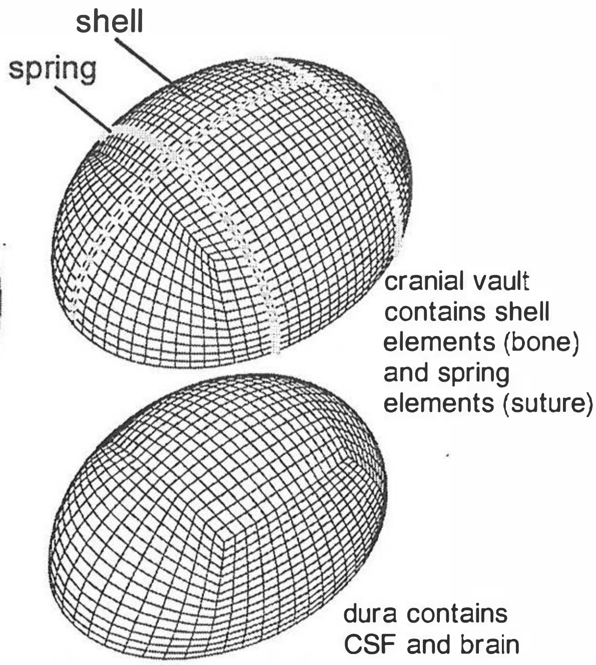


Figure 1. Isometric view of three-dimensional idealized finite element model of the infant head impacted laterally at 45°. The mesh was generated in the TrueGrid platform-independent modeling environment.

Figure 2. Exploded view of three-dimensional idealized finite element model of the infant head depicting the separate structures of the brain and skull.

ANALYSIS. A finite element model of the three-month-old infant head was constructed using TrueGrid (XYZ Scientific Applications, Livermore, CA) to simulate 45° lateral and posterior impacts (Figure 1). The infant skull consisted of five deformable bone plates (occipital bone, left and right parietal bones, and two frontal bones) which were interconnected by linear springs to model the sutures. Thus, the sutures in the model were designed to support tensile forces but were not intended to resist bending moments. The foramen magnum was included in the base of the skull. Contacting interfaces were included between the rigid impactor and the skull, as well as the skull and the dura.

The model consisted of 30,153 nodes: 25,279 eight-noded hexagonal solid elements for the indentor, brain, and CSF; 5,514 four-noded quadrilateral shell elements for the bone plates, foramen magnum, and dura; and 137 two-noded, one-dimensional spring elements for the sutures (Figure 2). Frictionless contact between the indentor, skull bones, and the dura-brain continuum was modeled using LS-DYNA3D (Livermore Software Technology Corporation, Inc., Livermore, CA). Zoning studies were conducted to verify convergence. Typical solution times were approximately 9 hours on an Origin 200 Silicon Graphics server equipped with a 180 MHz R10000 CPU.

The elasto-plastic material properties of infant cranial bone were based on our skull tissue experiments performed on tissue obtained from a three-month-old infant donor. For elastic deformations, the Poisson ratio of the bone was taken to be 0.28. The suture stiffness was based on the tensile properties of coronal suture tissue for a three-month-old infant. The brain material properties were based on the experimentally determined mechanical response of human brain tissue (Galford, 1970). The brain was modeled as a linear viscoelastic solid: $G(t) = G_\infty + (G_o - G_\infty)e^{-\beta t}$, where G_o = 5.99×10^{-3} MPa, G_∞ = 2.32×10^{-3} MPa, and $\beta = 9.43\times10^{-2}$. The brain was assumed incompressible with a bulk modulus of 2110 MPa. A membrane spanning the foramen magnum was used to simulate the mechanical impedance of the spinal cord resisting brainstem herniation. A conservative approximation of the membrane stiffness was 100 MPa, which is an order of magnitude stiffer than central nervous system tissue. Because this is still an order of magnitude more compliant than pediatric cranial bone, further reduction of this value, even to zero, did not substantially affect the overall response of the model.

The finite element model was exercised with two load cases to simulate lateral and posterior impacts. A peak impact force of 1000 N was used for each of the load cases, based on accelerations measured by Duhaime and colleagues (Duhaime, 1987). The load was applied to the peak level in a half-sinusoidal, 10 ms pulse. In the simulations, loads were applied to the parietal and occipital regions of the skull, respectively, at 45° relative to the vertical axis. For both cases, the base of the cranial vault was fixed to simulate pure impact loading with no rotation or translation of the head following impact.

RESULTS

EXPERIMENTS. Our data thus far shows that human cranial bone is extremely compliant during the first few months of life. When the current data are presented in concert with values presented in the literature, it becomes apparent that the elastic modulus of cranial bone increases dramatically during pre-pubescent growth and development (Figure 3).

The inverse finite element simulation of the three-point bending test of three-month-old parietal bone resulted in a force-deflection curve that accurately matched the experimental data up to failure (r^2>0.995, Figure 4). The numerically determined properties for the simulated bone sample were E=880 MPa, σ_y=12 MPa and E_h=50 MPa. The hardening modulus, E_h, corresponded to a tangent modulus of 47 MPa in the plastic portion of the model. The maximum principal strain at failure was 0.158, and the maximum principal stress at failure was 18.5 MPa, corresponding to a plastic strain of 0.126 at failure (Figure 5).

Coronal suture specimens from the three-month-old donor, tested in tension, indicated a non-linear behavior typical of biological materials (Figure 6). The slope of the linear portion of the curves shown was determined and averaged to obtain a mean stiffness of 189 N/mm.

ANALYSIS. For both the lateral and posterior impact sites, the 1000 N impact load resulted in plastic deformations of the skull bones. During the lateral impact, the compliant infant brain case permitted plastic skull deformations in locations remote from the site of impact (Figure 7). The parietal bone was subjected to plastic bending in the basilar region, with a maximum effective plastic strain of 0.07. Plasticity was involved, albeit to a much lesser extent, in the deformations of adjacent bone plates which communicated with the lateral impact site via the sutures. The deformable brain case also resulted in a diffuse distribution of strain throughout regions representing the cortex and brainstem (Figure 8).

The deformations of the skull and braincase were sensitive to the loading orientation, with the posterior loading case producing the greatest strains in the bone and brain. Plastic strains in the posterior impact site were found to exceed 0.126, which was experimentally determined to be the maximum effective plastic strain for infant cranial bone.

DISCUSSION

A finite element model of the three-month-old infant head was developed with sufficient versatility to enable systematic exploration of the effects of infant braincase material properties on impact response and injury thresholds for the skull and brain. This is the first study to explore the effects of skull plasticity on the mechanical response of the infant brain case to impact. Previous finite element analyses of the infant skull have assumed linear elastic behavior for the cranial bone (Thibault, 1997b), which can be expected to overpredict the stiffness of the skull, and hence underpredict the strains in the underlying brain tissue. The elasto-plastic model of the skull, when coupled with a failure criterion (such as the threshold for maximum plastic strain) will enable estimation of skull fracture tolerances as a function of load, loading rate, and impact site.

The ultimate goal of this research is to develop a family of structural models for the pediatric population as a function of age. Validation of these models is complicated by the difficulty in direct measurement of injury tolerances. In the absence of experimental dynamic impact data for infant skulls with which to validate the global structural model, the importance of using experimentally measured material properties for the constituents of the model is paramount. In this study, the nonlinear material properties of the skull and sutures were based on experiments on tissue from a three-month-old infant. The remaining material properties, including those for the brain and dura, were based on prior experiments on adult tissue. Future parametric studies will establish the sensitivity of the human cranial impact response as a function of the dynamically changing anatomy, physiology and material properties of the developing pediatric head.

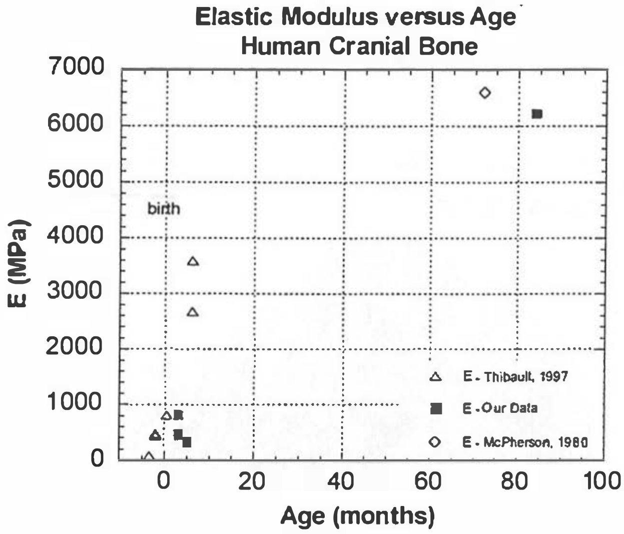


Figure 3. Elastic modulus of human cranial bone as a function of age, our data compared with values from the literature. Zero (0) months indicates birth (i.e., 40 weeks gestational age).

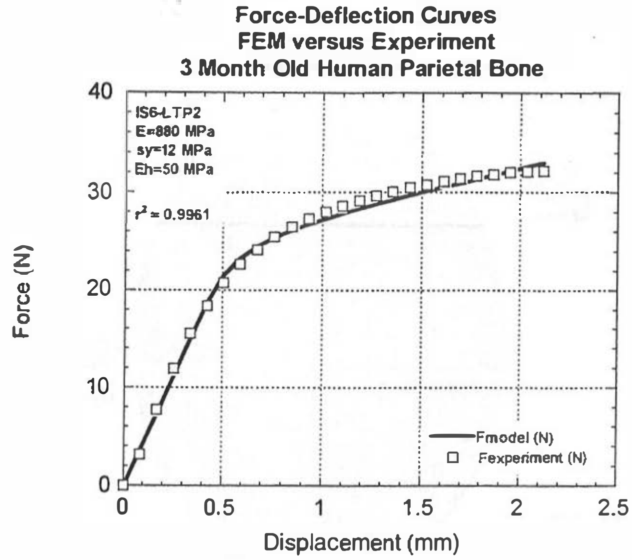


Figure 4. Force-deflection behavior of three-month-old parietal bone tested in 3-point bending at 30 mm/sec loading rate. Hollow squares represent the experimental data, the line represents the numerical material simulation. The simulation accurately matched the experiment ($r^2 > 0.995$).

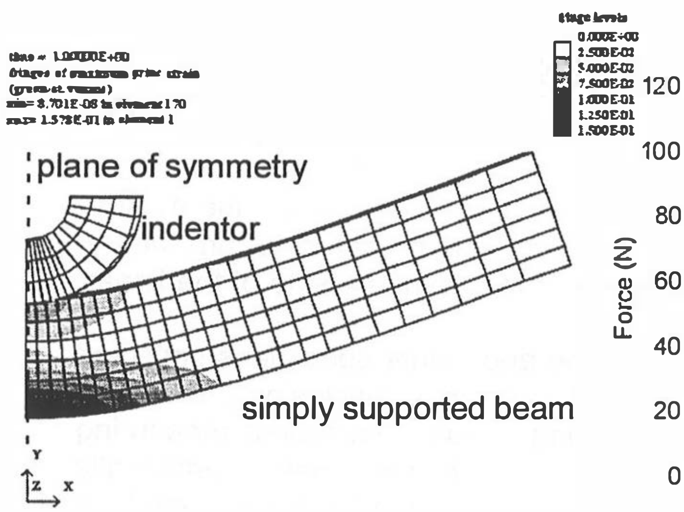


Figure 5. FEM beam bending simulation with fringes of maximum principal strain depicted at the failure deflection for parietal bone from a three-month-old donor. The maximum principal strain at failure was 0.158 on the tensile surface under the indentor location (midspan).

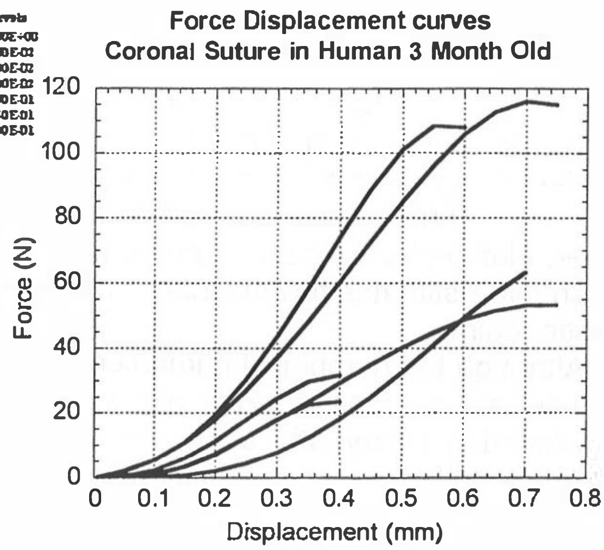


Figure 6. Typical force-displacement curves for coronal sutures in tension from a three-month-old donor. The mean value of the stiffness (linear portion of the curves) was approximately 189 N/mm.

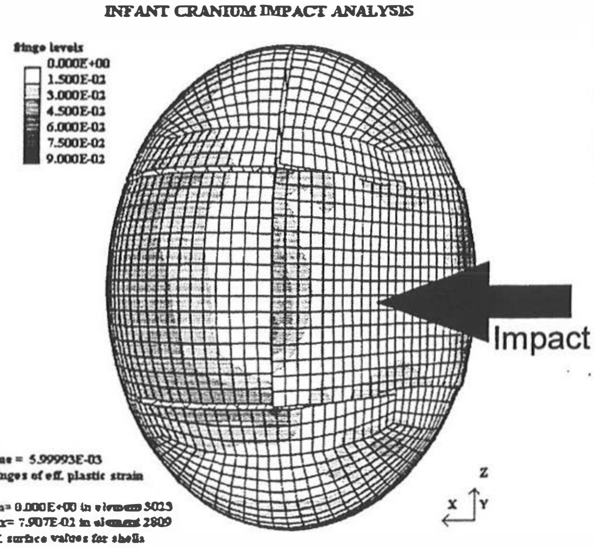


Figure 7. Fringes of effective plastic strain for the three-month-old numerical simulation of lateral impact loading, top view of the cranial vault.

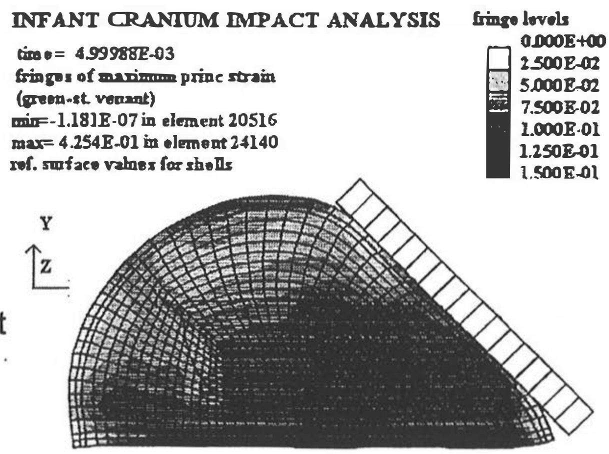


Figure 8. Fringes of maximum principal strain in a mid-brain coronal section of the three-month-old infant brain, posterior view. Deformations are shown to scale.

Our laboratory continues to define the age-dependent mechanical properties of infant cranial bone and suture. These data are an integral part of modeling the response of the infant head to traumatic loading. We have established that three-point bending tests of cranial bone in combination with inverse FEM allows us to determine the unique material properties of cranial bone as a function of age. Developing cranial bone is not exclusively linear and/or elastic to failure. Through inverse FEM optimization we are able to determine its non-linear, elasto-plastic material properties without relying on simple beam theory, which overestimates the mechanical behavior of the tissue outside of the linear, elastic regime.

Although the number of human cranial bone and suture specimens collected to date is too small to draw age-dependent statistical conclusions, we have developed a systematic approach to acquiring tissue specimens, measuring experimentally the mechanical properties of each specimen, numerically simulating each experiment and determining the material behavior based on optimized inverse FEM. As evidenced in Figure 1, there are many data yet to be collected. As we continue to collect human tissue specimens of various ages, we will develop a comprehensive database of mechanical properties of the pediatric head for use in three-dimensional, age-dependent FEM representations of the pediatric head.

ACKNOWLEDGEMENTS

The authors wish to thank the Centers for Disease Control for partial funding of this work (R49CCR304684/5). In addition, the authors gratefully acknowledge Tara Khatua, Ph.D., for his generous support of this project with an internal research grant from Exponent Failure Analysis Associates. Special thanks to Brad Maker at LSTC for his technical assistance and helpful discussions regarding LS-NIKE3D. Special thanks are also due to Bob Rainsberger, Ph.D., at XYZ Scientific Applications for technical assistance and many helpful discussions.

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