Modeling Neck and Brain Injuries in Infants

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Each year, there are many reported cases of brain and cervical lesions in infants, not only from falls or other accidents but also from child abuse as a result of violent shaking (called *shaken baby syndrome*, or SBS) or a blow to the head (called *abusive head trauma*, or AHT). The most prevalent types of brain lesions due to SBS are subskull bleeding and bleeding into the retina. Vigorous shaking can also cause neck injuries. Between 7 and 30 percent of SBS cases result in death; 30 to 50 percent result in serious cognitive and neurological disorders. Roughly 30 percent of cases have a chance of recovery, but with a high risk of long-term neurological problems.

We used the finite element method (FEM) to simulate how the vibrations due to shaking alters infants' first through fourth cervical vertebrae (numbered C1 through C4, from the skull to the spine). We hypothesized that by combining FEM and vibration theory, we can predict and quantify such alterations. We also performed simulations of AHT. Here, we hypothesized that by using FEM and simulating the effects of dynamic impact, we can predict, locate, and quantify diffuse alterations in the brain.

An economical modeling tool for detecting the effects of child abuse would allow fast simulations for forensic studies. It would also facilitate medical treatment, education, legal prosecution, and the design of safety restraints and automobile air bags. In addition, it would likely help prevent infant head injuries by showing their effects.

Building the Model

There's little experimental information on the heads of babies; most investigations have been on adults. (For more on research on head injuries and modeling them, see the related sidebar.) So, we created our model of an infant brain and upper cervical spine.

To achieve an accurate model, we had to consider five things. First, a baby's head is large, relative to the rest of its body. Second, a baby's brain contains a higher proportion of water than an adult's. Third, the brain tissue is more delicate because it's developing. Fourth, the skull is thinner and less dense, which makes it less protected. Finally, the neck muscles and tendons are weak, limiting their ability to absorb sudden movements.

We created our model using 2D computed tomography (CT) slices and 3D reconstructions on the basis of our own and other research.[1,2] We imported vertical slices (transversal to the brain axis) to AutoCAD. The slices were at millimeter resolution to distinguish bone from other tissues. Then, we drew a closed line around each slice and assigned a distance of 4 to 10 mm between the lines, to form a wire structure. (We placed all 2D contour lines in their 3D positions.) Finally, we assembled the model using AutoCAD's 3D tool, which built a solid between the closed contour lines. We used a similar method to create the internal brain parts. We imported the entire volume to the Algor (now Autodesk Simulation) FEM software.

The model includes the scalp, skull, brain, spinal cord, and cerebrospinal fluid (CSF). We modeled the scalp, skull, and CSF as eight-node cube elements with a uniform thickness, with the brain matter as the internal tissue. The skull's thickness is 2 mm, based on the CT measurements. We didn't include the fontanels (areas without bone). Because the brain is immersed in CSF, we built the model to show how the fluid holds the brain. We modeled the head components as continuous

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Related Work in Head Injuries

Sebastien Roth and his colleagues recently studied *abusive head trauma* (AHT)—the effects of blows to the head's occipital area—which often happens at the end of aggression against infants such as shaking.[1] Sujan Fernando and his colleagues identified the neurologic consequences of such injuries.[2]

Frank Meyer and his colleagues compared the most commonly used injury criteria—von Mises stress (equivalent mechanical tensile stress), maximum linear acceleration, the maximum-pressure criterion, and the head injury criterion.[3] However, none has been properly validated for infants.

Giovanni Belingardi and his colleagues, among others, studied the mechanical properties of the materials forming the head, brain, skull, and cerebrospinal fluid (CSF), and the related characterizations and constitutive models for the finite element method (FEM).[4] Timothy Horgan and Michael Gilchrist studied the overall effect of damping in an FEM model that included the skull base, subskull membrane (arachnoid), and its liquid reservoirs (arachnoid cisterns).[5] Svein Kleiven and Warren Hardy developed a model of an adult human head consisting of a scalp, skull, subskull membranes, CSF, and 11 pairs of brain vessels with internal interconnection.[6] Jean-Sébastien Raul and his colleagues demonstrated how to use FEM in forensic practice.[7]

The evolution of FEM models of brain injuries in children has reached quantitative levels—that is, it's now possible to obtain numerically precise results. Frank Meyer and his colleagues created a neck model of a three-year-old child using real geometry.[8] Roth and his colleagues created a head model of a three-year-old child.[1] Frank Meyer and his colleagues used these two models, along with a torso model, to analyze the effects of a simulated car crash, with impact against an air bag.[3]

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Table 1. Input data for the head materials.[1,3]

	Density (kg/m^3)	Elastic modulus (N/mm^2)	Poisson modulus	Shear modulus (N/mm^2)
Brain	1,040	1,930	0.4	0.00083
Cerebrospinal fluid	1,020	2,280	0.4	0.00002
Skull	1,100	7,600	0.2	2,800
Scalp	1,010	20	0.4	0.002

volumes connected by a 3D mesh. The complete model comprises approximately 22,500 elements and 25,000 nodes.

Our model incorporates the mechanical properties described in the few studies that have investigated these properties for a child's head (see Table 1). The baby's skull properties were estimated to be somewhat lower than the properties of a three-year-old child's skull because the elastic modulus varies between a newborn's skull and a three-year-old child's skull.

The biological materials (scalp, skull, and spinal cord) are isotropic and elastic, but the brain is viscoelastic (a more complex model that can include the damping effect and a nonlinear relation between stress and displacement). Giovanni Belingardi and his colleagues described shear-modulus relaxation in the viscoelastic model as

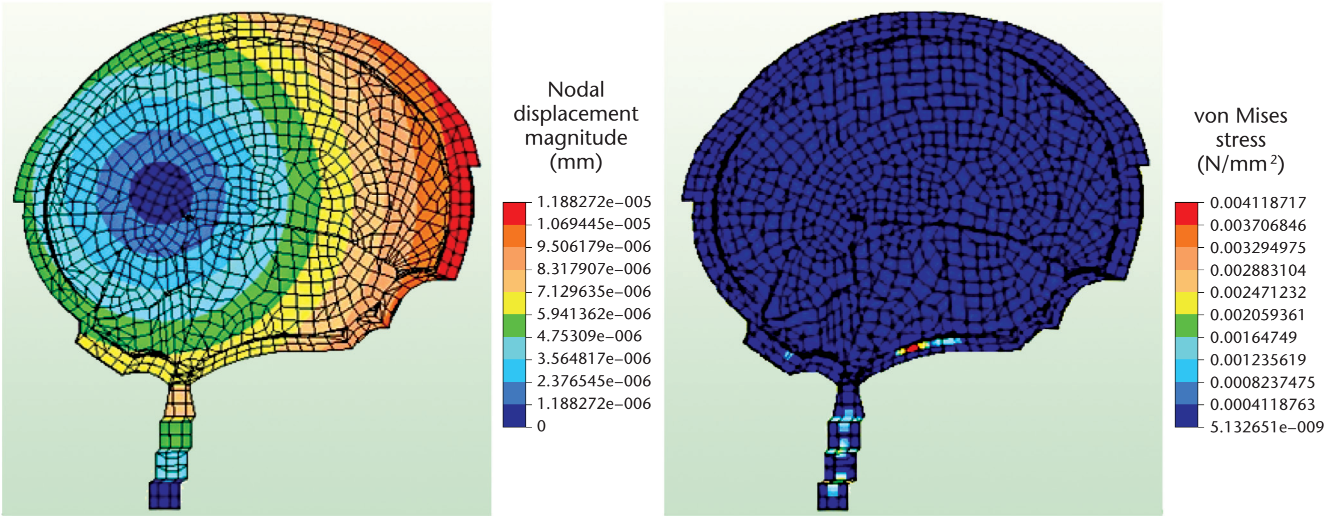

Figure 1. The simulation of vibration mode 1 of shaking a baby. The results show no risk of injury.

$$G(t) = G(\infty) + ((G(o) - G(\infty)) \exp(-\beta t),$$

where $G(t)$ is the shear modulus, as defined in static conditions; $G(\infty)$ is the long-time shear modulus (167 kilopascals); $G(o)$ is the short-time shear modulus (490 kPa); β is the decay coefficient (0.145 m/s); and t is time.[3]

Simulating Injuries

We used Algor software to simulate vibration and a head blow. The simulations yielded static images, but our analysis software let us create a short animation from the nodal-displacement images.

To analyze SBS, we subjected our model to a simulated 3-Hz vibration and 200-Newton horizontal sinusoidal force. We measured the frequency and force at the University of Tarapaca's Mechanics Laboratory in a study involving 18 students aged 22 to 25. They simulated shaking a baby, using a dummy. We analyzed six vibration modes—that is, the six natural frequencies of oscillation of the baby's head model. The simulations placed the boundary conditions at the C4 vertebra (the farthest from the head) and used the average weight of a baby at six to nine months. The simulations took roughly 35 minutes each.

For AHT, we treated the blow as a distributed load on the occipital area (a diffuse blow that leaves no outward signs). The total dynamic load was 400 N (the force of a normal adult male, extended with a slight impact coefficient equal to 2). The boundary conditions took into consideration that the head moves freely above the neck. The blow was so fast that it wasn't influenced by the restrictions on C4, so we discarded the boundary conditions restricting C4 to a fixed location for the dynamic response.[4] To avoid incorrect results, the same restrictions were also discarded, with a degree of freedom in the axis of impact. This simulation took 30 minutes.

Although knowing the real damping factor in infant head injuries is difficult, the AHT simulation had to include the damping factor. It had to take into account

- the scalp's protective action;
- the CSF;
- the brain's behavior as an elastomer;
- the absorbing effect of the ventricles (big cavities containing CSF) and other cavities; and
- the deformation of the skull, veins, arteries, and other tissues as shock absorbers.

Generally, the recommended damping factor in dynamic analysis for low-impact structures (such as the head) is 0.5.

To determine the presence of cervical and brain injuries, we measured the von Mises stress. For each node, the von Mises stress combines into one value the stresses predicted in each axis. Frank Meyer and his colleagues posited that the von Mises stress was the best indicator of brain injury for babies.[5] They determined that a von Mises stress of 0.048 N/mm^2 presents a 50-percent probability of injury and that a force of 0.080 N/mm^2 presents a 100-percent probability of injury.

SBS Results

Figures 1 through 6 show the nodal displacement and von Mises stress for the six vibration modes. The nodal displacement equals the movement in millimeters of the cube nodes in our model. The displacement and stress values can reveal the areas most likely to undergo alterations and can guide how safety equipment is designed.

Stresses exceeding 50 percent (0.048 N/mm^2) and 100 percent (0.080 N/mm^2) probability of

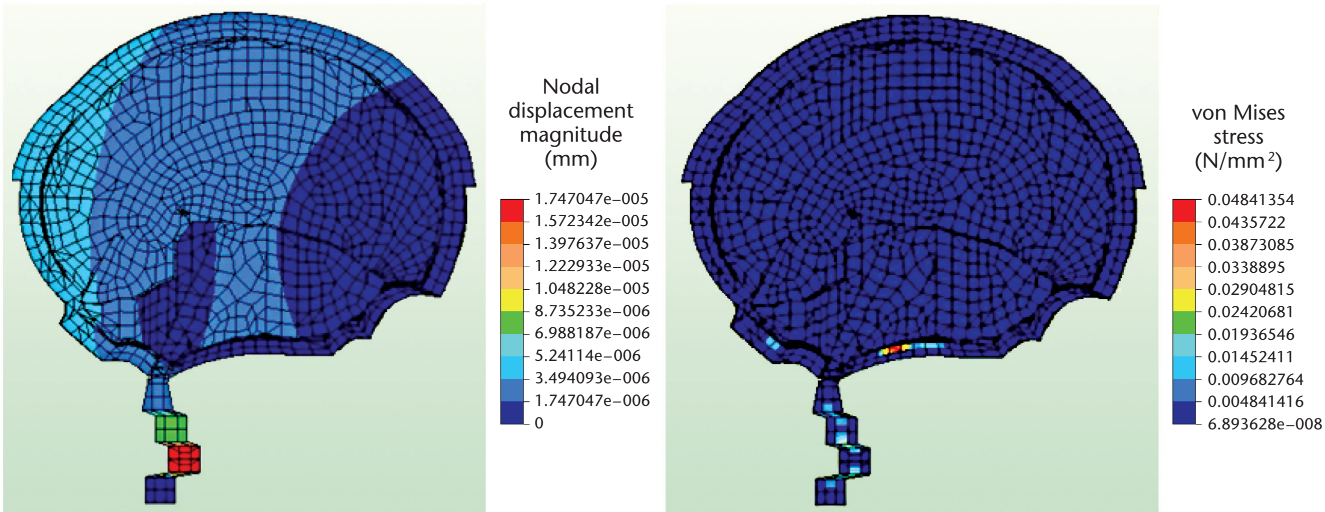

Figure 2. The simulation of vibration mode 2. The results show no risk of injury.

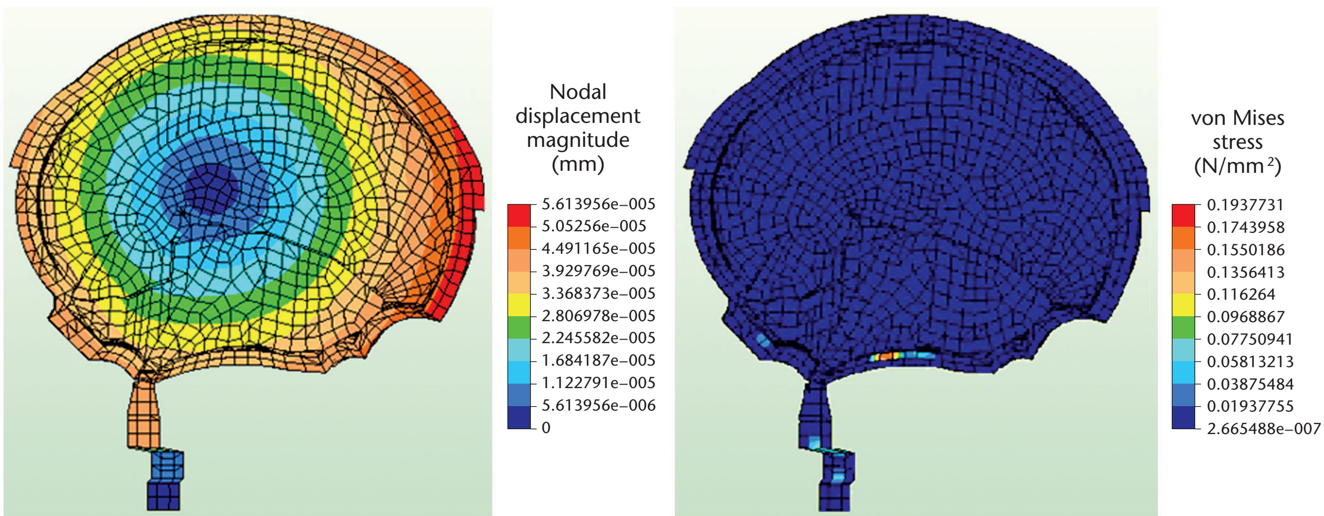

Figure 3. The simulation of vibration mode 3. The risk of injury on vertebrae C2 and C3 ranged from 0.058 to 0.077 N/mm^2, exceeding a 50 percent probability of injury (0.048 N/mm^2).

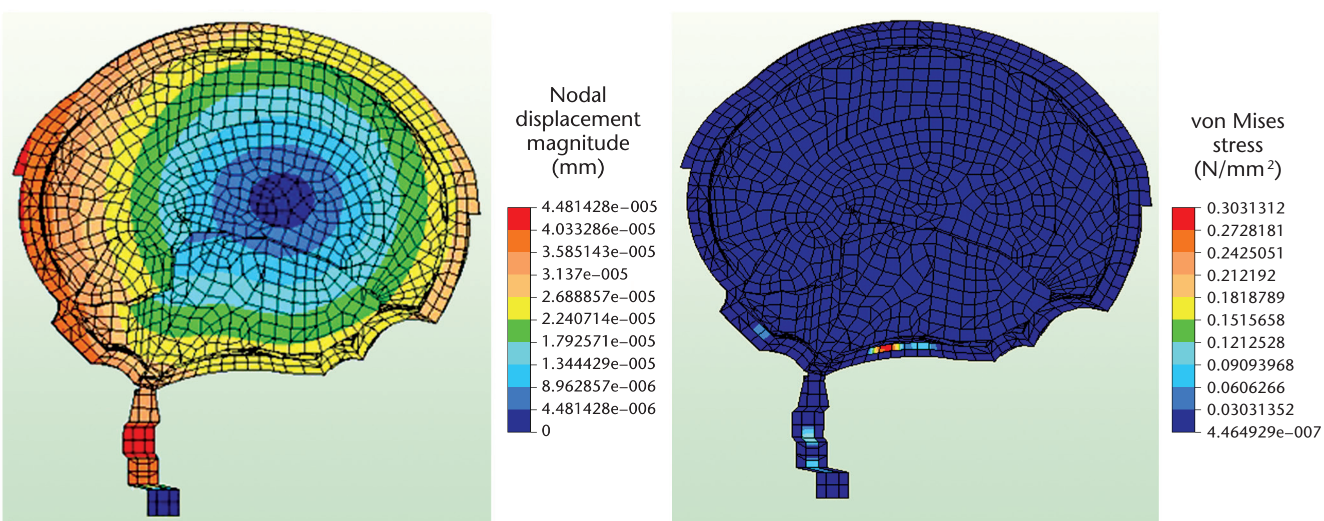

Figure 4. The simulation of vibration mode 4. The risk of injury on vertebrae C3 and C4 ranged from 121 to 0.151 N/mm^2, exceeding a 100 percent probability of injury (0.080 N/mm^2).

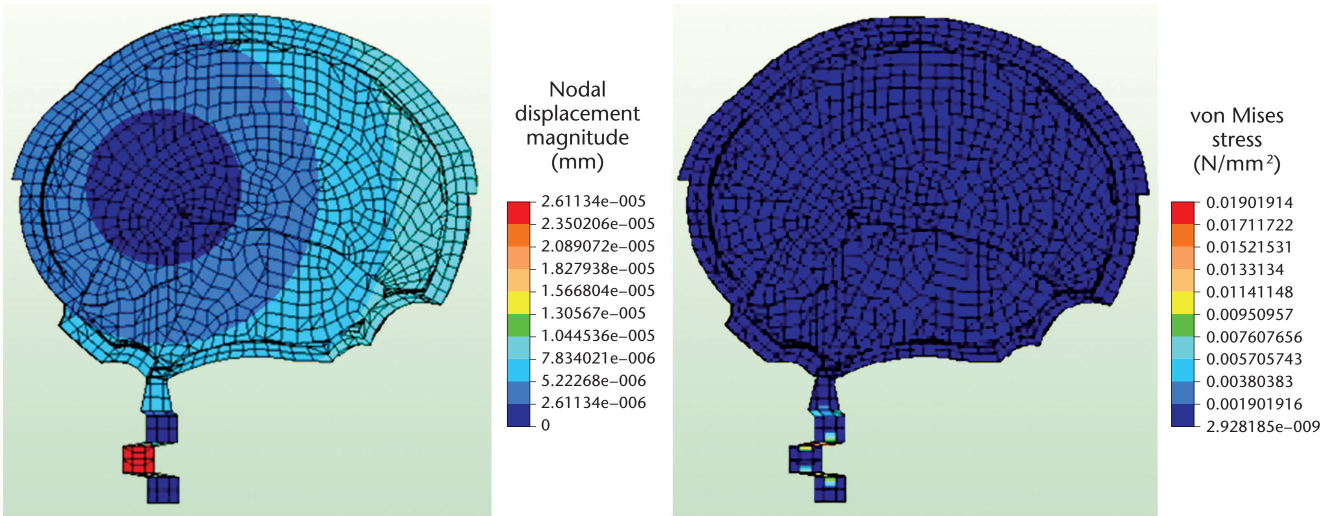

Figure 5. The simulation of vibration mode 5. The results show no risk of injury.

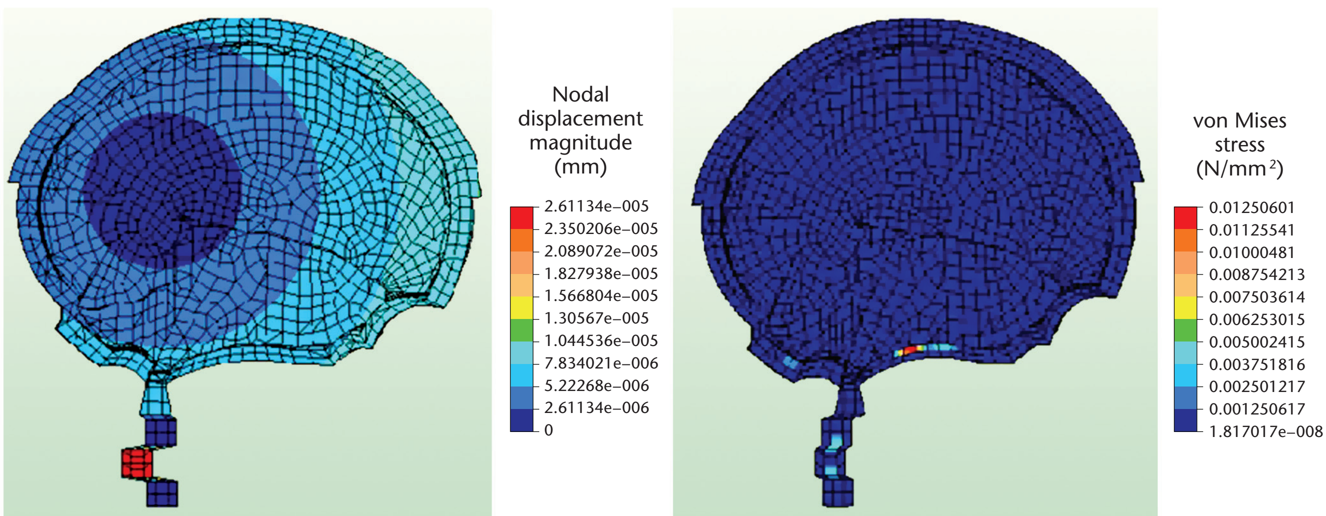

Figure 6. The simulation of vibration mode 6. The results show no risk of injury.

injury occurred in modes 3 and 4 (see Figures 3 and 4). Predictions for mode 3 were 0.058 to 0.077 N/mm^2, between C2 and C3; predictions for mode 4 were 0.121 to 0.151 N/mm^2, between C3 and C4. Only these two modes predicted alterations. In them, the vibration waves were reinforced, which weakened the vertebrae.

The CSF reached high stresses at the brain's base. However, because it's a liquid, it can absorb large shock loads and isn't destroyed under compression. The liquid stresses were 0.0041 N/mm^2 for mode 1, 0.0484 N/mm^2 for mode 2, 0.1937 N/mm^2 for mode 3, 0.3031 N/mm^2 for mode 4, 0.0190 N/mm^2 for mode 5, and 0.0125 N/mm^2 for mode 6. The stress compression limit exceeded 200 N/mm^2.

The results indicate that our model can predict and quantify cervical alterations in infants. The cervical alteration that our FEM model predicted agrees with Nima Sana and colleagues' research on cervical cord whiplash injuries, which showed that SBS can injure the cervical cord.[6]

None of the six vibration modes predicts alterations within the brain, owing to the roughness of the model, which doesn't include the network of arteries and veins in the head. But this type of injury can disturb the head's vessels. A future model should include a proper network of blood circulation with blood pressure simulation in the brain, arteries, and veins; modeling of transmitted vibrations, such as internal pressure waves; and data for blood vessel pressure.

Our model also doesn't include a layer of biological material to simulate the connections between the interior wall of the skull membranes and the brain. The relative motion between the brain and skull could lead to stresses on the arachnoid net (the skull membrane nearest the brain), which consists of fine blood vessels and nerve connections. To improve the model, we could introduce a layer between the arachnoid and the brain that resists movement. This would simulate friction, so we wouldn't need prior knowledge of the relative motion's direction. This layer should have features reflecting the arachnoid's strength during displacement between the skull and brain.

We plan to include the jaws and facial tissue in this model. This will lead to more accurate simulation of the head's moment of inertia, giving us more accurate results.

AHT Results

Figure 7 shows the nodal displacement, with the predicted maximum in the area of the blow (the occipital area). A relative displacement of 0.13 mm occurred between the tissue of the occipital lobe. From the impact area to the face, the nodal (element) displacement ranged from 1.317 to 0.790 mm—a difference of 0.527 mm.

Figure 8 shows the final diffuse stresses in a sagittal slice. The simulation predicted maximum stress from 0.23 to 0.26 N/mm² in the hypothalamus (the brain's inner part) and from 0.19 to 0.23 N/mm² in places as scattered as the frontal lobe and the cerebellum (just below the brain). The model predicted alterations to the spinal cord between C1 and C2, with stresses from 0.11 to 0.15 N/mm². Both exceeded the stress limit. The model also predicted alterations in the occipital lobe from 0.07 to 0.11 N/mm².

The results indicate we can use FEM and the dynamic effects, which model a blow, to predict, locate, and quantify diffuse alterations in the brain (and alterations to the spinal cord). When a model indicates a large nodal displacement, it can also predict the maximum injury risk in the impact area (see Figure 7). The high relative displacement between the impact area and face (0.527 mm) occurred because an infant's biological tissue is less rigid than an adult's. A 400-N blow has relatively little power (a heavyweight boxer's blow is approximately 5,000 N) but can seriously injure a toddler.

In the brain, the results indicate stresses beyond the acceptable limit (see Figure 8) for areas far from each other, suggesting neurological tissue damage. This indicates the type of alterations that shaking or blunt trauma can cause in the brain.

To validate this FEM simulation, we compared it with medical research on brain injuries due to head impacts.[7–9] In our simulation, the big nodal displacement near the impact site was consistent with large brain-skull separation that causes subskull bleeding. The simulation's accuracy was comparable to the medical researchers' real-world results.

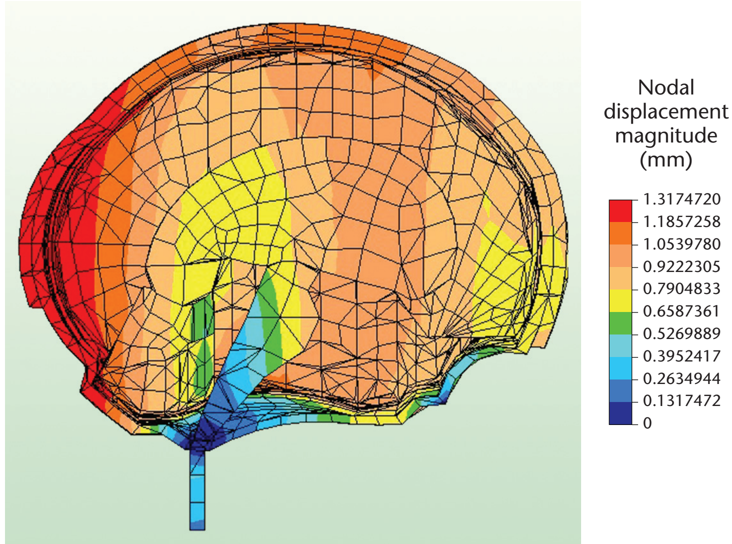

Figure 7. The initial nodal displacement caused by a 400-Newton blow to the back of the head. The red zone shows the biggest displacement at the site of the blow.

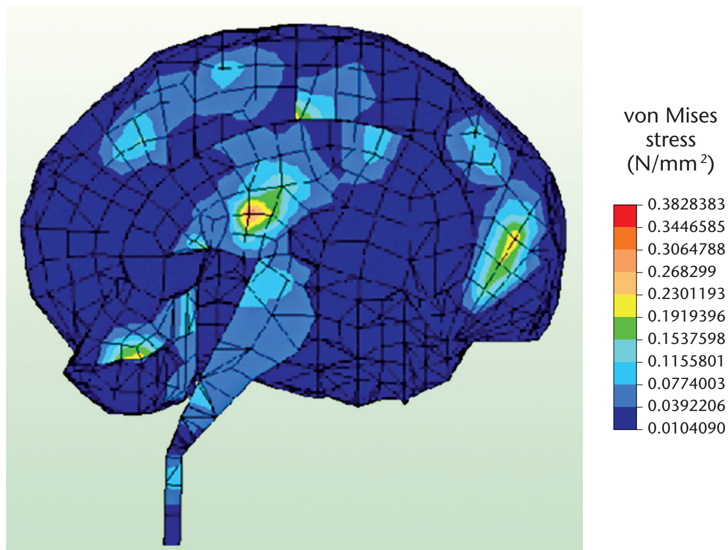

Figure 8. A sagittal slice showing the final diffuse stresses caused by a 400-N blow. There are some zones where the stress is concentrated: principally in the frontal lobe (information-processing center) and the hypothalamus (emotion-control center).

Our impact analysis model produced medically accurate findings. It works correctly when simulating impacts and shouldn't require modification as long as we apply the same dynamic loads.

FEM appears to be a practical, universal, economic, and fast tool with important forensic uses beyond detecting evidence of child abuse. For head and brain injuries, determining the damage mechanism is often difficult and normally requires sophisticated, expensive neuroimaging techniques. A good comprehension of FEM should help experts understand different head impact scenarios and impacts with different objects—including ballistic injuries—and correlate models with

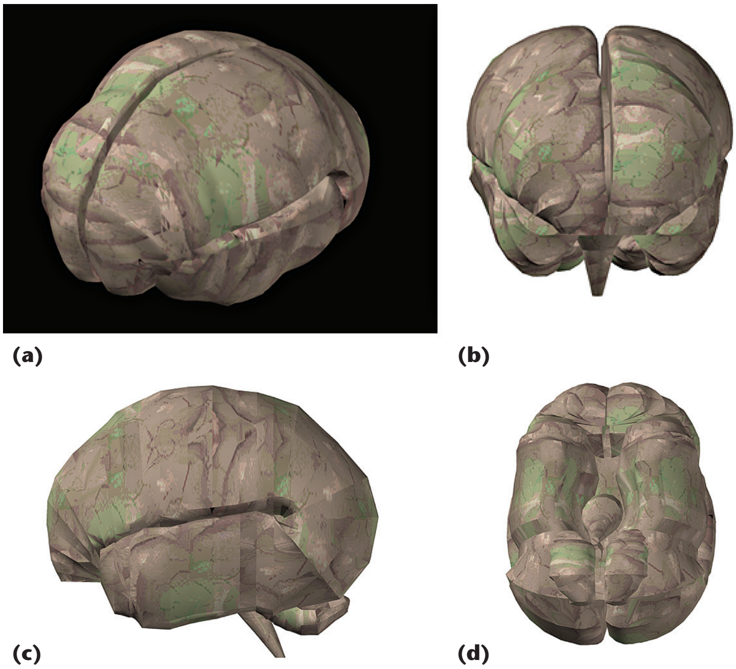

Figure 9. 3D views of the brain model we used in the simulation: (a) isometric, (b) posterior, (c) lateral, and (d) bottom. The images exclude the skull and cerebrospinal fluid. The improved model resolution permits a good calculation using FEM.

observed behavior in injuries. Such an objective method could be used to sustain evidence, support hypotheses, and verify post mortems.

In addition, we've used our mathematical model to create high-quality images of the brain (see Figure 9). Such quality is important for improving the resolution of images of specific parts of the brain and improving the study of head injuries.

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