

Received:
19 August 2018
Revised:
31 October 2018
Accepted:
30 November 2018

Cite as: Tom Brooks,
Jung Eun Choi, Mark Garnich,
Niels Hammer,
John Neil Waddell,
Warwick Duncan,
Mark Jermy. Finite element
models and material data for
analysis of infant head
impacts.
Heliyon 4 (2018) e01010.
doi: [10.1016/j.heliyon.2018.e01010](https://doi.org/10.1016/j.heliyon.2018.e01010)



Review Article

Finite element models and material data for analysis of infant head impacts

Tom Brooks^a, Jung Eun Choi^b, Mark Garnich^a, Niels Hammer^{c,d,e},
John Neil Waddell^b, Warwick Duncan^b, Mark Jermy^{a,*}

^a Department of Mechanical Engineering, University of Canterbury, Private Bag 4800, Christchurch, 8140, New Zealand

^b Sir John Walsh Research Institute, Faculty of Dentistry, University of Otago, PO Box 56, Dunedin, 9054, New Zealand

^c Department of Anatomy, University of Otago, PO Box 56, Dunedin, 9054, New Zealand

^d Department of Orthopaedic and Trauma Surgery, University of Leipzig, Liebigstr. 20, 04103, Leipzig, Germany

^e Fraunhofer Institute for Machine Tools and Forming Technology, Medical Division, Nöthnitzer Str. 44, 01187, Dresden, Germany

* Corresponding author.

E-mail address: mark.jermy@canterbury.ac.nz (M. Jermy).

Abstract

Finite element (FE) models of the infant human head may be used to discriminate injury patterns resulting from accidents (e.g. falls) and from abusive head trauma (AHT). Existing FE models of infant head impacts are reviewed. Reliability of the material models is the major limitation currently. Infant head tissue properties differ from adults (notably in suture stiffness and strain-to-failure), change with age, and experimental data is scarce. The available data on scalp, cranial bone, dura, and brain are reviewed. Data is most scarce for living brain. All infant head model to date, except one, have used linear elastic models for all tissues except the brain (viscoelastic or Ogden hyperelastic), and do not capture the full complexity of tissue response, but the predicted whole-head response may be of acceptable accuracy. Recent work by Li, Sandler and Kleiven has used

hyperelastic models for scalp and dura, and an orthotropic model for bone. There is a need to simulate falls from greater than one metre, and blunt force impacts.

Keywords: Biomedical engineering, Pathology, Pediatrics, Anatomy, Biophysics, Computational biology, Physiology, Mechanics

1. Introduction

Head injury is the most common cause of death in infant and child homicides in the world (Butchart et al., 2006). In the United States, Abusive Head Trauma (AHT) resulted in the deaths of an estimated 2250 infants and children (less than five years old) during a 15 year period between 1999 and 2014 (Spies and Klevens, 2016). When a child who has suffered AHT is admitted to Accident and Emergency Departments (A&E), accidental falls are often cited as the cause of the injury (Chadwick et al., 1991). Johnson et al. (2005) reviewed cases of children (less than five years old) admitted to one A&E department over an eight month period with head injuries resulting from accidental falls and noticed that many of the injuries a child sustains from AHT can also result from domestic accidents. However without the history being corroborated by a second, reliable, witness, making a clear determination of the cause is difficult. Johnson et al. (2005) also found that children falling from less than half a metre are extremely unlikely to suffer a skull fracture; while for falls over one metre, 95% of children had visible head injuries, with this increasing to 100% over one and a half metres. Similar studies were also carried out by Chadwick et al. (1991), Wilkins (1997) and Williams (1991). Studies by Reece and Sege (2000) and Roach et al. (2014) found that children who suffered from AHT had higher rates of diffuse axial injury and subdural haemorrhage (SDH) than children who suffered head injuries from accidental causes. Castellani and Schmidt (2018) reviewed the literature on acceleration thresholds required for fatal brain trauma. They found that translational forces rarely cause the rupture of bridging veins, which may explain why SDH does not often occur in the head injuries obtained when a child experiences an accidental fall. Rather, it is the magnitude of a rotational force that is more likely to produce the inflicted head injury pattern. From the above studies, it is possible, with certain injury patterns and subject to certain assumptions, to make estimates of the probability that a child's injuries were a result of an accident or AHT; however, without reliable witnesses to the history of the injury, this probability alone is not enough to reliably determine the cause.

The human head is a complex structure composed of a series of well-organised layers (Fig. 1), including the skin (dermis) of the scalp, muscles and galea aponeurotica, the osseous skull, the three layers of meninges, and deeper the brain and cerebrospinal fluid (CSF). The skull forms the protective cavity for the brain and is made up of the eight different bones of the neurocranium that are fused together by the

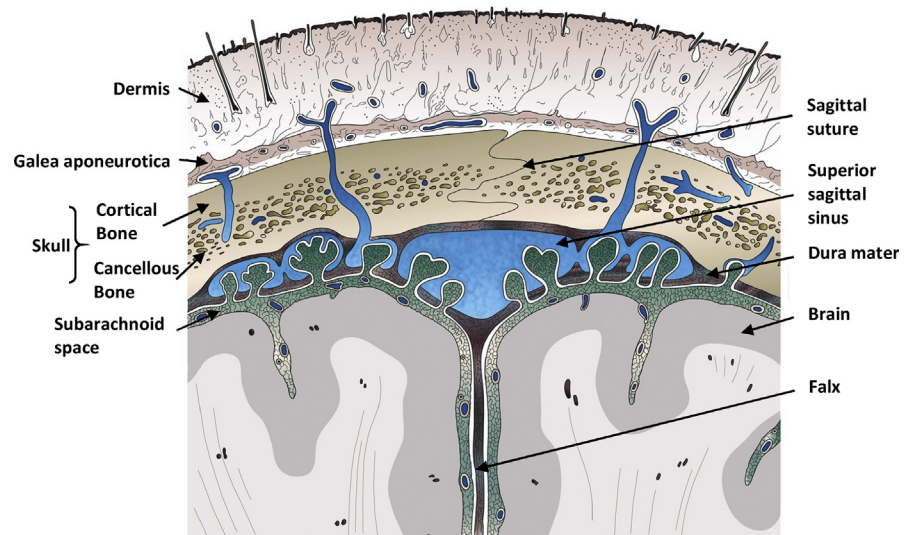


Fig. 1. Cross-section of the human head (modelled after [Waschke and Paulsen \(2013\)](#)).

syndesmotomic sutures. Each plate consists of cancellous bone sandwiched between two layers of cortical bone. The infant skull differs from that of an adult in that the cranial bones are thin, pliable plates that are separated by relatively more fibrous sutures. Areas of these more fibrous tissue sutures, called fontanelles, are present in the infant skull, between the cranial bones; these allow for both compression of the skull during birth and stretching of the skull as the brain grows. At approximately the age of two years, the sutures have largely fused together, with the largest fontanelle (frontal fontanelle) closing at around two and a half years old ([Ridgway and Weiner, 2004](#)). Inside the skull are the meninges, which consists of three layers that support and cover the brain and spinal cord externally ([Schmitt, 2014](#)). The outermost layer of the meninges is the dura mater, followed by the arachnoidea mater and pia mater (not shown in [Fig. 1](#)). The brain is surrounded by CSF, which cushions it against mechanical shock loading ([Schmitt, 2014](#)).

Finite Element (FE) modelling ([Bathe, 2006](#); [Mac Donald, 2011](#); [Reddy, 1993](#)) is a numerical method of reaching a solution to the set of equations which approximate the stress and deformation fields in a physical object, subject to simplifying assumptions about the geometry of the object and the mechanical behaviour of the materials which constitute it. The FE method involves the creation of a geometry that represents a problem of interest, discretising the geometry into subdomains ('elements'), applying boundary conditions and material constitutive laws, and then solving the resulting equations numerically. In the context of infant and child head injuries, an FE model can be used to better understand the injury patterns resulting from head impacts such as accidental falls or blunt force trauma. Compared to experiments, FE models offer control of loading cases, the ability to study many cases with the same effort required for one experiment, and to study cases that are

impossible to do experimentally for ethical reasons. Insights from FE models are expected to enable medical professionals to make more reliable inferences as to the cause of injury when a child is admitted to A&E.

Finite element models of the adult head have been developed, largely for studying the effects on the human head of motor vehicle accidents, ballistic impacts, blasts, pedestrian falls and sports related injury. As summarised by [Raul et al. \(2008\)](#) and [Tse et al. \(2014\)](#), the first two dimensional (2D) FE models were developed in the early 1970s, with three dimensional (3D) models being developed in the 1990s due to the increased availability of computational resources. The most detailed adult models include the Wayne State University Brain Injury Model ([Ruan et al., 1993](#); [Zhou et al., 1995](#); [Zhang et al., 2001](#)), the Louis Pasteur University - Finite Element Head Model ([Kang et al., 1997](#)), the Kungliga Tekniska Högskolan model ([Kleiven and Hardy, 2002](#)), the Simulated Injury Monitor Head Finite Element Model ([Takhounts et al., 2003, 2008](#)) and the University College Dublin Brain Trauma Model ([Horgan and Gilchrist 2003, 2004](#)).

There is a need for a validated FE model of the infant head that can be used to investigate injuries that may have been caused by accident or abuse, as well as in the design of products for impact safety (such as cars, car seats, flooring or cots). The present paper reviews the literature to date on infant head FE models. There is a particular focus on the suitability of the material models that have been previously used, as this is the greatest limiting factor on the accuracy and validity in current infant FE models.

2. Main text

2.1. Finite element models of the infant head

Biological systems are inherently difficult to model due to complex geometry, heterogeneity, nonlinear behaviours of the materials involved and often poorly defined boundary conditions. Generalised FE models of the infant head are far more challenging than those for the adult head due to the rapid development a child undergoes in the first few years of life, and the material properties of the infant's head tissues change as the infant grows. Exacerbated by the scarce availability of infant cadaver samples, determining the material properties of the tissues that constitute the infant head is the primary limiting factor in infant head FE models. To date, only a handful of published studies have been carried out that involve infant head FE models ([Margulies and Thibault, 2000](#); [Coats et al., 2007](#); [Roth et al., 2007, 2008, 2010](#); [Ponce and Ponce, 2011](#); [Li et al., 2013a,b, 2015a, 2017](#)). [Table 1](#) provides a brief summary of these. A valid FE model for the infant head needs the correct geometry, material properties and boundary conditions. This section will outline how these requirements have been tackled by previous infant FE models.

Table 1. Summary of FE models of the infant head.

Study	Age of Infant being Modelled	Anatomic Features Modelled	Purpose and Application	FE Code
Margulies and Thibault (2000)	1 Month	Skull, suture, brain	Impact loading to investigate the effect of cranial bone properties.	LS-Dyna
Coats et al. (2007)	5 Weeks	Skull, suture, brain	Study skull fracture from occipital impacts	ABAQUS/Explicit
Roth et al. (2007)	6 Months	Skull, fontanel, CSF, scalp, brain, bridging veins	Compare vigorous shaking and inflicted impact.	Not Stated
Roth et al. (2008)	6 Months	Scalp, skull, face, sutures, CSF, brain	Comparison of child head model from CT scans and scaled adult head. Impact against rigid wall.	Radioss
Roth et al. (2010)	17 Days	Scalp, skull, sutures, meninges, CSF, brain	FE model of newborn head, with validation against experimental data.	Radioss
Li et al. (2013b)	1.5 and 3 Months	Scalp, skull, sutures, dura mater, CSF, brain	Create statistical head geometry. Parametric study to determine material parameter sensitivity.	LS-Dyna
Li et al. (2013a)	6 Months	Scalp, skull, suture, dura mater, pia mater, CSF, brain	Investigate effects of different drop heights and impact surfaces.	LS-Dyna
Li et al. (2015a)	From (Li et al., 2013b)	From (Li et al., 2013b)	Reconstruct 50 cadaver drop tests using parametric infant FE head model.	LS-Dyna
Li et al. (2017)	New-born, 5 and 9 Months	Scalp, skull, suture, dura mater, CSF, brain	Addition of nonlinear material models. Head impact and compression simulation.	LS-Dyna

2.2. Geometry

Human head geometries are more complex than most geometries used in traditional engineering structures due to the large number of curved surfaces. As a result, it is difficult to create an accurate head geometry with Computer Aided Drawing (CAD) software. Instead, most studies begin from Computed Tomography (CT) or Magnetic Resonance Imaging (MRI) data. CT uses X-ray absorption measurements on many different lines of sight to create cross sectional images. Multiple cross-sectional images are stacked to construct a 3D image (Cierniak, 2011). MRI scanners use oscillating magnetic fields to stimulate radio frequency emissions from hydrogen nuclei. Gradients in a steady magnetic field allow the location of the emitting nuclei to be determined (Brown et al., 2014). While CT usually has a greater resolution, it involves a dose of ionising radiation, therefore, MRI is preferred if it yields satisfactory images. With the data from either CT or MRI, the shape of individual organs or structures of interest can be extracted (segmentation) and converted into 3D geometry formats that can be imported into FE software packages so that a mesh can be generated.

Most FE models involving infant head impacts use geometric data from a single individual of a given age (Coats et al., 2007; Roth et al., 2007, 2008, 2010; Li et al., 2013a,b, 2017). This ensures realistic representation of that specific individual but may not reliably account for variations in geometric features between individuals

and the anatomical differences of the infant skull as it develops. In contrast, [Li et al. \(2015b\)](#) created a statistical model for the zero to three month old infant skull, using CT scan data from 56 children. Geometric features of the infant skull, such as the skull size, shape and thickness, as well as suture width, were quantified using CT images and statistical analysis. From this, a model was created so that the geometry of an actual infant skull could be generated based on the age and head circumference of the infant. [Li et al. \(2013b\)](#) created a statistical model of an infant's cranium geometry using multiple CT scans and a combination of multivariate regression and principal component analyses. The geometry was then used to create a baseline FE model that could be morphed into FE models with geometries representing a new-born, one and a half month old and three month old infants. This removes the need to expose further infants to radiation from a CT scan; and, due to the ease of generation, multiple geometries can be modelled at any one time. [Li et al. \(2016\)](#) shortened the time taken to generate FE models by creating a method to extract anatomical landmarks from CT scans, with these points representing the morphology of the infant head. A mesh-morphing method was then created to automatically morph a baseline FE model into an FE model with the geometry extracted from the CT scans.

The anatomic features that were commonly modelled were the skull, sutures and brain; however, as the models became more advanced, the scalp, dura mater and CSF were added. These six features are the most important to model as together they make the greatest contribution to the global stiffness of the infant head, due to making up the bulk of the head. Features such as the eyes, neck, jaw, teeth, tongue and vertebrae are not modelled as they are generally not part of the load path in a head impact. [Table 1](#) includes a summary of the anatomic features modelled in previous studies.

2.3. Finite element meshes

When the geometry of a model is discretised, it is broken up into subdomains called elements. The complete set of elements make up the mesh. Each element consists of a series of nodes (the number of nodes depends on the element type), each node being a location at which adjacent elements are connected to each other. The nodes are points in the mesh where the degrees of freedom (DOFs) are defined. The DOFs are the possible directions that the node can move in, up to a maximum of six (three translational and three rotational). In an FE model, the nodes are also the points at which the displacements are calculated and the forces and moments are transferred between elements. The displacement field over the volume of the element is defined by an assumed interpolation function that is generally a polynomial. It is often termed a 'shape function' as it defines the possible deformed shapes of the element. The computed nodal displacements define the polynomial coefficients. More nodes

in an element enables higher order interpolation. However, first and second-order interpolation are most common because of their computational efficiency and straightforward application.

2.3.1. Elements

In an FE mesh, different elements are used depending on the structure being modelled. Common elements for a 3D model include shell elements and solid elements such as tetrahedral and hexahedral.

2.3.1.1. Shell elements

Shell elements are commonly used to model thin walled structures where simplifying assumptions can be made about the through thickness behaviour. A thickness is associated with them but they are generally represented by nodes located on the mid-thickness of the structure. Shell elements inherently simplify the kinematics of deformation in the thickness direction; a simplifying assumption that should be validated. In addition, the use of shell elements may limit the choices of material models that can be employed. Typical applications include the modelling of sheet-metal structures, pressure vessels and aircraft wings.

2.3.1.2. Solid elements

Sometimes simplifying assumptions about structure behaviour cannot be made and fully 3D representations are required. In these cases, solid (or volume) elements must be used where a 3D volume representation of the structure is created. This generally results in more nodes and therefore higher computational expense. However, very accurate representations of structure behaviour can be achieved.

Solid elements are generally tetrahedral or hexahedral in shape. Hexahedral elements are shaped like a brick and are generally favoured due to their higher computational efficiency, but they lose accuracy when they are significantly distorted from the ideal rectangular shape. Geometric complexity often results in highly distorted hexahedrons, therefore, the less computationally efficient tetrahedral (pyramidal) elements are often employed due to the relative ease of creating a mesh, especially on curved surfaces.

2.3.1.3. Meshes used in infant FE models

Table 2 summarises the meshes used in previous studies, with descriptions of the meshing method outlined in most of them. Using the correct element type for the specific application is an important step in any FE model as different elements

Table 2. Element types used in previous infant FE model studies.

Study	Element Type				Number of Elements
	Scalp	Skull	Suture	Brain	
Margulies and Thibault (2000)		4 node linear shell	4 node linear shell	8 node linear brick	12 772
Coats et al. (2007)	Hexagonal 8 node solid	Hexagonal 8 node shell	2D 4 node membrane	Tetrahedral 10 node	32881
Roth et al. (2007)	Brick	Shell	Shell	Brick	69 324 brick, 9187 shell
Roth et al. (2008)		As Above			
Roth et al. (2010)	Brick	Shell	Shell	Brick	30 000 solid, 3700 shell
Li et al. (2013b)		Mesh Morphing Model			38 916 solid, 7 680 shell
Li et al. (2013a)	Shell	Shell	Shell	Hexahedral	50 404
Li et al. (2015a)		Baseline model same as Li et al. (2013a) above			
Li et al. (2017)	Hexahedron				5.3 million for nine month old infant

model deformation differently. None of the studies provided justification for the choice of element type.

2.4. Implicit and explicit solvers

Depending on the type of FE model being created, an implicit solver or an explicit solver can be used. In implicit methods, the dependent variables are defined by coupled equations which must be solved simultaneously, typically with iterative or matrix methods. They have an advantage of being more numerically stable than explicit methods and so larger step sizes can be used. Implicit solvers are, for example, used in analyses involving static loading conditions. An explicit solver solves a system of equations each describing a variable in the later state in terms of the known current state. The dependent variables can be directly calculated using the independent variables. This has an advantage of being relatively quick to solve, however, is less stable and so small step sizes are required. Explicit solvers are used in FE models that simulate dynamic events (such as impacts or blasts). They can deal with highly nonlinear problems without the convergence problems that are common when using an implicit solver. All of the infant head impact studies reviewed used an explicit solver, with Table 1 summarising the specific code used.

2.5. Material models and properties

In an FE model, a ‘constitutive model’ is a mathematical model that relates strain to stress. The term ‘material model’ is synonymous and is used here. Material

models are selected from a set of alternatives which employ different assumptions about the behaviour of the material. More complex models faithfully reproduce the behaviour of complex materials, but generally require longer computations and more sophisticated testing to obtain the required material properties. For example, isotropic linear elastic models are commonly used in analyses involving metals and loading conditions that do not exceed the yield strength, and may adequately model biological materials at low strain. Viscoelastic and hyperelastic models reproduce some of the important behaviours of biological materials at high strains, or variable strain rates. High strain rates are common in injury-causing impacts.

2.5.1. Linear elastic models

A linear elastic model can be used to represent the behaviour of a material where the stress is proportional to strain, the strain is small, there is no dependence on the rate of loading, and the material will undergo no permanent deformation (that is, returns to its original shape). The model will represent the material up to its elastic limit, which, for traditional engineering materials, is the yield stress.

If the material is isotropic and linear elastic then only two material constants characterise the material behaviour. For example, Young's modulus and Poisson's ratio. If the material is anisotropic (i.e. the properties are different in different directions), then, depending on the assumed type of anisotropy, different numbers of elastic constants are required. For a transversely isotropic material model, there are five independent constants, while for an orthotropic model, there are nine independent constants. Other types of anisotropic material models are rarely used.

2.5.2. Viscoelastic models

A viscoelastic material adds a viscous, dissipative component to the elastic behaviour. Loading or strain rate dependent behaviour can be represented. On a stress-strain curve of a viscoelastic model, the loading and unloading curves are different, the hysteresis being due to energy dissipation by the viscous behaviour. The most common type of viscoelastic model calculates the shear modulus, G , as a function of time, t , by (Eq 1):

$$G(t) = G_{\infty} + (G_0 - G_{\infty})e^{-\beta t} \quad (1)$$

Where G_{∞} is the long term shear modulus, G_0 is the instantaneous shear modulus and β is the viscoelastic decay constant. This time dependence of material stiffness results in creep strain under constant stress, and stress relaxation under constant strain. Biological tissues typically exhibit noticeable viscoelastic behaviour.

2.5.3. Hyperelastic models

Most materials deviate from a linear stress-strain relationship at large strains. Hyperelastic material models reproduce this nonlinearity by deriving the stress-strain relationship from a strain energy function. They are inherently nonlinear because of the large strain regime they are intended for and different hyperelastic models are accurate over different strain ranges. Different models may also be selected depending on the experimental data that is available. For example, the Ogden model requires at least uniaxial and equibiaxial test data. For an Ogden hyperelastic model the strain energy function is defined as (Ogden, 1984) (Eq 2):

$$W = \sum_i^N \frac{\mu_i}{\alpha_i} (\lambda_1^{\alpha_i} + \lambda_2^{\alpha_i} + \lambda_3^{\alpha_i} - 3) + \frac{1}{d}(J - 1)^2 \quad (2)$$

Where λ_i are the principal stretches, μ and α are the Ogden constants, J is the determinant of the elastic deformation gradient, d is a material incompressibility factor determined from the bulk modulus and N is the order of the Ogden model used. For a Mooney-Rivlin hyperelastic model (Mooney, 1940; Rivlin, 1948), the strain energy function is expressed in terms of the principal invariants, I_i , and can be defined by an infinite series (Eq 3):

$$W = \sum_m \sum_n C_{mn} (I_1 - 3)^m (I_2 - 3)^n + \frac{1}{d}(J - 1)^2 \quad (3)$$

Where C_{mn} is a material constant defined by curve fitting stress-strain data from physical testing. The number of parameters in the expansion of the series depends on the accuracy required.

2.5.4. Linear versus nonlinear material models

Linear elastic material models are the most common models used in infant head impact studies to date. The required parameters are easier to measure than those required for more advanced nonlinear models. Linear models are computationally inexpensive and therefore require relatively shorter solver times. However, they cannot accurately represent the material behaviour at high strains, as well as account for the effect of nonlinear or time dependent material behaviour.

Nonlinear material models are better able to model material behaviour at high deformations and strains, as well as rate dependent behaviour (although rate dependence can be linear). This allows for accurately modelling of the material under dynamic loads where the rate at which the load is applied can be varied, resulting in different material behaviour. Nonlinear models are however more complex to fit to experimental data. For implicit solvers, nonlinear models are more computationally

expensive because an iterative solution is required for each increment in load. They have been used in only one published infant head FE model to date (Li et al., 2017).

2.6. Properties of tissues measured from human and animal samples

As an infant grows, the material properties of the tissues making up their head also change (Coats and Margulies, 2006; Margulies and Thibault, 2000). This, along with the lack of availability of infant cadaver samples, are the limiting factors in all previous infant head FE models as there is a severe scarcity of test data for infant tissue mechanical properties.

Margulies and Thibault (2000) investigated the age dependent changes in the material properties of the infant skull and sutures by carrying out three point bend tests on human and porcine infant cranial bones. The samples were frozen at $-4\text{ }^{\circ}\text{C}$ during storage and then defrosted to room temperature in a bath of saline before the tests were conducted. Once it was confirmed that the material properties of the porcine and infant cranial bones agreed, further three point bend and tensile tests on porcine samples were carried out to represent the infant skull. From these tests, and comparisons with previously published data on the adult human head, it was found that the elastic modulus, ultimate tensile stress and energy absorbed to failure increase with the age of the cranial bone; while the ultimate strain decreases. For the sutures, it was also found that the elastic modulus, ultimate stress and energy absorbed to failure increase with the age of the cranial bone; however, there are significant differences in the magnitudes of these mechanical properties between the cranial bone and sutures. For infant cranial bone and suture, the elastic modulus and stress at fracture (termed by Margulies and Thibault (2000) as rupture modulus) increases with loading rate; however, the energy absorbed to failure does not. Jaslow (1990) demonstrated experimentally that in mature skulls, the suture absorbs a much larger amount of energy to failure than cranial bone during an impact, thus showing that the suture plays a shock absorbing role in the skull. Margulies and Thibault (2000) found that this does not occur for the infant skull as infant suture absorbs less energy before failure than adult suture.

Coats and Margulies (2006) conducted dynamic three point bend and tension tests on infant cranial bone and suture, respectively, at impact velocities between 1.2 and 2.8 ms^{-1} . This allowed for the determination of the rate dependent material properties of infant cranial bone and suture for rates associated with low height falls. During storage, the samples were frozen and then thawed in mock CSF solution before testing. It was found that as the age of the infant increased, the elastic modulus and ultimate stress of cranial bone also increased. Similar to Margulies and Thibault (2000), it was also found that infant suture deforms by a significantly greater amount (30 times) than infant cranial bone before failure. This means that due to the

flexibility of the skull, brain injury can still occur from the forces of the impact even though there is no skull fracture. Based on the difference in material property data for the cranial bones making up the skull, the impact location can influence whether skull fracture occurs or not. It was also found that strain rate does not affect the modulus of elasticity or ultimate stress of infant cranial bone; although the effect of fibre orientation on the material properties at different strain rates needs further investigation. For the infant suture, the material properties were not affected by strain rate or age which contradicts Margulies and Thibault (2000), but it is suggested that this may be due to Coats and Margulies (2006) testing a bone-suture-bone segment rather than just a suture segment like Margulies and Thibault (2000). The reported failure locations of the bone-suture-bone samples were at the bone-suture boundary, with no visible damage to the suture, therefore suggesting that the reported suture data for the ultimate stress and strain is that of the suture when the boundary fails.

Davis et al. (2012) carried out four point bend tests on 47 specimens of cranial bone from one six year old human head to investigate the effects of loading rates and the structure of cranium bone on the elastic modulus and bending stiffness. They found that the elastic modulus varies between the suture, cortical bone and the sandwich bone of cortical-cancellous-cortical (1.10 GPa, 9.87 GPa and 3.69 GPa respectively), with the loading rate having no effect. The bending stiffness of the sandwich bone was found to be much greater than that of the cortical bone and suture ($12.32 \text{ Nm}^2\text{m}^{-1}$, $5.58 \text{ Nm}^2\text{m}^{-1}$ and $3.7 \text{ Nm}^2\text{m}^{-1}$ respectively), where the bending stiffness was defined as EI where E is the elastic modulus and I is the second moment of area. Due to a difference in the widths of each specimen (resulting from the harvesting procedure), the bending stiffness was normalised by the width (giving the reported units of Nm^2m^{-1}). The variation in the elastic modulus and bending stiffness for the two types of cranial bone needs to be accounted for in future FE models.

Prange et al. (2004) conducted compression experiments on the heads of one, three and 11 day old infant cadavers that were unembalmed and fresh-frozen. Impact testing was also carried out by dropping the specimens from heights of 0.15 and 0.30 m. It was found that the infant head is a lot more compliant than an adult head as, compared to previous adult test data, there were longer pulse durations and lower peak accelerations for the infant. For the infant drop tests, it was found that the impact response did not depend on the location of impact. From the compression tests, it was found that the compression direction did not affect the stiffness, but rather, the stiffness was affected by the velocity of the compression. The infant head was modelled as a simple mass-spring system and a three parameter viscoelastic model. Data from the compression test and one of the drop tests was used to determine the unknown parameters of each model. To validate the models, predictions of the force-time response for the second drop test were made and compared to the measured data. The spring-mass model did not accurately represent the measured data, however, the three parameter model performed better with a 9.5%

average absolute error for the peak head acceleration and 36% error for the predicted pulse duration. The main limitation to the experiment was that the cadaver samples were also used for other experiments and so could not be loaded to failure. Therefore, the peak accelerations recorded were below that of the fracture tolerance of the skull. Both the mass-spring system model and three parameter viscoelastic model considered the head as a whole, not individual tissues. This does not allow for an accurate representation of the deformation of the individual tissues and the contribution they make to the global head stiffness.

Wang et al. (2014) conducted three point bend tests to determine the mechanical properties of cranial bone and sutures from one to two year old infants. Samples were obtained from seven human infant cadavers, with eight samples from each cadaver from the frontal and parietal bones, and the sagittal and coronal sutures. The mechanical properties of interest were the elastic modulus, ultimate stress and ultimate strain. It was found that for the frontal bone, the elastic modulus and ultimate stress was higher than those of the parietal bone. With the sutures, there was no difference in the properties of interest between the two locations. The ultimate stress and elastic modulus in the cranial bones were higher than in the sutures, while the opposite was found for the ultimate strain.

2.7. Material models used in infant FE models to date

Due to the lack of material property data for the infant head, it has been common for isotropic, linear elastic material models to be used for the scalp, skull, sutures, dura and CSF, and a viscoelastic model for the brain. Table 3 summarises the material models used in previous infant FE models.

Li et al. (2017) are currently the only researchers who have used nonlinear material models in their infant FE modelling. They included the scalp, skull, sutures, dura mater, CSF and brain in their models. The scalp, sutures and dura mater contain mostly collagen and therefore exhibit nonlinear elastic behaviour, hence more

Table 3. Material models used for the infant head FE models.

Study	Tissue					
	Scalp	Skull	Suture	Dura	CSF	Brain
Margulies and Thibault (2000)		Linear Elastic	Linear Elastic			Linear Viscoelastic
Coats et al. (2007)	Linear Elastic	Orthotropic Elastic	Linear Elastic			Ogden Hyperelastic
Roth et al. (2007), (2008), (2010)	Linear Elastic	Linear Elastic	Linear Elastic		Linear Elastic	Linear Viscoelastic
Li et al. (2013a,b), (2015), (2016)	Linear Elastic	Linear Elastic	Linear Elastic	Linear Elastic	Linear Elastic	Linear Viscoelastic
Li et al. (2017)	Ogden Hyperelastic	Orthotropic Elastic -Age Dependent	Ogden Hyperelastic	Mooney-Rivlin Hyperelastic		Ogden Hyperelastic

advanced material models are needed. For the sutures, a first order Ogden hyperelastic model was created by fitting it to the stress-strain data of infant suture published by [Coats and Margulies \(2006\)](#). The scalp was modelled as two layers, an adipose tissue layer and the connective tissue layer. These layers were both modelled using a first order Ogden hyperelastic model with parameters adjusted from those of adult scalp presented in [Fahlstedt et al. \(2015\)](#). The adipose tissue layer Ogden parameters were assumed to be the same for infants as adults due to the lack of paediatric data, while the Ogden parameters for the connective tissue layer were assumed to be one tenth of those for adults due to the softer scalp in infants. For the dura mater, a Mooney-Rivlin hyperelastic model was used, with the parameters determined from data presented in [Bylski et al. \(1986\)](#). The skull was modelled using an orthotropic material model. The three point bend data of [Coats and Margulies \(2006\)](#) was used to determine the elastic modulus in the direction perpendicular to the direction of the fibres in the skull, where the fibre direction was determined by observation. Using an anisotropy ratio obtained from the data of [Kriewall \(1982\)](#) and the data from [Coats and Margulies \(2006\)](#), the elastic modulus in the direction parallel to the fibre direction was calculated. The elastic modulus in the through-thickness direction was assumed to be the same as that in the direction perpendicular to the fibres.

2.8. Material model parameters used in infant FE models to date

The model parameters currently existing in the literature are summarised in [Table 4](#). Many of the studies used values for the material properties of the skull from [Coats and Margulies \(2006\)](#), sutures from [Margulies and Thibault \(2000\)](#), CSF from [Willinger et al. \(1995\)](#) and brain from [Thibault and Margulies \(1998\)](#). [Roth et al. \(2007, 2008, 2010\)](#), [Li et al. \(2013a,b, 2015a\)](#) used adult values for the material properties of the CSF and the latter three studies also used adult values for scalp.

2.9. Loading cases modelled and predicted injuries

2.9.1. Comparing adult and infant material properties and geometries

[Margulies and Thibault \(2000\)](#) constructed FE models of a one month old infant head using the paediatric skull material properties for one simulation and the adult skull properties for another. Each model was subjected to impact loading based on accelerations measured in a study of shaken baby syndrome published by [Duhaime et al. \(1987\)](#). This consisted of half sinusoidal load-time input with peak magnitudes of 1000 N and 5000 N (to represent minor and major impacts respectively) and a pulse duration of 10 ms. The location of the impacts were in the parietal region, 45° from the vertical axis. From these models, it was found that the maximum intrusion of the impactor was 100% greater in the infant cranial bone compared to the adult cranial bone. The resulting strains on the brain, using the

Table 4. Material models parameters used in previous infant FE models.

Tissue	Linear Models			Study	
	Elastic Modulus (MPa)	Poisson's Ratio	Density (kg/m ³)		
Scalp	16.7	0.42	1200	Zhou et al. (1997) (Adult)	
Skull	500	0.22	2150	Coats and Margulies (2006)	
Suture	8	0.22	2150	Coats and Margulies (2006)	
Membranes	31.5	0.45	1140	Zhou et al. (1997) (Adult)	
CSF	0.012	0.49	1040	Willinger et al. (1995) (Adult)	
Nonlinear Models					
	Viscoelastic		Ogden Hyperelastic		
Brain	$G_0 = 5.99e-3$ MPa, $G_\infty = 2.32e-3$ MPa, $\beta = 0.09248$ s ⁻¹		$\mu_1 = 53.8$ Pa, $\alpha_1 = 10.1$, $\mu_2 = -120.4$ Pa, $\alpha_2 = -12.9$		Thibault and Margulies (1998), Li et al. (2017)
Suture			$\mu_1 = 1.48 \times 10^4$ Pa, $\alpha_1 = 6.9$		Li et al. (2017)
Scalp			Outer: $\mu_1 = 1.30 \times 10^4$ Pa, $\alpha_1 = 24.2$ Inner: $\mu_1 = 3.99 \times 10^3$ Pa, $\alpha_1 = 8.8$		Li et al. (2017)
Skull	(Orthotropic Elastic, Age Dependent) Five month old infant: $E_1 = 731.7$ MPa, $E_2 = E_3 = 266.2$ MPa, $G_{23} = 111.8$ MPa, $G_{12} = G_{31} = 194.8$ MPa				Li et al. (2017)
Mooney-Rivlin Parameters					
Dura			$C_1 = 1.18$ MPa, $C_2 = 0.295$ MPa		Li et al. (2017)

infant cranial material properties, caused diffuse, bilateral hemispheric distribution of maximum principal strains. Therefore, using this simplified model, it has been shown that impact loading may produce diffuse injury in infants.

Roth et al. (2008) carried out numerical simulations of a six month old infant head model using geometry derived from CT scans (real geometry) and comparing the results with models that used geometry based on scaling an adult head. Their model used an impact velocity of 1 ms⁻¹, against a rigid wall. Between the two geometries, the stress distributions in the skull and brain were very different, both for the magnitude and location of the maximum stress. For a frontal impact, the peak pressure in the occiput region was 19 kPa and 38 kPa for the real geometry and scaled adult geometry respectively. The von Mises stress in the brain, in the occiput region, was 0.8 kPa and 2 kPa respectively, while the maximum von Mises stress in the skull was 4.4 MPa and 3.7 MPa respectively. These differences show that a scaled adult head geometry is not suitable for infant head FE modelling. The major difference between

the two models was the thickness of the skull. For the scaled adult model, the skull thickness did not correspond to that of the CT scan from the six month old child. This means that in the scaled adult model, there was a smaller skull deflection and hence smaller stress in the brain. Overall, they found that scaling down the adult head does not appear relevant for use in child numerical simulations.

2.9.2. Validation studies

Roth et al. (2010) developed an FE model of a new-born head and validated it against previously published experimental data. The head was compressed by simulating a plate moving at 50 mms^{-1} contacting the head at the desired location, with the reverse side of the head placed against a rigid wall. Overall, good correlations were found for the force-displacement curves of the simulated impact and the experimental results. They also conducted a parametric study on the parameters of the brain viscoelastic model to determine the influence of the brain tissue on the skull deformation. It was found that large variations in the viscoelastic parameters of the brain material model led to very small changes in the skull stresses, skull deformation and peak acceleration of the head. Only when the bulk modulus was significantly increased from 2.11 GPa to 21.10 GPa, was there more than a 15% difference in results. Therefore, they concluded that brain material properties have minimal influence on the stresses and skull deformation, which conflicts with the conclusions of Coats et al. (2007) (outlined in the next subsection). To demonstrate the capability of their model, they also simulated a real world accidental fall where a one month old infant fell from one metre onto a concrete surface. Comparisons of the resulting skull fracture lines showed good accordance with those observed in the medical images in the patient's medical file. However, they noted that validation of skull fractures cannot be performed as the cadaver tests do not investigate them.

Li et al. (2013b) created a statistical model of an infant's cranium geometry using multiple CT scans and a combination of multivariate regression and principal component analyses. This model was then used to create geometries of a new-born, one and a half month old and three month old infants to be used in FE models. The FE models were used to carry out a parametric study where the sensitivity of various material parameters were quantified, under near-vertex impact loading conditions. Boundary conditions included an initial velocity calculated based on a drop height of 0.3 m and a frictional contact (with a coefficient of 0.2) for contact between the head and rigid surface. The elastic modulus of the skull, suture, dura mater and scalp, along with the long and short term shear moduli and decay constant for the brain were the parameters of interest in the parametric study. The maximum principal stress and strain in the skull and suture, as well as the peak head acceleration, were used to evaluate the changes in the above parameters. From the parametric

study, it was found that changes in the skull elastic modulus resulted in significant differences in each of the resulting parameters. An increase in the elastic modulus resulted in all output parameters increasing except for the maximum principal strain of the skull, which decreased. Model validation and material model parameter optimization was carried out by simulating the drop test experiments carried out by Prange et al. (2004). A parametric study found that the viscoelastic material properties of the brain had little effect on the result parameters, so were excluded from the optimisation study. Values for the elastic moduli of the skull, suture, scalp and dura mater were optimised to fit the data from the experimental tests by Prange et al. (2004). These optimised material model parameters were then used to create simulations replicating the tests from Prange et al. (2004) to validate the models, as discussed in the next section.

Li et al. (2013a) created an FE model of a six month old infant to simulate compression and drop tests that have been experimentally carried out by Loyd (2011). For the compression simulations, the head was compressed between two plates from the anterior-posterior direction and left-right direction, with velocities of 15 mms^{-1} and 45 mms^{-1} . A frictional contact boundary (with a friction coefficient of 0.2) was prescribed for contact between the head and plate. Comparison of the force-time plots for the simulations with those from the experimental data showed that the compression forces were slightly higher in the simulation. Therefore, the global head stiffness was a little stiffer than in reality. For the drop test simulations, an initial velocity was applied to the head based on drop heights of 0.15 and 0.30 m. The impact occurred between the head and an aluminium plate and had the same contact conditions as the compression tests. Five different impact orientations were simulated (forehead, occipital, vertex and parietal). There was an acceptable correlation between the simulation and experimental acceleration-time curves, with peak resultant accelerations around 80g for all head orientations from the 0.3 m drop height. These were around 10–20% higher than those recorded by Loyd (2011). This again shows that the FE model is globally slightly stiffer than the cadaver head. A parametric study of the elastic modulus for each tissue was also conducted. For each tissue the elastic modulus was decreased by half of the original value, as well as increased by a factor of two. It was found that the elastic modulus of the skull and scalp had the most significant effect on the peak acceleration, and the von Mises stress and maximum principal strain in the skull. For example, the increase in the modulus of the skull resulted in an increase of the peak acceleration from approximately 75 g to around 85–90 g, and the von Mises stress increased from around 30 MPa to 50 MPa. Simulations were also carried out to determine the effects of drop height and the stiffness of the impact surface on the head responses. Impact surface types included concrete, wood fibre board and hard, soft and rigid foam. Drop heights of 0.15, 0.30, 0.60, 0.90 and 1.20 m were used, with the impact location being in the occipital region. Overall, the head peak acceleration, maximum von Mises

stress and first principal strain of the skull all increased with increasing drop height and surface stiffness.

2.9.3. Modelling falls, inflicted impacts and shaking

Coats et al. (2007) developed a geometrically accurate FE model to predict skull fractures in infants resulting from impacts. An initial velocity that represented a fall from 0.3 m was applied to the head, and the impacting surface modelled as a fixed, rigid plate. They found that small variations in the thickness or width of the suture did not affect the principal stress in the infant cranial bone. However, large sutures (>10 mm) decreased the estimated occurrence of fracture resulting from an impact, thus showing that there is a relatively significant injury risk due to unusual anatomic variations. Through a parametric study of the parameters for the visco-hyper-elastic material model used for the brain (based on the work of Prange and Margulies (2002)), they found that decreasing the shear modulus of the infant brain by half does not affect the principal stress; but decreasing the stiffness by greater than one order of magnitude will significantly increase the principal stress in the cranial bone. They also found that changing the Poisson's ratio (from 0.499 to 0.49 and 0.4999) changes the bulk modulus and significantly varied the principal stress by 30–77%, thus showing the importance of determining the compressibility of the brain in a numerical model. Overall, their model was able to predict skull fracture resulting from an impact with a hard surface to good agreement with previous studies using infant cadavers.

Roth et al. (2007) created an FE model of a six month old infant head to compare vigorous shaking and an inflicted impact. The shaking was modelled by taking one angular velocity cycle from the data recorded by Prange et al. (2003) and applying it to the centre of rotation of the system. For the inflicted impact, the head was modelled as hitting a rigid wall at 3 ms^{-1} . The von Mises stresses and pressure were significantly higher in the inflicted impact model (14 kPa occurring in the occipital region, and 80 kPa respectively) than the shaking model (3.2 kPa occurring in the vertex region, and 22 kPa respectively). Both the shaking and inflicted injury models experienced similar relative displacements of the brain in the sagittal plane, leading to the rupture of bridging veins. The strain in the bridging veins was determined to be 100% and 90% for the impact and shaking simulations respectively. The strain calculation was based on the original lengths of the bridging veins and the relative displacement in the sagittal plane (for the elongation of the bridging veins). The similarity in the bridging vein strains shows that shaking can cause subdural haemorrhaging, even though there may not be the physical trauma (such as skull fractures) present from a possible inflicted impact.

Ponce and Ponce (2011) used FE models to simulate the effects of impacts to an infant's head to predict, locate and quantify diffuse brain injuries. The original FE

model was to simulate the vibrations of an infant's head when they are shaken so that its effect on the first through to fourth cervical vertebrae could be investigated. A 400 N load was applied to the occipital region of the head, while the head was free to rotate about the spinal cord in the axis of impact. The resulting stress in the brain was greater than the acceptable limit for areas far from each other, thus indicating potential damage to the neurological tissue.

Li et al. (2015a) used a parametric infant FE model based on Li et al. (2013b) to predict paediatric skull fractures. By using the model to reconstruct previous infant cadaver tests, skull fracture risk curves were able to be generated for children less than nine months old. Using mesh morphing techniques, head geometries were created from data of age, head size/shape and skull thickness that was reported in the cadaver tests. Overall, it was found that the stress responses in the skull were better at predicting skull fracture rather than kinematic based measures, such as peak head acceleration.

Jiang et al. (2017) created a simplified computational model of the infant head to simulate the skull response to blunt impacts. Only the skull and sutures were included in the geometry and a linear elastic material model was used. Initial simulations were carried out for an impact test replicating experimental tests on piglet skulls. It is claimed that this simulation showed a good match for the skull fracture patterns between the simplified model and piglet heads. However, there is very little in regards to describing the boundary conditions used in this model. As commented by Johnson and Auer (2018), this FE model is overly simplified and cannot be considered a qualitatively or quantitatively validated model.

2.9.4. Li, Sandler and Kleiven's model: nonlinear material models

As discussed in the material models and properties section earlier, Li et al. (2017) used nonlinear material models for the scalp, sutures, dura mater and skull. They modelled drop and compression tests for infant heads as carried out experimentally by Loyd (2011) so that the material models could be validated against cadaver tests. FE models of a new-born, five and nine month old infant were created, with the same material properties used for each age. For the drop tests, an initial velocity was applied to the model head that simulated a drop height of 0.3 m, and five different head impact orientations were used (forehead, occipital, vertex, left and right parietal). The acceleration-time curves from the drop tests correlated well with the experimental data across all impact locations and ages. For the compression tests, the head was simulated to be compressed between a fixed plate and a plate moving at a velocity calculated to obtain a strain rate of 0.3 mm (mm⁻¹s⁻¹). The force-deflection curves all showed an increase in the stiffness at large displacements, also seen in the experimental tests. To further validate the nonlinear material models,

the same FE simulations were carried out for the frontal and parietal impacts, with linear elastic material models for the scalp, sutures and dura mater. For the frontal impacts, the linear elastic suture model resulted in an increase of 22.4% in the peak acceleration, while the linear elastic scalp model resulted in an increase of 49.2% when compared to the use of their respective nonlinear models. A 22.4% decrease in the von Mises stress in the skull occurred when using the linear elastic scalp model due to the stiffening effect of the linear elastic model (there was less bending in the skull). The linear elastic model for the dura resulted in an increase of 20% for the peak acceleration and little change in the von Mises stress. Absolute values for the von Mises stress in the skull were not presented. Overall these linear elastic models resulted in an overall stiffer model. This is due to the linear elastic models not allowing for the uncrimping in soft tissues, which causes a period of lower stiffness.

2.10. Discussion

To date, only five complete, distinct FE models have been previously created for the infant head. These include [Margulies and Thibault \(2000\)](#), [Coats et al. \(2007\)](#) and [Li et al. \(2017\)](#), as well as the models that have been used in multiple studies by [Roth et al. \(2007, 2008, 2010\)](#) and [Li et al. \(2013a,b\)](#), [Li et al. \(2015a, 2016\)](#). These models include the relevant anatomy and boundary conditions subject to simplifying assumptions which appear to be consistent with the goal of modelling impact injury at a computational demand which can be achieved on at least high-end desktop computers.

Two further models should be noted. The model of [Jiang et al. \(2017\)](#) considered only the skull and sutures, which is overly simplified given that the whole infant head, with all of its tissues, is of interest. As a result of their limitations, these two studies are not considered as being as important as the other studies focusing on the modelling of the infant head. The model by [Ponce and Ponce \(2011\)](#) focused on modelling the vibration from shaking in an infant's cervical vertebrae, but also modelled the effects of an impact to the infant head, with the head rotating about the spinal cord. The kinematic effects of the head and neck are important if the primary force (impact or shaking) is applied to the torso, rather than the head. These features come at the cost of added complexity and computational expense.

The above five complete models consist of suitable geometry, mesh, material models, boundary conditions, solution method and validation to simulate head impacts in infants. A reasonable amount of research has already been completed to achieve the state of current infant head FE models to date. However, there is room to further improve various aspects of the models so that they can be used as another tool to investigate head injuries that may have been caused by accident or abuse.

Any FE analysis requires inputs of geometry, discretisation method (meshing), material models, boundary conditions and solver, along with suitable validation (MacDonald, 2011). Some points regarding these inputs from the current FE models are discussed here to identify areas for further improvement.

2.10.1. Geometry

In regards to geometry, Li et al. (2016, 2015b) have developed methods to create geometries of infant heads quickly, efficiently and accurately based on statistical models and CT scans. If required, specific geometry of a particular infant's head can be obtained from CT scans, and software used to convert the data into 3D geometry formats. Obtaining geometries of an infant head is relatively straight forward if scan data is available and therefore is not an important limiting factor to creating an accurate and valid infant head FE model.

2.10.2. Meshing

For meshing, there is room to further investigate the use of suitable element types to model the various tissues. In the studies listed in Table 2, there were descriptions of the discretisation methods used to generate the meshes used in their respective models. However, there was no justification for the types of elements used to discretise each tissue.

2.10.3. Material models

The material models currently available for the various tissues of the infant head are the most important factors limiting the accuracy of FE models to date. Given the scarcity of experimental data on the mechanical properties of infant tissues, material models are generally determined from animal tissues (such as pigs). While infant porcine tissues such as cranial bone have been shown to be similar to human infants (Margulies and Thibault, 2000), there can be no substitute for directly determining material properties from human infant tissues. To date, the model created by Li et al. (2017) using nonlinear material models is the most valid given that its results show a better correlation against cadaver test results, compared to those for a similar model using linear elastic materials by Li et al. (2013a) that was validated against the same cadaver tests. All other studies used linear elastic material models, which resulted in FE models that were not as well validated against experimental data compared to the model by Li et al. (2017). As Li et al. (2017)'s peak accelerations were within 10% of those measured in the cadaver drop tests (Li et al. (2013a) had a difference of 10–20%), there is clearly an increase in the accuracy when using nonlinear models.

In the studies using linear elastic material models, only the skull and suture models contain parameters measured from infant tissue samples (along with those for the

viscoelastic model for the brain). As noted in [Table 4](#), the parameters used for the scalp, membranes and CSF are those for the adult tissues. This is due to the lack of specific data for the infant tissues and the scarcity of human infant tissue samples to conduct the required experiments on. Adult skull has an elastic modulus of around 8 GPa ([Hubbard, 1971](#)), while 500 MPa has been previously used for the infant skull. Adult suture has been shown to have similar properties to the adult skull and therefore also has an elastic modulus of around 8 GPa, while 8 MPa has been used for the infant suture ([Hubbard et al., 1971](#)). These significant differences in the elastic moduli for the infant and adult tissues show the need for specific data relating to the infant tissues. Hence the use of adult data for the scalp, membranes and CSF are a limitation to the respective FE models.

2.10.4. Boundary conditions: falls, inflicted impacts and shaking

Many of the FE models to date use boundary conditions replicating low height falls and compression of the head ([Roth et al., 2010](#), [Li et al., 2013a,b, 2017](#)). Low height falls are often given as the history of the cause of an infant's head injuries ([Chadwick et al., 1991](#)), therefore modelling these types of head impacts is of value. However, only [Roth et al. \(2007\)](#) modelled an impact velocity higher than that experienced in a fall from one metre. To estimate the maximum impact speeds of blunt weapons in deliberate assaults, [Williams \(2008\)](#) found that the maximum speed of a fit adult male swinging a baseball bat (the longest weapon commonly employed in assault) was 36.2 ms^{-1} . Impacts with similar characteristics to that of a baseball bat swung at 36 ms^{-1} should be investigated to understand the injury patterns which may result from deliberate physical abuse. The parameters which determine the acceleration experienced, peak force and energy transferred in an impact are the velocities, masses and stiffness of the two (or more) colliding objects.

In the five studies noted above, the boundary conditions were typically applied to the models by specifying an initial velocity calculated from the simulated drop height and the mass of the head. The impacting surface was generally a fixed, rigid plate and where specified, had a frictional boundary condition. In most of the studies reviewed, little detail was given in terms of how the boundary conditions were implemented. These details are important for an FE model as the assumptions used in the model require different amounts of modelling effort and computational cost. In order to use an FE model to investigate the differences in the nature of different head impacts, it will be important to correctly model the boundary conditions.

2.10.5. Validation of the complete head models

Validation of an FE model is important to determine the overall accuracy of the model in simulating real world problems. [Li et al. \(2013a,b, 2017\)](#) and [Roth et al.](#)

(2010) all validated their FE models against experimental cadaver tests. Li et al. (2013b) and Roth et al. (2010) validated their models against the experimental results from (Prange et al., 2004). For the compression tests, both Roth et al. (2010), and Li et al. (2013b) described their models as having a good correlation with the experimental results for the force-displacement curves. However, Prange et al. (2004) describes the curve as being an exponential function, but the curves from the FE models do not accurately reflect this. Therefore, there is room for improving the accuracy and validity of these FE models. For the drop tests, Li et al. (2013b) had a range of both small and large variations in the results for peak head acceleration when compared to the cadaver drop tests. The smallest difference was around 2 g and the largest being approximately 20 g. Roth et al. (2010) conducted a statistical analysis based on the standard deviations provided by Prange et al. (2004) and found that the error between their results and the cadaver tests was 51% and 45% for the occipital and parietal impacts respectively. As with the compression tests, there is also room to improve the FE models to better predict the peak acceleration responses.

Li et al. (2013a, 2017) validated their FE models against the cadaver tests from Loyd (2011). For the compression tests, Li et al. (2013a) found that their model predicted slightly higher forces on the force-displacement plot, while the model created by Li et al. (2017) produced a curve that followed the general exponential shape of the experimental curve. For the drop tests, Li et al. (2013a) found that the peak acceleration was 10–20% higher than the cadaver tests, while Li et al. (2017) found a difference of, at most, 10%. As Li et al. (2017) used nonlinear material models in their FE model, their results indicate the importance of accurately modelling the materials to improve the accuracy and validity of the infant head FE models. It should also be noted that Li et al. (2013a) used a model for a six month old infant compared to the five month old cadavers used by Loyd (2011), which could explain the higher stiffness of their model, given that the stiffness of the infant skull increases with age (Margulies and Thibault, 2000; Coats and Margulies, 2006).

The model by Li et al. (2017) is the only model that is validated to within 10% of experimental tests. Validation occurred for the combined FE model consisting of all tissues, however, the individual models for each of the tissues were not validated in isolation. There is scope to validate the individual tissue material models to assess whether the model parameters are suitable. This could involve a range of physical testing of an individual tissue and using the resulting data (such as displacements or strains) to validate an FE model replicating the physical testing. Other FE models are largely limited by the material models used as they are often limited to linear elastic models, which do not include rate dependent behaviour of the biological tissues involved. The use of linear elastic models can also result in FE models that have a relatively larger stiffness as they do not allow for uncrimping in soft tissues. As soft tissues are placed under load, the individual collagen fibres

start to elongate and align with the load direction. This creates a toe-region on the stress-strain curve where the stiffness is lower than when the fibres are uncrimped (Meyers et al., 2008).

For the studies that were validated against cadaver tests, global parameters, such as acceleration for the drop tests and force-deflection curves for the compression tests, were compared. There are an infinite number of combinations of local parameters that could satisfy these global parameters. That is, highly inaccurate local parameters could lead to accurate global results. This means that local stresses and strains are not necessarily modelled with the same level of accuracy. Therefore, the validation of individual material models will reduce the likelihood of a combination of inaccurate local parameters resulting in accurate global parameters.

2.11. Recommendations for future work

Overall, the greatest limitation in the current infant head FE models is the modelling of the tissues' material behaviour. There are many factors influencing the behaviour of the tissues under a given load, including the loading rate, age dependence and the orientation of fibrous tissues. Accounting for each of these factors is the challenge for future models.

2.11.1. Tissue tests

In determining the influence of loading rate and age dependence on the behaviour of the tissues, experimental testing of infant tissues is required. This will allow the determination of the parameters for the material models of infant tissues. However, the greatest limitation to this is the availability of suitable specimens for physical testing. Obtaining specimens of infant head tissues is often difficult due to their limited availability and for ethical reasons. This results in limited experiments that can be conducted to determine different material model parameters, and any available specimens are likely to be used for other studies deemed to be more significant. Often, animal tissue substitutes (such as infant porcine tissues) are used for such physical testing as they have similar properties to that of infant tissues. The use of simulant materials could also be used, but these require their validity to be proven, so do not eliminate the need for animal or human tissue tests.

When tissue specimens are available, it is critical to ensure that they are preserved in a state that allows them to exhibit the same mechanical properties as living tissues. Due to technical and logistical reasons, human tissues cannot usually be tested mechanically on the site of harvest and in sufficient sample sizes. In order to preserve the tissues and prevent autolysis or bacterial contamination, they are usually treated chemically, using ethanol or formaldehyde, or frozen. Tissues used in previous

research have been frozen and then defrosted to room temperature in saline baths or mock CSF solution (Margulies and Thibault, 2000; Coats and Margulies, 2006).

Chemical fixation is known to impair tissue mechanics of both soft and hard tissues, in particular of the organic matrix. In a series of studies, Hammer et al. (2014) showed that both ethanol and formaldehyde cause changes in the properties of human bone and that the mechanisms of denaturation vary for the chemicals. The effects of freezing of biological tissues on their mechanical properties have given controversial results, of which the freezing protocol appears to play a major role. A key aspect in the limited validity of results obtained from tissues that have been frozen appears to be the formation of ice needles in the tissues. If a relatively slower cooling rate is used, the ice needles tend to increase in size, resulting in the tissues being intrinsically destroyed by their formation, with subsequent loss of water. Vice versa, it appears that a rigorous pre-cooling of the tissues close to the freezing point prior to snap freezing, and a rapid warming of the frozen tissues prior to testing, may minimise such alterations of tissue integrity. In determining the behaviour of the infant head tissues, these preservation effects need to be considered in order to be able to accurately model the material behaviour in an FE model.

2.11.2. Age dependence

In terms of the effect of age dependence, only Li et al. (2017) use a model for the skull that is a function of age, with most other FE models using parameters that have been obtained from specimens of a similar age to that of which they are trying to model. Given that the material properties of the infant head tissues change with age (Coats and Margulies, 2006; Margulies and Thibault, 2000) (and are significantly different to those for adults), material models need to be a function of the age of the infant. An ideal infant head FE model would be one that can be parameterised in terms of age so that model components such as the geometry and material models can be adapted for the desired age. Therefore, there is a future need for the material models to be age dependent. However, this is likely to require a large amount of work and resources as each tissue, at a range of ages, would have to undergo physical testing and a model created based on the results.

2.11.3. Anisotropy

Infant tissues such as the skull are anisotropic. Current models use geometry that does not necessarily incorporate any direction dependence for the tissues. While orthotropic material models have been used to account for direction specific material properties (Li et al., 2017), direction dependence is not accounted for in the geometry. For example, the infant skull has fibres orientated radially outwards from the ossification centres. This means that a material model involving radial coordinates, with a defined centre (the ossification centre) is more suitable. Therefore, being able

to incorporate this into the geometry of the skull is likely to be complex and involve another avenue of research.

2.11.4. Impact with different objects

Another area of interest is the simulation of blunt force impacts from weapons. As discussed earlier, previous FE models simulated falls onto flat surfaces. Roth et al. (2007, 2008) simulated an inflicted impact, but the object contacted was a rigid wall, which is mechanically identical to a fall with the same impact velocity. There have been no previous studies simulating blunt force impacts from non-flat weapons. These types of impacts are important to model in future research so that there is a greater understanding of the differences in the injury patterns resulting from impacts such as falls from less than a metre, falls greater than a metre and from potential weapons. Impacts from objects with curved or angular surfaces are likely to exhibit injury patterns that differ from those with large flat surfaces (such as a floor). Therefore, understanding the differences will allow medical professionals to make more informed observations on whether an infant is likely to have suffered accidental or non-accidental head injuries.

3. Conclusions

A valid infant head FE model requires accurate geometric representation of the head, a suitable discretisation method, material models that model the effect of loading rate, age dependence and directional dependence, boundary conditions that accurately model the different loading conditions, and validation against experimental tests. There are only five complete FE models in the current literature that are close to meeting these requirements.

- For the geometry, a lot of work has already been carried out for obtaining both specific and generalised geometry of the infant head. Further improvements can be made by accounting for the directional dependence of some tissues in the geometry.
- A variety of finite element types have been used to create the meshes in these models, however, there has been little reported in terms of justification for the type used. Future models can be improved by investigating suitable types of elements for the different tissues.
- Li et al. (2017)'s model is the only model to date that uses material models that can model the effect of loading rate, age dependence and to some degree, directional dependence. The other models are limited to linear elastic and isotropic for tissues such as the scalp, skull and dura, and linear viscoelastic for the brain, which do not model these effects. Improvements can be made to the material models by carrying out further experimental tests on human infant tissues. However, infant tissues suitable for the required tests are scarce.

- The boundary conditions need to be modelled correctly in order to be able to use the FE model to distinguish between different injury patterns resulting from different impacts. The current FE models lack some details around the specific boundary conditions used. Most of the models are used principally to model low height falls. There is value in using an FE model to simulate falls from greater heights than one metre and blunt force impacts from weapons to understand the different injury patterns.
- Validation of the FE models to date largely consists of comparing acceleration and force-deflection plots to cadaver drop and compression tests. This allows for validation of the models at a global level, but does not necessarily mean that the local parameters are accurate. Validation of individual material models for each of the tissues can be further improved to increase the global accuracy of the FE model, as well as ensuring they can model the effects of loading rate, age and direction dependence.

Overall, the most significant deficiency in infant head FE models is the lack of validated combinations of material models and element types for the different tissues that make up the head, as these are age dependent up to maturity. As the gap is filled by further research, FE modelling of the head offers the possibility of accurate, reliable prediction of the injury site and severity from defined impacts. FE modelling of infant head impacts is likely to find a role in the investigation of suspicious injuries, and in the design of safety equipment.

Declarations

Author contribution statement

All authors listed have significantly contributed to the development and the writing of this article.

Funding statement

This research did not receive any specific grant from funding agencies in the public, commercial, or not-for-profit sectors.

Competing interest statement

The authors declare no conflict of interest.

Additional information

No additional information is available for this paper.

Acknowledgements

The authors would like to thank Robbie McPhee for providing the illustrative drawing of Fig. 1.

References

- Bathe, Klaus-Jürgen, 2006. Finite Element Procedures. Klaus-Jurgen Bathe.
- Brown, Robert W., Cheng, Yu-Chung N., Haacke, E. Mark, Thompson, Michael R., Venkatesan, Ramesh, Wiley, InterScience, 2014. Magnetic Resonance Imaging: Physical Principles and Sequence Design, second ed. John Wiley & Sons, Inc, Hoboken, New Jersey.
- Butchart, A., Kahane, T., Phinney Harvey, A., Mian, M., Furniss, T., 2006. Preventing Child Maltreatment: a Guide to Taking Action and Generating Evidence. WHO and International Society for the Prevention of Child Abuse and Neglect, Geneva.
- Bylski, Donita I., Kriewall, Timothy J., Akkas, Nuri, Melvin, John W., 1986. Mechanical behavior of fetal dura mater under large deformation biaxial tension. *J. Biomech.* 19 (1), 19–26.
- Castellani, Rudy J., Schmidt, Carl J., 2018. Brain injury biomechanics and abusive head trauma. *J. Forensic Med. Sci. Law* 4 (2), 91.
- Chadwick, D.L., Chin, S., Salerno, C., Landsverk, J., Kitchen, L., 1991. Deaths from falls in children - how far is fatal. *J. Trauma Inj. Infect. Crit. Care* 31 (10), 1353–1355.
- Cierniak, Robert, 2011. X-ray Computed Tomography in Biomedical Engineering. Springer, London.
- Coats, B., Margulies, S.S., Ji, S., 2007. Parametric study of head impact in the infant. *Stapp Car Crash J.* 51, 1–15.
- Coats, Brittany, Margulies, Susan S., 2006. Material properties of human infant skull and suture at high rates. *J. Neurotrauma* 23 (8), 1222–1232.
- Davis, Matthew T., Andre, M. Loyd, Shen, Han-yu Henry, Mulroy, Maura H., Nightingale, Roger W., Myers, Barry S., Dale Bass, Cameron, 2012. The mechanical and morphological properties of 6 year-old cranial bone. *J. Biomech.* 45 (15), 2493–2498.
- Duhaime, A.C., Gennarelli, T.A., Thibault, L.E., Bruce, D.A., Margulies, S.S., Wiser, R., 1987. The shaken baby syndrome. A clinical, pathological, and biomechanical study. *J. Neurosurg.* 66 (3), 409–415.

- Fahlstedt, Madelen, Depreitere, Bart, Peter, Halldin, Jos Vander Sloten, Kleiven, Svein, 2015. Correlation between injury pattern and finite element analysis in biomechanical reconstructions of traumatic brain injuries. *J. Biomech.* 48 (7), 1331–1335.
- Hammer, Niels, Voigt, Christian, Werner, Michael, Hoffmann, Falk, Bente, Klaus, Kunze, Holger, Scholz, Roger, Steinke, Hanno, 2014. Ethanol and formaldehyde fixation irreversibly alter bones' organic matrix. *J. Mech. Behav. Biomed. Mater.* 29, 252–258.
- Horgan, T.J., Gilchrist, M.D., 2004. Influence of FE model variability in predicting brain motion and intracranial pressure changes in head impact simulations. *Int. J. Crashworthiness* 9 (4), 401–418.
- Horgan, T.J., Gilchrist, Michael D., 2003. The creation of three-dimensional finite element models for simulating head impact biomechanics. *Int. J. Crashworthiness* 8 (4), 353–366.
- Hubbard, Robert P., 1971. Flexure of layered cranial bone. *J. Biomech.* 4 (4), 251–263.
- Hubbard, Robert P., Melvin, John W., Barodawala, Iqbal T., 1971. Flexure of cranial sutures. *J. Biomech.* 4 (6), 491–496.
- Jaslow, C.R., 1990. Mechanical properties of cranial sutures. *J. Biomech.* 23 (4), 313–321.
- Jiang, Binhui, Zhu, Feng, Cao, Libo, Presley, Barbara R., Shen, Ming, King, H., Yang, 2017. Computational study of fracture characteristics in infant skulls using a simplified finite element model. *J. Forensic Sci.* 62 (1), 39–49.
- Johnson, K., Fischer, T., Chapman, S., Wilson, B., 2005. Accidental head injuries in children under 5 years of age. *Clin. Radiol.* 60 (4), 464–468.
- Johnson, L., Auer, Roland, 2018. Commentary on: Jiang B, Zhu F, Cao L, Presley BR, Shen MS, Yang KH. Computational study of fracture characteristics in infant skulls using a simplified finite element model. *J Forensic Sci* 2017; 62 (1): 39–49. *J. Forensic Sci.* 63 (1), 345–348.
- Kang, Ho-Sung, Willinger, Rémy, M Diaw, Baye, Chinn, Bryan, 1997. Validation of a 3D Anatomic Human Head Model and Replication of Head Impact in Motorcycle Accident by Finite Element Modeling. SAE Technical Paper.
- Kleiven, Svein, Hardy, Warren N., 2002. Correlation of an FE model of the human head with local brain motion: consequences for injury prediction. *Stapp Car Crash J.* 46, 123–144.

- Kriewall, Timothy J., 1982. Structural, mechanical, and material properties of fetal cranial bone. *Am. J. Obstet. Gynecol.* 143 (6), 707–714.
- Li, Xiaogai, Sandler, Håkan, Kleiven, Svein, 2017. The importance of nonlinear tissue modelling in finite element simulations of infant head impacts. *Biomech. Model. Mechanobiol.* 16, 823–840.
- Li, Z.G., Luo, X., Zhang, J.H., 2013a. Development/global validation of a 6-month-old pediatric head finite element model and application in investigation of drop-induced infant head injury. *Comput. Methods Progr. Biomed.* 112 (3), 309–319.
- Li, Z., Liu, W., Zhang, J., Hu, J., 2015a. Prediction of skull fracture risk for children 0-9 months old through validated parametric finite element model and cadaver test reconstruction. *Int. J. Leg. Med.* 129 (5), 1055–1066.
- Li, Zhigang, Han, Xiaoqiang, Ge, Hao, Ma, Chunsheng, 2016. A semi-automatic method of generating subject-specific pediatric head finite element models for impact dynamic responses to head injury. *J. Mech. Behav. Biomed. Mater.* 60, 557–567.
- Li, Zhigang, Hu, Jingwen, Reed, Matthew P., Rupp, Jonathan D., Hoff, Carrie N., Zhang, Jinhuan, Cheng, Bo, 2013b. Erratum to: development, validation, and application of a parametric pediatric head finite element model for impact simulations. *Ann. Biomed. Eng.* 41 (1), 215–220.
- Li, Zhigang, Park, Byoung-Keon, Liu, Weiguo, Zhang, Jinhuan, Reed, Matthew P., Rupp, Jonathan D., Hoff, Carrie N., Hu, Jingwen, 2015b. A statistical skull geometry model for children 0-3 years old. *PLoS One* 10 (5).
- Loyd, Andre Matthew, 2011. *Studies of the human head from neonate to adult: an inertial, geometrical and structural analysis with comparisons to the ATD head.* Citeseer.
- Mac Donald, Bryan J., 2011. *Practical Stress Analysis with Finite Elements*, second ed. Glasnevin Publishing, Dublin.
- Margulies, S.S., Thibault, K.L., 2000. Infant skull and suture properties: measurements and implications for mechanisms of pediatric brain injury. *J. Biomech. Eng.* 122 (4), 364–371.
- Meyers, Marc André, Chen, Po-Yu, Lin, Albert Yu-Min, Seki, Yasuaki, 2008. Biological materials: structure and mechanical properties. *Prog. Mater. Sci.* 53 (1), 1–206.
- Mooney, Melvin, 1940. A theory of large elastic deformation. *J. Appl. Phys.* 11 (9), 582–592.

- Ogden, Ray W., 1984. *Non-linear Elastic Deformations*. Elsevier.
- Ponce, E., Ponce, D., 2011. Modeling Neck and Brain Injuries in Infants, pp. 90–96.
- Prange, Michael T., Coats, Brittany, Duhaime, Ann-Christine, Margulies, Susan S., 2003. Anthropomorphic simulations of falls, shakes, and inflicted impacts in infants. *J. Neurosurg.* 99 (1), 143–150.
- Prange, Michael T., Luck, Jason F., Dibb, Alan, Chris, A. Van Ee, Nightingale, Roger W., Myers, Barry S., 2004. Mechanical properties and anthropometry of the human infant head. *Stapp Car Crash J.* 48, 279.
- Prange, Michael T., Margulies, Susan S., 2002. Regional, directional, and age-dependent properties of the brain undergoing large deformation. *J. Biomech. Eng.* 124 (2), 244–252.
- Raul, Jean-Sébastien, Deck, Caroline, Willinger, Rémy, Bertrand, Ludes, 2008. Finite-element models of the human head and their applications in forensic practice. *Int. J. Leg. Med.* 122 (5), 359–366.
- Reddy, Junuthula Narasimha, 1993. *An Introduction to the Finite Element Method*, Vol. 2. McGraw-Hill, New York.
- Reece, Robert M., Sege, Robert, 2000. Childhood head injuries: accidental or inflicted? *Arch. Pediatr. Adolesc. Med.* 154 (1), 11–15.
- Ridgway, Emily B., Weiner, Howard L., 2004. Skull deformities. *Pediatr. Clin.* 51 (2), 359–387.
- Rivlin, R.S., 1948. Large elastic deformations of isotropic materials IV. Further developments of the general theory. *Phil. Trans. Roy. Soc. Lond.* 241 (835), 379–397.
- Roach, Jonathan P., Acker, Shannon N., Bensard, Denis D., Sirotnak, Andrew P., Karrer, Frederick M., Partrick, David A., 2014. Head injury pattern in children can help differentiate accidental from non-accidental trauma. *Pediatr. Surg. Int.* 30 (11), 1103–1106.
- Roth, Sébastien, Raul, Jean-Sébastien, Bertrand, Ludes, Willinger, Rémy, 2007. Finite element analysis of impact and shaking inflicted to a child. *Int. J. Leg. Med.* 121 (3), 223–228.
- Roth, Sebastien, Raul, Jean-Sebastien, Willinger, Remy, 2010. Finite element modelling of paediatric head impact: global validation against experimental data. *Comput. Meth. Progr. Biomed.* 99 (1), 25–33.

- Roth, Sébastien, Raul, Jean-Sébastien, Willinger, Rémy, 2008. Biofidelic child head FE model to simulate real world trauma. *Comput. Meth. Progr. Biomed.* 90 (3), 262–274.
- Ruan, Jesse S., Khalil, Tawfik B., King, Albert I., 1993. Finite element modeling of direct head impact. SAE Technical Paper.
- Schmitt, Kai-Uwe, 2014. *Trauma Biomechanics: an Introduction to Injury Biomechanics*, 4th. Springer, Heidelberg. 2014.
- Spies, E.L., Klevens, J., 2016. Fatal abusive head trauma among children aged < 5 Years - United States, 1999-2014. *MMWR Morb. Mortal. Wkly. Rep.* 65 (20), 505–509.
- Takhounts, Erik G., Eppinger, Rolf H., Campbell, J Quinn, Tannous, Rabih E., 2003. On the development of the SIMon finite element head model. *Stapp Car Crash J.* 47, 107.
- Takhounts, Erik G., Ridella, Stephen A., Hasija, Vikas, Tannous, Rabih E., Campbell, J Quinn, Malone, Dan, Danelson, Kerry, Joel Stitzel, Rowson, Steve, Duma, Stefan, 2008. Investigation of traumatic brain injuries using the next generation of simulated injury monitor (SIMon) finite element head model. *Stapp Car Crash J.* 52, 1.
- Thibault, Kirk L., Margulies, Susan S., 1998. Age-dependent material properties of the porcine cerebrum: effect on pediatric inertial head injury criteria. *J. Biomech.* 31 (12), 1119–1126.
- Tse, Kwong Ming, Lim, Siak Piang, Tan, Vincent Beng Chye, Lee, Heow Pueh, 2014. A review of head injury and finite element head models. *Am. J. Eng. Technol. Soc* 1 (5), 28–52.
- Wang, J.W., Zou, D.H., Li, Z.D., Huang, P., Li, D.R., Shao, Y., Wang, H.J., Chen, Y.J., 2014. Mechanical properties of cranial bones and sutures in 1-2-year-old infants. *Med. Sci. Mon. Int. Med. J. Exp. Clin. Res.* 20, 1808–1813.
- Waschke, Jens, Paulsen, Friedrich, 2013. *Sobotta Atlas of Human Anatomy*, Vol. 3. Elsevier.
- Wilkins, Barry, 1997. Head injury—abuse or accident? *Arch. Dis. Child.* 76 (5), 393–397.
- Williams, E., 2008. *The Biomechanics of Blunt Force Trauma*. Masters Thesis in Forensic Science. University of Auckland.
- Williams, R.A., 1991. Injuries in infants and small children resulting from witnessed and corroborated free falls. *J. Trauma Acute Care Surg.* 31 (10), 1350–1352.

Willinger, R., Taleb, L., Kopp, C.M., 1995. Modal and temporal analysis of head mathematical models. *J. Neurotrauma* 12 (4), 743–754.

Zhang, Liying, Yang, King H., Ramesh, Dwarampudi, Omori, Kiyoshi, Li, Tieliang, Chang, Kun, Hardy, Warren N., Khalil, Tom B., King, Albert I., 2001. Recent advances in brain injury research: a new human head model development and validation. *Stapp Car Crash J.* 45 (11), 369–394.

Zhou, Chun, Khalil, Tawfik B., King, Albert I., 1995. A New Model Comparing Impact Responses of the Homogeneous and Inhomogeneous Human Brain. SAE Technical Paper.

Zhou, Chun, Khalil, Twafik B., King, Albert I., Dragovic, Ljubisa J., 1997. Head Injury Assessment of a Real World Crash by Finite Element Modeling. NASA (19980003886).