Probabilistic description of infant head kinematics in abusive head trauma


To link to this article: https://doi.org/10.1080/10255842.2017.1403593

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Probabilistic description of infant head kinematics in abusive head trauma


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ABSTRACT

Abusive head trauma (AHT) is a potentially fatal result of child abuse, but the mechanisms by which injury occur are often unclear. To investigate the contention that shaking alone can elicit the injuries observed, effective computational models are necessary. The aim of this study was to develop a probabilistic model describing infant head kinematics in AHT. A deterministic model incorporating an infant's mechanical properties, subjected to different shaking motions, was developed in OpenSim. A Monte Carlo analysis was used to simulate the range of infant kinematics produced as a result of varying both the mechanical properties and the type of shaking motions. By excluding physically unrealistic shaking motions, worst-case shaking scenarios were simulated and compared to existing injury criteria for a newborn, a 4.5 month-old, and a 12 month-old infant. In none of the three cases were head kinematics observed to exceed previously-estimated subdural haemorrhage injury thresholds. The results of this study provide no biomechanical evidence to demonstrate how shaking by a human alone can cause the injuries observed in AHT, suggesting either that additional factors, such as impact, are required, or that the current estimates of injury thresholds are incorrect.

Introduction

Abusive head trauma (AHT) is a leading cause of traumatic brain injury in children (Duhaime et al. 1998; Parks et al. 2012a, 2012b) and is typically characterised by a triad of symptoms that include subdural and retinal haemorrhage and acute brain dysfunction (Harding et al. 2003). In those cases where an adult confesses to having injured the child, the mechanism described is often that of violent shaking (Adamsbaum et al. 2010). However, there is debate regarding whether shaking alone is sufficient to elicit the triad of injuries observed (Geddes et al. 2001, 2003, 2004; Punt et al. 2004; Smith and Bell 2008; Squier 2008). In part, this debate is based on biomechanics (Duhaime et al. 1987; Prange et al. 2003). The present biomechanical study was designed to investigate whether a probabilistic model could demonstrate how shaking can cause the injuries characteristic of AHT.

It has been proposed that rotational components of head motion are primarily responsible for the injuries described in AHT (Holbourn 1943; Ommaya and Gennarelli 1974). Experimental studies using non-human primates described concussion and subdural haemorrhage (SDH) injury thresholds as a function of the peak angular acceleration and velocity of the head (Gennarelli et al. 1982; Ommaya and Hirsch 1971; Ommaya et al. 1968). Scaling these thresholds by the relative brain mass and complex shear moduli of the brain allowed paediatric injury thresholds to be estimated (Duhaime et al. 1987; Margulies and Thibault 1992; Thibault and Margulies 1998).

Several computational studies and mechanical surrogate experiments have been performed to assess whether infant head kinematics during shaking exceed the proposed injury thresholds (Bondy et al. 2014; Cory and Jones 2003; Duhaime et al. 1987; Prange et al. 2003; Wolfson et al. 2005). Although rigid-body computational models have been used to describe AHT (Wolfson et al. 2005), the impact of different shaking motions on head kinematics has not been thoroughly investigated. The present study employed a deterministic coupled rigid-body computational model, developed in OpenSim (Delp et al. 2007), where realistic shaking motions were used to simulate a human infant's head kinematics during shaking. The present study has built upon previous studies, in which the rigid-body computational modelling approach was validated using in vivo shaking experiments (Lintern et al. 2015).
Using this approach, the peak angular acceleration and velocity of an infant’s head were estimated in the present study and compared with published injury thresholds.

Previous studies have investigated head kinematics using a narrow set of models and shaking input motions, but there is doubt about whether worst-case shaking scenarios had been attained. To complement these existing results, the present study proposes the first probabilistic description of infant head kinematics during shaking. Probabilistic models have been used in a number of areas in biomechanics to account for variability and uncertainty in model parameters (Laz and Browne 2010). The use of probability distributions to describe the variation in infant mechanical properties and imposed shaking motions has enabled a range of infant head kinematics to be simulated. This approach allowed the probability of exceeding existing injury thresholds, in worst-case shaking scenarios, to be evaluated. Worst-case shaking scenarios were also compared between infants of different ages by adapting the probability distributions to their respective age ranges. The methodology used in this study enables head kinematics to be simulated during different shaking inputs, and provides a computational framework to complement subsequent investigations of the injury mechanisms in AHT.

Materials and methods

Infant deterministic model

A computational model of a 4.5 month-old human infant’s head and torso geometry was developed using anonymised computed tomography (CT) scans obtained from an existing clinical database at Starship Children’s Health, Auckland, New Zealand. The collection and use of images for this study was approved by the New Zealand Health and Disability Ethics Committees. The upper thoracic vertebrae (t1-t2), cervical vertebrae (c1-c7), skull, and torso were individually segmented and surface meshes were generated with the surface rendering tool in OSIRIX (Rosset et al. 2004). The relative positions and orientations of images for this study was approved by the New Zealand Health, Auckland, New Zealand. The collection and use of anonymised computed tomography (CT) scans obtained from an existing clinical database at Starship Children’s Health, Auckland, New Zealand. The collection and use of images for this study was approved by the New Zealand Health and Disability Ethics Committees. The upper thoracic vertebrae (t1-t2), cervical vertebrae (c1-c7), skull, and torso were individually segmented and surface meshes were generated with the surface rendering tool in OSIRIX (Rosset et al. 2004). The relative positions and orientations of each bone were used to build a computational model in OpenSim (Delp et al. 2007). A schematic of the model is included in Figure 1.

Each vertebral joint within the model was restricted to a single rotational degree of freedom (DOF) in the sagittal plane. A lumped-parameter description of joint torque was used to represent the passive force contributions of the muscles, ligaments, and other soft tissues for each vertebral joint within the neck. Exponential loading relationships have been demonstrated to be suitable for the paediatric human spine (Dibb et al. 2014; Luck et al. 2008, 2013; Luck 2012; Nuckley and Ching 2006; Nuckley et al. 2005; Ouyang et al. 2005). The torque (τ) within each joint was therefore defined to be exponentially related to the joint angle (Equations (1) and (2)), where $A_f$ and $A_e$ are gradient coefficients applied during flexion and extension, respectively, $k_f$ and $k_e$ are the coefficients that control the mechanical response during flexion and extension, respectively, and θ is the joint angle, defined to be positive during neck extension. The atlantoaxial-occipital (c0-c2) joint complex was modelled as a single functional unit (Nightingale et al. 1998, 2002) with loading properties described by Equation (1), while the remaining joints had loading properties that were described homogenously using Equation (2).

\[
\tau_{c0c2} = \begin{cases} 
-A_f (e^{-k_f \theta} - 1); & \theta \leq 0 \\
A_e (e^{k_e \theta} - 1); & \theta > 0 
\end{cases} 
\] (1)

\[
\tau = \begin{cases} 
-A_f (e^{-k_f \theta} - 1); & \theta \leq 0 \\
A_e (e^{k_e \theta} - 1); & \theta > 0 
\end{cases} 
\] (2)

Hunt–Crossley contact spheres (Hunt and Crossley 1975; Sherman et al. 2011) were included in the model to describe the contact forces that occur during impact of the occiput with the back, and the chin with the chest. The use of this contact mechanism to reproduce the accelerations during shaking was validated in using an in vivo lamb experimental model (Lintern et al. 2015). The position and radius of the contact spheres were approximated by inspection and fixed during the simulations. The contact stiffness and dissipation were described by the parameters $K_{cont}$ and $\alpha_{diss}$, respectively.
The head mass and moment of inertia (MOI) of the infant's head were described using a logarithmic age regression model (Loyd et al. 2010). A representative example of vertebral inertial properties was obtained using another logarithmic age regression model (Dibb 2011). The age-dependent estimates of the vertebral properties were scaled by an order of magnitude to account for the surrounding soft tissues in the neck using the relative anatomical dimensions of the vertebrae and the neck cross-section. Age-dependent estimates of the torso mass of the infant were prescribed to be the difference between the weight of a 50th percentile male (The World Health Organization 2006), and the age-dependent head mass estimates. A linear coefficient (c) describing viscous damping within the joints was based on *in vivo* shaking results that were obtained using a lamb experimental model (Lintern et al. 2015).

A computational model of an average male adult upper body (Holzbaur et al. 2005; Steele et al. 2013) was included with the infant model to prescribe kinematic (displacement) boundary constraints to the infant. The boundary constraints produced realistic shaking motions in the sagittal plane by varying the elevation of the shoulder joint (\(x_{\text{elev}}\)), flexion of the elbow (\(x_{\text{flex}}\)), and the deviation of the wrist (\(x_{\text{dev}}\)). All joints were perturbed sinusoidally about the configuration in Figure 1 (Equations (3)–(5)), where \(X_{\text{elev}}\), \(X_{\text{flex}}\) and \(X_{\text{dev}}\) denote the amplitudes of the sinusoids that describe angles of the shoulder, elbow and wrist joints, respectively. Other parameters used to describe the loading were the shaking frequency (\(f\)) and the phase difference (\(\Phi\)) between the sinusoids describing the shoulder elevation and elbow flexion. The elbow flexion and wrist deviation were assumed to be in phase (using the same value for \(\Phi\)), and a single constant frequency was prescribed to all three sinusoids.

\[
x_{\text{elev}} = X_{\text{elev}} \sin (2\pi f)
\]  
\[
x_{\text{flex}} = X_{\text{flex}} \sin (2\pi f + \Phi)
\]  
\[
x_{\text{dev}} = X_{\text{dev}} \sin (2\pi f + \Phi)
\]  

By varying \(\Phi\), shaking motions could be produced that occurred along an arc (\(\Phi = 0\) radians) or along an approximately linear path (\(\Phi = \pi\) radians). By describing the input motion as a function of these five shaking parameters, a full range of shaking motions, which are constrained to be within anatomical limits, can be produced. This approach enables worst case shaking scenarios to be investigated.

Appropriate shaking frequencies were taken from previous studies that involved manual shaking of a 1.5 month-old infant surrogate (Coats and Margulies 2008). The shaking frequency from five independent shaking experiments varied between 1.6 Hz and 2.5 Hz with an average of 2 Hz (personal communication with B. Coats, 2013).

The infant model was rigidly attached to the hands of the adult upper body model, and the hand motion was prescribed as a kinematic boundary constraint to the infant’s torso. A forward simulation was performed where the equations of motion were solved using the *Cpodes* integrator in OpenSim (Sherman et al. 2011). Peak angular acceleration and peak angular velocity of the infant’s head were recorded. An inverse dynamics analysis was performed to estimate the joint torques necessary to produce the shaking motion. Because of the kinematic constraint between the infant and the hands of the adult, the inverse dynamics analysis was simplified to consider only the inertial forces necessary to accelerate the infant and the shakers arms, and excluded the large impact forces that were propagated throughout the limb segments. The implications of these assumptions are addressed in the discussion.

**Infant probabilistic model**

To account for uncertainty in model parameters, probabilistic methods were used to analyse the infant head kinematics during shaking by combining the OpenSim API with NESSUS (SwRI, San Antonio, TX, USA). The OpenSim forward simulations were initiated by NESSUS by generating and providing a set of model parameters to an executable code developed in the OpenSim API. The model parameters were sampled from a set of probability density functions (PDFs) to obtain cumulative probability distributions (CDFs) describing the peak angular acceleration (\(a_{\text{peak}}\)) and peak angular velocity (\(\omega_{\text{peak}}\)) of the infant’s head.

PDFs were prescribed for all parameters including those describing mass properties, inertial properties, joint loading properties, contact properties, and the shaking boundary conditions (Table 1). The mean parameter values (\(\mu\)) were estimated from the literature, and the standard deviations (\(\sigma\)) were chosen using published experimental uncertainties, or were approximated arbitrarily to be 10% of their respective means. A lognormal distribution was used for the shaking frequency with the mean and standard deviation estimated from the average and the variance in the surrogate shaking experimental data-set (personal communication with B. Coats, 2013). This distribution allowed a bias towards higher frequencies to clearly demonstrate the effect of frequency on the head kinematics. The lognormal distribution was also chosen for the vertebral MOIs to allow the values to vary across
Table 1. Probabilistic model parameters with probability distribution functions, mean (μ) and standard deviation (σ) for a 4.5 month-old infant.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Description</th>
<th>μ</th>
<th>σ</th>
<th>Distribution</th>
</tr>
</thead>
<tbody>
<tr>
<td>ε</td>
<td>Viscous damping (N·m·rad⁻¹·s)</td>
<td>0.2</td>
<td>0.05</td>
<td>Normal</td>
</tr>
<tr>
<td>$A_f$</td>
<td>Flexion joint loading coefficient (N·m)</td>
<td>0.0032</td>
<td>0.00032</td>
<td>Normal</td>
</tr>
<tr>
<td>$k_f$</td>
<td>Flexion joint loading coefficient (rad⁻¹)</td>
<td>23.87</td>
<td>2.387</td>
<td>Normal</td>
</tr>
<tr>
<td>$A_e$</td>
<td>Extension joint loading coefficient (N·m)</td>
<td>0.0277</td>
<td>0.00277</td>
<td>Normal</td>
</tr>
<tr>
<td>$k_e$</td>
<td>Extension joint loading coefficient (rad⁻¹)</td>
<td>12.73</td>
<td>1.273</td>
<td>Normal</td>
</tr>
<tr>
<td>$A_{fe}^{c0-c2}$</td>
<td>Flexion c0–c2 joint loading coefficient (N·m)</td>
<td>0.0036</td>
<td>0.00036</td>
<td>Normal</td>
</tr>
<tr>
<td>$k_{fe}^{c0-c2}$</td>
<td>Extension c0–c2 joint loading coefficient (rad⁻¹)</td>
<td>6.37</td>
<td>0.637</td>
<td>Normal</td>
</tr>
<tr>
<td>$k_{ke}^{c0-c2}$</td>
<td>Extension c0–c2 joint loading coefficient (N·m)</td>
<td>0.0041</td>
<td>0.00041</td>
<td>Normal</td>
</tr>
<tr>
<td>$\kappa$</td>
<td>Contact stiffness (MPa)</td>
<td>13</td>
<td>13</td>
<td>Lognormal</td>
</tr>
<tr>
<td>$\alpha_{diss}$</td>
<td>Contact dissipation (s·m⁻¹)</td>
<td>0.5</td>
<td>0.1</td>
<td>Normal</td>
</tr>
<tr>
<td>$m_{head}$</td>
<td>Head mass (kg)</td>
<td>2.29</td>
<td>0.18</td>
<td>Normal</td>
</tr>
<tr>
<td>$I_{head}$</td>
<td>Head MOI - sagittal plane (kg·m²)</td>
<td>0.0068</td>
<td>0.0014</td>
<td>Lognormal</td>
</tr>
<tr>
<td>$m_{vert-vert}$</td>
<td>c0–c2 vertebral mass (kg)</td>
<td>0.06</td>
<td>0.006</td>
<td>Normal</td>
</tr>
<tr>
<td>$I_{vert}$</td>
<td>General vertebral mass (kg)</td>
<td>0.03</td>
<td>0.003</td>
<td>Normal</td>
</tr>
<tr>
<td>$f$</td>
<td>Shaking frequency (Hz)</td>
<td>2</td>
<td>0.5</td>
<td>Lognormal</td>
</tr>
<tr>
<td>$m_{torso}$</td>
<td>Mass of infant’s torso (kg)</td>
<td>5.2</td>
<td>5.2</td>
<td>Deterministic</td>
</tr>
<tr>
<td>$\chi_{dev}$</td>
<td>Shoulder rotation amplitude (rad)</td>
<td>0</td>
<td>π/6</td>
<td>Uniform</td>
</tr>
<tr>
<td>$\chi_{elb}$</td>
<td>Elbow rotation amplitude (rad)</td>
<td>0</td>
<td>π/4</td>
<td>Uniform</td>
</tr>
<tr>
<td>$\phi$</td>
<td>Shoulder-to-elbow phase difference (rad)</td>
<td>0</td>
<td>5π/36</td>
<td>Uniform</td>
</tr>
</tbody>
</table>

*Estimated using lamb shaking experimental results (Lintern, 2015).
*Representative joint loading properties for a 5 month-old infant (Luck 2012).
*Estimated using elastic moduli ranges for the infant rib cage (Tsai et al. 2012).
*Estimated contact dissipation (Hunt and Crossley 1975).
*Head mass and MOI for a 4.5 month-old using an age-regression model (Loyd et al. 2010).
*Vertebral mass and MOI for a 4.5 month-old using an age-regression model (Dibb 2011).
*Anthropometric dummy shaking data (personal communication with B. Coats, 2013).
*Torso mass obtained from 50th percentile male child growth charts (The World Health Organisation 2006).

The Monte Carlo probabilistic method was applied in this study to ensure convergence of the results as the number of simulation trials was increased (Haldar and Mahadevan 2000). In this method, each parameter was randomly sampled according to its PDF, and statistics describing the output kinematics were evaluated. The results of these Monte Carlo simulations were compared with existing computational and experimental studies, and the probability of exceeding published injury thresholds was investigated. Sensitivity factors were used to describe how the variability in the model parameters impacted the variability in the output kinematic metrics. The sensitivity factors calculated using the Monte Carlo analysis were defined as the absolute values of the correlation coefficients between each model parameter and the output kinematics (Haldar and Mahadevan 2000; Saltelli et al. 2004). A correlation coefficient of 1 indicates that all variance in the output can be accounted for by the variance in the input parameter. Conversely, a correlation coefficient of 0 implies that the output is unrelated to the input parameter.

Additional Monte Carlo simulations were performed to investigate the range of head kinematics produced for infants of different ages. The mass, inertial, and joint loading properties were scaled to a newborn and a 12 month-old infant (Table 2). The mass and inertial parameters were scaled using the age-dependent models (Dibb 2011; Loyd et al. 2010; The World Health Organisation 2006), and the loading properties were estimated using representative experimental measurements (Luck 2012). The flexion loading parameters were not scaled for the 12 month-old infant as the experimental data was observed to be within an order of magnitude, and to ensure that the MOI was greater than zero. Uniform distributions were assumed for the shaking amplitudes and the phase difference as there was no experimental justification to use any other PDFs.

Table 2. Age-scaled probabilistic model parameters for a newborn and a 12 month-old infant. The model parameters not listed here were not scaled, and were set to the values for the 4.5 month-old infant (see Table 1).

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Newborn</th>
<th>12 month-old infant</th>
</tr>
</thead>
<tbody>
<tr>
<td>$k_f$ (rad⁻¹)</td>
<td>14.32</td>
<td>23.87</td>
</tr>
<tr>
<td>$k_e$ (rad⁻¹)</td>
<td>8.91</td>
<td>19.10</td>
</tr>
<tr>
<td>$k_{e-c0}$ (rad⁻¹)</td>
<td>0.25</td>
<td>6.37</td>
</tr>
<tr>
<td>$k_{e-c2}$ (rad⁻¹)</td>
<td>5.73</td>
<td>14.32</td>
</tr>
<tr>
<td>$m_{head}$ (kg)</td>
<td>5.25</td>
<td>10.5 × 10⁻⁶</td>
</tr>
<tr>
<td>$I_{head}$ (kg·m²)</td>
<td>0.00021</td>
<td>0.0006</td>
</tr>
<tr>
<td>$m_{vert}$ (kg)</td>
<td>10⁻⁶</td>
<td>10⁻⁶</td>
</tr>
<tr>
<td>$m_{vert-c0}$ (kg)</td>
<td>0.0006</td>
<td>0.12</td>
</tr>
<tr>
<td>$m_{vert-c2}$ (kg·m²)</td>
<td>5×10⁻⁶</td>
<td>15×10⁻⁶</td>
</tr>
</tbody>
</table>

*Estimated using lamb shaking experimental results (Lintern, 2015).
*Representative joint loading properties for a 5 month-old infant (Luck 2012).
*Estimated using elastic moduli ranges for the infant rib cage (Tsai et al. 2012).
*Estimated contact dissipation (Hunt and Crossley 1975).
*Head mass and MOI for a 4.5 month-old using an age-regression model (Loyd et al. 2010).
*Vertebral mass and MOI for a 4.5 month-old using an age-regression model (Dibb 2011).
*Anthropometric dummy shaking data (personal communication with B. Coats, 2013).
*Torso mass obtained from 50th percentile male child growth charts (The World Health Organisation 2006).
the joint loading variation prescribed for the 4.5-month-old infant. The torso masses \( m_{\text{torso}} \) were prescribed as deterministic model parameters in all simulations. The injury thresholds were also scaled to each age range from those described by Duhaime et al. (1987) and Thibault and Margulies (1998) using age-dependent brain mass estimates proposed by Dobbing and Sands (1973). A description of the injury threshold scaling is provided as supplementary material.

The peak torques estimated using inverse dynamics were compared to the maximum isometric joint torques that can be produced by a 50th percentile male. Estimates of the maximum joint torques were obtained from existing simulation results using the upper body computational model (Holzbaur et al. 2005; Steele et al. 2013) (Table 3).

Table 3. Maximum isometric joint torques that were used to constrain the simulated shaking motions.

<table>
<thead>
<tr>
<th>Maximum torque (N-m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shoulder flexion</td>
</tr>
<tr>
<td>Shoulder extension</td>
</tr>
<tr>
<td>Elbow flexion</td>
</tr>
<tr>
<td>Elbow extension</td>
</tr>
<tr>
<td>Wrist deviation (radial)</td>
</tr>
<tr>
<td>Wrist deviation (ulnar)</td>
</tr>
</tbody>
</table>

The maximum torque limit was used to identify and eliminate physically unrealistic shaking results of the Monte Carlo simulations, and to identify the worst-case-shaking scenarios.

**Results**

The results of 3000 Monte Carlo trials for the 4.5-month-old infant model were compared to existing experimental and computational results (Figure 2) (Bondy et al. 2014; Cory and Jones 2003; Duhaime et al. 1987; Wolfson et al. 2005). These simulations results are consistent with previous surrogate shaking experiments (Cory and Jones 2003) and a rigid-body computational study (Wolfson et al. 2005). The maximal head kinematics were observed to exceed scaled concussion and SDH injury thresholds. The probabilities of exceeding the angular accelerations and velocities necessary to elicit SDH were calculated to be 0.01 and 0.05, respectively (Figure 3).

The sensitivity factors demonstrated that the variability of the output kinematics were impacted most strongly by the variabilities in the shaking frequency and the contact stiffness (Figure 4). The sensitivities of the peak angular acceleration and velocity to \( f \) were similar (0.651 and 0.654, respectively), but only the peak acceleration demonstrated any sensitivity to the contact stiffness (0.32). The joint rotation amplitudes \( X_{\text{elev}}, X_{\text{flex}}, X_{\text{dev}} \) and the head inertia \( I_{\text{head}} \) also had a modest impact upon the output kinematics. These results are illustrated in Figure 5, and demonstrate that the output kinematics are highly sensitive to the applied shaking motion and the contact stiffness, but are relatively insensitive to the model parameters that describe the properties of the infant.

The maximal kinematics occurred when the shaking frequency exceeded 4.5 Hz, which significantly exceeded the experimental data described in a personal communication with B. Coats in 2013 (unreferenced). The simulations were compared with the peak isometric torque constraint to identify which shaking results were physically realistic. The constrained output kinematics for the three infant ages studied here (Figure 6(a)–(c)) demonstrated that the peak angular velocity varied inversely with age. The maximum peak angular velocity was limited to approximately 60, 50, and 45 rad s\(^{-1}\) for the newborn, 4.5 month-old, and 12 month-old infants, respectively. The highest accelerations were simulated using the newborn infant model (up to 35,000 rad s\(^{-2}\)), but the peak angular velocity was insufficient to exceed the scaled SDH injury thresholds. The results were similar for the older infants, with no simulated head kinematics exceeding SDH injury thresholds.
Figure 3. Cumulative distribution functions describing the statistics of the $\alpha_{\text{peak}}$ (a) and the $\omega_{\text{peak}}$ (b) obtained from the Monte Carlo analysis. The probabilities of producing kinematics that do not exceed the published injury thresholds are indicated by the corresponding solid (concussion) and dashed (SDH) lines. The probabilities of $\alpha_{\text{peak}}$ and $\omega_{\text{peak}}$ exceeding the SDH injury threshold were 0.01 and 0.05, respectively.

Figure 4. Sensitivity factors calculated from the Monte Carlo analysis. Much of the variance in the output kinematics was described by the shaking parameters and the contact stiffness ($K_{\text{cont}}$).

Figure 5. (a) Variation in the output kinematics when the shaking parameters (Shaking) and infant model parameters (Infant) were varied using individual Monte Carlo analyses. Results indicate a high sensitivity to the shaking parameters in the model, and a comparatively low sensitivity to parameters describing the mechanical properties of the infant. (b) Classifying the results for a 4.5 month-old infant according to contact stiffness ($K_{\text{cont}}$) indicates that this parameter has a strong influence on peak angular acceleration ($\alpha_{\text{peak}}$). (c) Classifying the results for a 4.5 month-old infant according to shaking frequency ($f$) indicates that this parameter has a strong influence on both $\alpha_{\text{peak}}$ and $\omega_{\text{peak}}$ (peak angular velocity).
frequency (Figure 4) is consistent with the infant head kinematics being proportional to the input acceleration. Furthermore, the peak angular acceleration was highly sensitive to the contact stiffness, which is likely due to the peak accelerations occurring during impact of the head with the torso. This head impact mechanism was observed during \textit{in vivo} shaking experiments (Lintern et al. 2015), and has been postulated to be a possible injury mechanism in AHT (Cory and Jones 2003). It is likely that this head-to-torso impact mechanism would be necessary to produce kinematics that exceed current estimates of injury thresholds in AHT.

This study predicted that the probabilities of producing peak angular accelerations and velocities that exceed SDH injury thresholds are 0.01 and 0.05, respectively. These estimates are limited by the input probability distributions and the lack of kinetic constraints imposed upon the prescribed shaking motions. Nevertheless, the results indicate that further research should focus on the shaking inputs, rather than the material properties of the infant, when investigating head kinematics during shaking. The likelihood of injury depends upon the injury metrics and thresholds used, and there is controversy regarding the validity of the primate-derived injury thresholds used in this study (Cory and Jones 2003; Pierce and Bertocci 2008). As \textit{in vivo} experimental investigations of brain injury thresholds in infants are unlikely, the field must turn to computational modelling studies to identify more appropriate biomechanical injury metrics. Such research will be complemented by the rigid-body shaking simulation framework developed in this study, which provides realistic worst-case boundary conditions for subsequent analysis.

\textbf{Discussion}

The probabilistic model used in this study has provided statistical descriptions of infant head kinematics during shaking. Worst-case shaking scenarios were investigated by excluding simulations that required non-physiological joint torques to be generated. The results were compared to estimated injury thresholds, derived from published data, for infants of three different ages.

This study demonstrated that the head kinematics is strongly influenced by the type of shaking motion imposed upon the infant. The peak angular acceleration was observed to be highly sensitive to the frequency of shaking and the contact stiffness controlling the impact between the head and torso, while the peak angular velocity was most sensitive to the shaking frequency. The results indicated relative insensitivity to variations in any of the model parameters that describe the mechanical properties of the infant. The range of kinematics that was produced is consistent with previous experimental and computational results (Cory and Jones 2003; Duhaime et al. 1987; Wolfson et al. 2005). Discrepancies observed between the simulation results and the finite element modelling results reported by Bondy et al. (2014) (Figure 2) can be attributed to a difference in the choice of material properties of the neck (in the present study, we have scaled these parameters to a newborn infant). Our simulation results for the newborn infant (Figure 6(a)) fall within the ranges previously described by Bondy et al. (2014), indicating that the parameter scaling used in our model is appropriate. The sensitivity factors predicted by the model are also realistic. The high sensitivity to the shaking frequency (Figure 4) is consistent with the infant head kinematics being proportional to the input acceleration. Furthermore, the peak angular acceleration was highly sensitive to the contact stiffness, which is likely due to the peak accelerations occurring during impact of the head with the torso. This head impact mechanism was observed during \textit{in vivo} shaking experiments (Lintern et al. 2015), and has been postulated to be a possible injury mechanism in AHT (Cory and Jones 2003). It is likely that this head-to-torso impact mechanism would be necessary to produce kinematics that exceed current estimates of injury thresholds in AHT.

This study predicted that the probabilities of producing peak angular accelerations and velocities that exceed SDH injury thresholds are 0.01 and 0.05, respectively. These estimates are limited by the input probability distributions and the lack of kinetic constraints imposed upon the prescribed shaking motions. Nevertheless, the results indicate that further research should focus on the shaking inputs, rather than the material properties of the infant, when investigating head kinematics during shaking. The likelihood of injury depends upon the injury metrics and thresholds used, and there is controversy regarding the validity of the primate-derived injury thresholds used in this study (Cory and Jones 2003; Pierce and Bertocci 2008). As \textit{in vivo} experimental investigations of brain injury thresholds in infants are unlikely, the field must turn to computational modelling studies to identify more appropriate biomechanical injury metrics. Such research will be complemented by the rigid-body shaking simulation framework developed in this study, which provides realistic worst-case boundary conditions for subsequent analysis.

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{figure6.png}
\caption{The results of three Monte Carlo analyses were performed with the model parameters selected to be representative of a newborn infant (a), a 4.5 month-old infant (b), and a 12 month-old infant (c). All simulation results that exceeded the peak isometric torque constraint were excluded. The peak isometric torque constraint imposed a limit on the peak angular velocity ($\omega_{\text{peak}}$) that was inversely proportional to the infant’s age. No results were observed to exceed the scaled SDH thresholds (dashed lines) for any age. Solid lines indicate scaled concussion injury thresholds, which were rarely exceeded.}
\end{figure}
The simulation results have demonstrated that an infant’s head kinematics scale with age (Figure 6), and that the worst-case shaking scenarios are dependent on the strength of the shaker. A large increase in peak angular acceleration was observed for the younger infant model, and this can be attributed to the logarithmic scaling of the mass and inertial properties. The smaller head mass resulted in smaller inertial forces, which caused the head to decelerate more rapidly during impact. The dependence of the peak angular acceleration on head inertia was also demonstrated by a relatively high sensitivity factor (Figure 4). The infant’s age, however, was observed to only have a minor effect on the peak angular velocity, and the maximal physically realistic head kinematics were more strongly influenced by the strength of the shaker. A stronger shaker can generate higher peak isometric joint torques, which results in shaking motions with larger peak angular velocities of the head. There is potential to use these scaling methods to adapt the modelling framework to describe worst-case shaking scenarios for individual shaking situations. The torque constraint could be scaled, as the peak isometric joint torque in the arms has been shown to scale linearly with muscle volume (Holzbaur et al. 2007a). Existing tools in OpenSim could be used to scale the length of the upper arms, and if combined with individualised muscle volumes, which could be derived from MRI (Holzbaur et al. 2007b), then customised joint torque constraints could be developed. It is worth noting that a recent clinical series reported that the severity of the injuries was worse when the shaker was male, and hypothesised that the size and strength of male shakers was one possible explanation (Esernio-Jenssen et al. 2011).

There are several limitations of the model that must be recognised. The model consisted of joints that were restricted to a single rotational DOF. In vivo shaking kinematics have been reproduced well using a similar computational model during sagittal plane shaking (Lintern et al. 2015) and, although the model does not reproduce any out-of-plane motion, the shaking inputs simulated in this study are consistent with those thought to occur in AHT (Cory and Jones 2003; Duhaime et al. 1987; Prange et al. 2003). The joints in the model were described using a representative anatomical data-set that was not scaled with an infant’s age throughout the simulations. It was assumed that the joint separation would have a negligible effect on the head kinematics compared to other inertial and contact forces. However, additional age-dependent information regarding the morphology of the head and neck may improve the applicability of the model when scaling the model across different age groups.

Another limitation was the use of a lumped-parameter description of joint torque, rather than modelling the passive and active force contributions of the paediatric cervical musculature. Although the passive muscle forces, and associated joint torques, were approximated by an exponential constitutive relationship, active muscle forces were not considered in this study. This was not considered a major limitation as the sensitivity results in this study demonstrated that the type of shaking input, and the inertia of the head had the largest influences on the results, whereas the joint stiffness had a comparatively minor effect.

The infant was rigidly attached to the shaker’s hands in the model, and the shaking motion was kinematically constrained. A limitation of this is approach that, the joints were occasionally forced through motions that the muscle forces could not support. An inverse dynamics analysis produced large spikes in the joint torques upon contact between the head and torso. In the physical situation, these torque spikes would be attenuated when the muscles can no longer produce the necessary force to continue the smooth sinusoidal motion. Estimating appropriate activation patterns for the shaking motions in a full musculoskeletal model (such as those proposed by Holzbaur et al. (2005)) would allow a joint torque boundary condition to be applied to the model, and would allow more realistic joint torques to be calculated. The inclusion of muscle models would also allow strength variations to be considered in the probabilistic analysis. By neglecting the contact forces in the inverse dynamics analysis, and by enforcing a smooth shaking motion, it is likely that the peak torque estimates were underestimated. The maximum isometric joint torque constraint provides an upper limit on motion due to the inverse relationship between contractile velocity and force production in skeletal muscle. Although skeletal muscle can generate a larger force during an eccentric contraction, compared to during isometric contraction, it has been shown that the joint torques across the elbow and other joints do not increase significantly above the peak isometric values (Amis et al. 1980; Hortobágyi and Katch 1990; Westing et al. 1988). Prescribing an overestimated torque limit would help compensate for the potential underestimation of peak joint torque in the model. This indicates that the shaking constraint imposed on the model can be considered realistic.

This study has provided the first probabilistic framework for investigating infant head kinematics during AHT. Probabilistic modelling was demonstrated to be scalable to different shaking scenarios, and provides a versatile tool to complement other modelling techniques in subsequent AHT research. Worst-case shaking scenarios were investigated across different age ranges, and the simulated head kinematics for physically plausible human shaking inputs did not exceed published SDH injury thresholds. Therefore, the results of this study provide no biomechanical evidence to demonstrate how shaking alone can cause
the injuries observed in AHT, suggesting either that additional factors, such as impact, are required to illicit injury, or that the published estimates of the injury thresholds are incorrect.

Acknowledgements

We would like to acknowledge the support of CureKids New Zealand, the Health Research Council of New Zealand Māori Health Doctoral scholarship (T.O. Lintern), Dr. McTavish from Starship Children’s Health, Auckland, New Zealand for providing the anatomical images of infants, and Associate Professor Brittany Coats from the University of Utah for sharing results from experimental studies on shaking of surrogates.

Disclosure statement

No potential conflict of interest was reported by the authors.

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