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Modelling 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. In this thesis, a novel computational framework for investigating head kinematics during AHT was developed using OpenSim (Delp et al., 2007), and was validated with kinematic measurements during shaking of an experimental phantom. The framework was used to investigate the biomechanics of AHT using model-based interpretation of animal shaking experiments, and computational studies simulating the shaking of human infants.

The lamb was used as an in vivo experimental analogue of AHT. An OpenSim computational model of the lamb was developed and used to interpret biomechanical data from shaking experiments. Sagittal plane acceleration components of the animal’s head during shaking were used to provide in vivo validation of the computational framework. Results demonstrated that peak accelerations occurred when the head impacted the torso and produced acceleration magnitudes exceeding 200 m·s⁻². The computational model demonstrated good agreement with the experimental measurements and was able to reproduce the extreme accelerations that occur during impact. The biomechanical results demonstrate the utility of using a coupled rigid-body modelling framework to describe infant head kinematics in AHT.

To investigate AHT in human infants, a novel probabilistic analysis of head kinematics during shaking was performed. A deterministic OpenSim model, incorporating an infant’s mechanical properties, was subjected to a variety of shaking motions. Monte Carlo analyses were 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, 4.5 months, and a 12 months infant. None of these cases produced head kinematics that
exceeded previously estimated subdural haemorrhage injury thresholds. 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, or that the current estimates of injury thresholds should be interpreted with caution.
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<td>ATD</td>
<td>Anthropometric test dummy</td>
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<td>CDF</td>
<td>Cumulative distribution function</td>
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<td>Characteristic length</td>
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Introduction

1.1 Motivation

Abusive head trauma (AHT) is a term used to describe brain injuries inflicted upon the most vulnerable members of society. The violent abuse of infants is, unfortunately, all too common. Inflicted injuries form a significant proportion of traumatic brain injury in young children (Billmire & Myers, 1985; Duhaime et al., 1992). In New Zealand, the annual incidence has been reported to be up to 19.5 cases per 100,000 infants, and is over-represented in the Māori population (Kelly & Farrant, 2008). The epidemiology of AHT is similar worldwide, with annual incidences being reported between 12.5 and 40 cases per 100,000 infants (Barlow & Minns, 2000; Hobbs et al., 2005; Jayawant et al., 1998; Keenan et al., 2003; Kesler et al., 2008; Talvik et al., 2006). AHT is most prevalent in infants under one year of age, but it is difficult to identify accurate statistics describing the epidemiology of AHT as many cases are unreported (Barr & Runyan, 2008; Theodore et al., 2005). The diagnosis of AHT is often missed because there are no pathognomonic symptoms, and diagnosis instead depends on a subjective interpretation of inconsistent or incomplete medical histories. Often the medical history is limited if the incident occurred when the accused perpetrator was alone, with no witnesses to provide collaborative information, and if the infant is unable to provide a history. It has been estimated that for every one infant who sustains an injury as a result of shaking, there may be up to 150 infants who are shaken, but the incidents go unreported (Theodore et al., 2005).

Infants who have injuries that are sustained as a result of shaking have high mortality and morbidity rates. A mortality rate of 25% has been reported (Barr & Runyan, 2008), and the injuries classified as AHT are significantly more severe than those inflicted accidentally (Irazuzta et al., 1997; Libby et al., 2003). Infants who survive typically do
very poorly. In a Canadian retrospective study, only 22% of infants were described as having no signs of health or developmental impairment upon discharge from hospital (King et al., 2003). Long-term symptoms include neurological and developmental problems, and may include motor impairment, visual impairment, or an inability to regain consciousness. Many infants who experience long-term injuries as a result of shaking require on-going medical care. In addition to emotional distress that is experienced by those directly involved, a large economic burden also exists. In retrospective studies performed in West Virginia and Colorado (USA), the direct costs of treatment have been estimated to exceed USD30,000 per infant (Irazuzta et al., 1997; Libby et al., 2003). These costs, however, ignore those associated with rehabilitation, long-term treatment necessities, a loss of societal productivity, or any legal costs that may be incurred.

Because there are no pathognomonic symptoms of AHT, the injuries sustained cannot be confidently attributed to shaking. It has been proposed that the injuries observed may be associated with other factors, such as blunt impact of the head with a hard surface, but controversy remains (Bandak, 2005; Goldsmith & Plunkett, 2004; Margulies et al., 2006; Pierce & Bertocci, 2008; Smith & Bell, 2008; Squier, 2008). Without a full understanding of the mechanisms of injury, there has been contention within criminal trials (Findley et al., 2012; Tuerkheimer, 2009). Producing a standard of evidence that is beyond reasonable doubt has forced experts to express confident opinions in a field where little confidence exists. It is critically important to address the legal contention that exists to ensure that the correct classification has been made. Incorrect assertions may have wide-ranging and detrimental consequences including the failure to punish those responsible for the child abuse, or the incorrect prosecution of an innocent party.

Biomechanics-based mathematical models have been recognised as important tools to aid in furthering understanding of the injury mechanisms of AHT (Duhaime & Dodge, 2008). Much more biomechanical research is necessary to improve existing biomechanical models to help investigate whether there is any causal relationship between shaking and the injuries described in AHT. This thesis describes a novel
computational modelling framework that complements prior work, and can be used to improve the understanding of injury mechanisms in AHT.

1.2 Thesis overview

1.2.1 Objectives

The primary goal of this thesis was to develop a computational modelling framework to describe infant head kinematics during manual shaking, and to use this framework to investigate the biomechanics of AHT.

The specific objectives were:

- To develop a verified computational model of an in vivo animal experimental analogue incorporating anatomical information obtained from computed tomography (CT) scans;
- To use the computational model to interpret and reproduce experimental recordings of head kinematics during manual shaking;
- To develop a computational model of a human infant that incorporates anatomical information obtained from CT scans and applies previously published material properties;
- To use the infant model to simulate head kinematics during different shaking scenarios and to compare results to published head injury criteria.

1.2.2 Chapter outline

Chapter 2 provides an historical overview of AHT and discusses previous research including a review of the animal, mechanical surrogate, and computational models that have been developed. The chapter outlines how this thesis proposes to address some of the limitations of previous research, and introduces the computational and experimental tools that were employed throughout this work.
Chapter 3 outlines the important computational methodologies that were used in this research. An experimental phantom was developed, and a series of manual shaking experiments were performed. The phantom was motivated by the anatomy of a lamb and contained joints with well-defined mechanical properties. The shaking kinematics were reproduced with an OpenSim rigid-body computational model (Delp et al., 2007), and the experimental results provided verification of the computational modelling framework and the parameter identification methodology that was used to interpret in vivo experimental results.

Chapter 4 describes the development of individualised OpenSim rigid-body computational models of neonatal lambs using anatomical information obtained from CT scans. A lumped-parameter description of intervertebral joint loading characteristics was implemented, and quasi-static loading experiments were performed to allow the mechanical properties of the models to be identified. The quasi-static results provided model parameter estimates that were used as initial values/conditions when reproducing head kinematics during subsequent in vivo shaking experiments.

Chapter 5 describes a series of in vivo experiments involving manual shaking of anaesthetised lambs. The existing computational models were scaled using anatomical landmarks, identified using magnetic resonance imaging (MRI), and were used to reproduce the head kinematics during shaking. The sensitivity of the model parameters was subsequently analysed and discussed. The chapter also describes the existence of macroscopic injuries sustained during shaking in lambs, and the suitability of using a lamb as an animal analogue of AHT.

Chapter 6 describes a series of OpenSim rigid-body computational models of human infants, and uses the modelling framework to investigate their head kinematics during shaking. The computational models were based upon anatomical information obtained from clinical CT scans and published experimental results. A probabilistic analysis was proposed to compare worst-case shaking scenarios to published injury thresholds, and to
previous experimental and computational results. A method of scaling the shaking results to different infant ages and different types of shaking motions was also proposed in this chapter.

Chapter 7 summarises and concludes the work and insights obtained in this thesis. Methodological limitations are acknowledged, and future directions for research that address some of these limitations are suggested.

1.3 Thesis contributions

- The skull and vertebra of four neonatal lambs were segmented from CT images and the relative positions and orientations of successive bones were used to develop custom coupled rigid-body computational models in OpenSim (Chapter 4). The models provided a basis for the novel model-based interpretation of head kinematics in an in vivo animal analogue of AHT.

- The lamb models were used to interpret experimentally-measured head kinematics during manual shaking (Chapter 5) using a novel parameter identification method that compared simulated head accelerations directly to accelerations measured using inertial measurement units (method described in sections 2.4.2 and 5.4.1).

- In vivo head kinematics of manually shaken lambs were successfully reproduced using the computational model (Chapter 5). To the author’s knowledge, this is the first time a physics-based computational model has been used to reproduce such in vivo experiments.

- A parametric analysis of model results demonstrated that the peak accelerations of the lamb’s head were a result of impact of the head with the torso. Results also demonstrated that many of the model parameters describing the mechanical response of the lamb’s neck were specific to the lamb and not transferrable to the human infant (Chapter 5).

- A coupled rigid-body computational model of a 4.5 months infant was developed using clinical CT scans and published material properties (Chapter 6). An existing
upper body musculoskeletal model was used to produce realistic shaking motions that were prescribed as kinematic (displacement) boundary constraints to the infant.

- A probabilistic analysis was performed to evaluate the cumulative probability distributions of the peak angular acceleration and velocity of an infant’s head during shaking. The analysis enabled the probability of exceeding published injury thresholds to be evaluated (Chapter 6). To the author’s knowledge, this study is the first probabilistic analysis of infant head kinematics in AHT, and allowed worst-case shaking scenarios to be investigated.

- Physically unrealistic shaking simulations were excluded by imposing a kinetic constraint upon the joint torques of the upper body musculoskeletal model. A peak isometric torque constraint was used and enabled worst-case shaking scenarios to be identified (Chapter 6).

- A methodology of scaling the results to infants of different ages and to shakers of different strengths was proposed (Chapter 6).

1.4 Publication list

This body of research has been submitted to international peer-reviewed journals, and presented at the following international conferences:


Representative OpenSim computational models of the lamb and human infant that were described in this thesis are publicly available on the OpenSim website, https://simtk.org/home/opensim. The lamb and infant models are available through the following individual project pages:

- https://simtk.org/home/lamb_headneck/
- https://simtk.org/home/infant_headneck/
Chapter 2

Background

Aspects of this chapter have been published in 2011 Fifth International Conference on Sensing Technology (ICST); and were described in an United States Patent Application (US 2013/0138412 A1).

2.1 Historical overview of abusive head trauma

The pathology of abusive head trauma (AHT) was described in many studies to be a manifestation of child abuse (Caffey, 2011; Kempe et al., 1962; Roche et al., 2005), but the association of infant brain injury with inertial head motion was first recognised by Norman Guthkelch, a British paediatric neurosurgeon, in 1971. Guthkelch noted the existence of subdural and retinal haemorrhage in infants without any external signs of trauma (Guthkelch, 1971). In regard to the bleeding, Guthkelch suggested that “repeated acceleration/deceleration rather than direct violence is the cause of the haemorrhage, the infant having been shaken rather than struck by its parent”. These findings were corroborated by John Caffey, a paediatrician who worked extensively in paediatric radiology, and he later coined the term whiplash shaken infant syndrome in 1974 (Caffey, 1974). He reclassified earlier cases of “battered baby syndrome”, described by Kempe et al. (1962), to be a result of whiplash due to a range of signs which he felt were associated with inertial head motion (shaking) rather than head impact. Whiplash shaken infant (or baby) syndrome has since been shortened to shaken baby syndrome (SBS), and the terminology remained until the consensus that shaking alone can produce the injuries was challenged. To remove any legal predisposition, the description SBS was replaced by non-accidental head injury (NAHI) (Mackey, 2006), and more recently by abusive head trauma (AHT) (Christian & Block, 2009). AHT is the terminology that has been adopted in this thesis.
The pathology of AHT is typically recognised by a triad of symptoms: subdural haemorrhage (SDH), bleeding into the subdural and subarachnoid spaces in the brain; retinal haemorrhaging, bleeding from the blood vessels in the retina; and acute brain dysfunction (Harding et al., 2003). This triad is often associated with a lack of external abrasions or signs of injury, and an incomplete or inconsistent medical history. Other injuries observed to occur during AHT include long bone or rib fractures (Caffey, 2011), injuries to the neck and brain stem (Brennan et al., 2009), diffuse axonal injury (Duhaime et al., 1987), and injuries associated with increased intracranial pressure (Chiesa & Duhaime, 2009). The bleeding in the brain is often bilateral (which is different to traumatic brain injury in adults), and is thought to be due to the shearing of bridging veins between the brain cortex and dura. This was suggested to occur during symmetrical motion induced during manual shaking (Guthkelch, 1971).

The assertion that the injuries were a result of whiplash motion was consistent with results from primate experiments investigating car accident injuries that were performed in the late 1960’s by Ommaya and others (Gennarelli et al., 1982; Ommaya et al., 1968; Ommaya & Gennarelli, 1974; Ommaya & Hirsch, 1971). It was observed that concussion and SDH could occur purely through rotational displacement of the head and neck without significant impact occurring. These studies proposed concussion and SDH injury thresholds that were a function of the peak angular acceleration and peak angular velocity of the primates head. Although these studies were performed to investigate car accidents, scaled versions of the injury thresholds have been used extensively in the biomechanical research into AHT, and have been used as metrics to investigate the hypothesis that shaking alone is sufficient to elicit the injuries of AHT.

In the ensuing decades, a number of studies have proposed other mechanisms of damage and have challenged the hypothesis that shaking is the sole contributor to AHT injuries. Duhaime et al. (1987) performed a clinical and biomechanical assessment of infant head injury using clinical observations and mechanical surrogate testing and concluded “that the shaken baby syndrome, at least in its most severe acute form, is not
usually caused by shaking alone. Although shaking may, in fact, be a part of the process, it is more likely that such infants suffer blunt impact”. Cory & Jones (2003) criticised the methods that Duhaime et al. (1987) used to scale the published injury thresholds, but when reproducing the results, were unable to conclude that shaking was able to cause the injuries of AHT.

In spite of the clinical and biomechanical evidence, a number of case studies continued to report incidences of brain injury without any signs of impact (Alexander et al., 1990; Gilliland & Folberg, 1996; Hadley, et al., 1989). Other studies reported shaking infant deaths, and were able to correlate the cases to a perpetrator’s confession or the infant’s medical history (Adamsbaum et al., 2010; Gill et al., 2009; Starling et al., 2004). An additional injury mechanism that gained legal traction was the Geddes hypothesis, which suggested that subdural and retinal bleeding could be caused by the initiation of a hypoxic cascade of events that were physiological, rather than traumatic, in origin (Geddes et al., 2001; Geddes et al., 2001a; Geddes et al., 2003; Geddes & Whitwell 2004). A number of studies refuted the causal relationship between hypoxia and SDH (Byard et al., 2007; Matschke et al., 2009; Hurley et al., 2010; Punt et al., 2004), and aspects of the Geddes hypothesis were retracted by the authors in the High Court of England (Mackey, 2006).

Controversies regarding the injury mechanisms of AHT persist (Bandak, 2005; Goldsmith & Plunkett, 2004; Margulies et al., 2006; Pierce & Bertocci, 2008; Smith & Bell, 2008; Squier, 2008), and the biomechanical questions, first raised by Duhaime et al. (1987), continue to be central in the arguments both for, and against, the hypothesis that shaking is sufficient to cause the injuries of AHT. A major implication of this uncertainty is the contention that it has led to in criminal trials (Findley et al., 2012; Tuerkheimer, 2009), where a consensus on the injury mechanisms has still not been reached. More biomechanical research investigating the relationship between shaking and AHT is necessary.
2.2 Existing models of AHT

2.2.1 Injury thresholds

To further the understanding of the injury mechanisms of AHT, a number of animal, mechanical surrogate, and computational models have been developed. Early experiments were performed on primates to investigate concussion and SDH injury thresholds during whiplash (Ommaya & Hirsch, 1971; Ommaya et al., 1968), and the injuries caused by translational and rotational head motion (Gennarelli et al., 1982; Ommaya & Gennarelli, 1974). The injury thresholds were measured in different species of primates and scale factors based upon the brain mass were proposed. These thresholds were extrapolated to the brain mass of human adults and were used to investigate traumatic head injuries resulting from car accidents (Holbourn 1956; Ommaya et al., 1967; Margulies et al., 1990; Margulies & Thibault 1992). The scaling relationships are described in Equations 2.1 and 2.2 where the brain mass, peak angular acceleration and peak angular velocity are denoted by \( M \), \( \ddot{\theta} \), and \( \dot{\theta} \), respectively.

\[
\ddot{\theta}_{Human} = \ddot{\theta}_{Primate} \left( \frac{M_{Primate}}{M_{Human}} \right)^{2/3} \tag{2.1}
\]

\[
\dot{\theta}_{Human} = \dot{\theta}_{Primate} \left( \frac{M_{Primate}}{M_{Human}} \right)^{1/3} \tag{2.2}
\]

The same relationship was used to scale injury thresholds to human infants by Duhaime et al. (1987), but experimental work in piglets has shown that age-dependent material properties of brain tissue may have a large effect on biomechanical injury thresholds (Thibault & Margulies, 1998). Thibault & Margulies proposed additional scaling using the elastic portion of the complex shear modulus \( (G') \) of paediatric brains to account for the developmental changes that occur to the brain’s material properties (Equations 2.3-2.4).
For 2.5% shear strains, the ratio of shear moduli \( \frac{G'_{\text{Infant}}}{G'_{\text{Adult}}} \) in porcine brains was measured to be 0.667, although it was suggested that mass-scaling alone may be sufficient at larger shear strains. The pigs used in this study were 2-3 days old and were equivalent to infants aged less than one month (Thibault & Margulies, 1998). In the absence of any other experimental data, this shear moduli ratio was used in this study. A number of assumptions were made when describing these injury thresholds. It was assumed that the brain is a homogeneous, isotropic, and elastic solid, and that the material properties, and geometry, of the brain are the same between species. It was also assumed that the skull is very stiff, when compared to the brain, and that excessive shear strains are responsible for the injuries that occur (Morison, 2002). Many of these assumptions are not valid, but the injury thresholds have continued to be used extensively for biomechanical research into AHT. Because of the approximate nature of the injury thresholds, and the difficulty in ascertaining human infant threshold values through experiments, further animal studies to investigate the injury mechanisms have been performed.

### 2.2.2 Animal experimental analogues

Animal experimental analogues have been used to investigate the pathophysiology that occurs during non-impact injuries. Experiments were performed with other primates (Kim, 2009), rodents (Bonnier et al., 2002; Kim 2009; Smith et al., 1998), piglets (Raghupathi & Margulies, 2002; Raghupathi et al., 2004) and lambs (Finnie, 2012; Finnie et al., 2012, 2010; Sandoz et al., 2012), but none have reported all the clinical features of AHT as a result of shaking. The recent results described by Finnie and colleagues described some lambs dying as a result of shaking, and observed histological evidence of
injury, but macroscopic bleeding was not observed. The results of these experimental studies demonstrate that a suitable animal analogue of AHT has still not been identified.

2.2.3 Mechanical surrogates

To investigate whether an infant’s head kinematics are likely to exceed injury thresholds established using animal experimental analogues, studies have been performed using mechanical surrogates. A biofidelic surrogate representing a 1 month infant was used to demonstrate that shaking was insufficient to exceed scaled injury thresholds without head impact occurring (Duhaime et al., 1987). These results were revisited by Cory & Jones (2003), who performed similar experiments using a range of biofidelic surrogates. They demonstrated that surrogates with different neck properties were able to produce head kinematics exceeding those described by Duhaime et al. (1987), and found some to exceed concussion injury thresholds. Cory & Jones’ study refuted Duhaime et al.’s categorical claims that shaking could not produce the injuries of AHT, but conceded that exceeding injury thresholds as a result of shaking appeared unlikely. In further studies, the head kinematics were contrasted during shaking and impact episodes using a 1.5 months biofidelic surrogate (Prange et al., 2003). Once again, no shaking result was found to exceed the existing injury thresholds and the results continue to contradict the case studies reported in the clinical literature. A limitation of the surrogate experiments is that the biofidelity is less than what is obtained in vivo. The surrogates do not include the vertebrae morphology, soft tissue structures, or the viscoelastic properties of an in vivo neck, and hence are unable to describe all of the anatomical complexities that are observed in infants. The shaking experiments also relied upon a manual shaking input, and there is no evidence to suggest that the shaking input is representative of the worst-case head kinematics.

2.2.4 Computational models

It is clear that existing animal and surrogate models have been unable to adequately describe all of the injury mechanisms in AHT. Further investigation has resulted in a
number of computational models being developed. Finite element models have been used to find mechanical correlates to the pathophysiology observed in AHT. Loading on the brain during shaking was investigated by Roth et al. (2007), but the results were limited by simplified constitutive relations and a single representative shaking boundary condition. A more complex model was developed by Couper & Albermani (2010), and included individual descriptions of different regions within the brain in addition to the cerebral spinal fluid and bridging vein insertion points. The model was, however, limited by its computational tractability and a single 4 Hz sinusoidal boundary condition being applied. Other studies investigated the strains necessary to rupture bridging veins (Morison, 2002), and the effect that an infant’s fontanel and brain-skull interface has on the loading response (Cheng et al., 2010), but both studies used simplified shaking boundary conditions. Cheng et al. (2010) applied boundary conditions that were representative of the motion applied to human infant dummies, but these were limited to in-plane motion and there is no evidence to suggest that the boundary conditions were representative of worst-case scenarios. As an infant’s brain is less stiff and the skull is much more deformable than an adult, (Coats et al., 2007; Margulies & Thibault, 2000; Roth et al., 2008), it is likely that similar finite element models will prove instrumental in identifying the infant-specific injury mechanisms of AHT. In subsequent studies, however, it would be important to prescribe boundary conditions that more completely describe the torso kinematics throughout a shaking episode.

Rigid-body computational modelling has been proposed to be an appropriate method to simulate an infants’ head kinematics during a range of shaking motions. Rigid-body computational models of an infant’s head and neck have been developed to investigate car crash injuries (Dibb et al., 2013), and have been used to describe head kinematics during infant shaking (Wolfson et al., 2005). Wolfson et al. produced a model that consisting of a series of ellipsoids that was based upon the standard MADYMO (TNO Automotive, Delft, Holland) CRABI 1 year-old test dummy. A sensitivity analysis was performed to confirm that the simulated head kinematics depended on the loading
properties of the neck. It was also observed that contact between the head and torso during shaking was sufficient to produce head kinematics that exceeded the published concussion injury thresholds. A limitation of this study was that the shaking boundary conditions used were obtained from a single representative example, and the biofidelity of the computational model was limited. Once again, there is no evidence to suggest the boundary conditions chosen produced worst-case shaking scenarios. The modelling results compared well with previous mechanical surrogate results (Cory & Jones, 2003; Duhaime et al., 1987), but a comparison to *in vivo* shaking kinematics was not made.

In summary, AHT is a very sensitive issue. Because *in vivo* infant shaking experiments are highly unlikely to ever be performed, there is a need to validate the computational methods by other means. Although existing animal models have not exhibited all the features of AHT, they can be used to validate the computational techniques necessary to investigate the injury mechanisms further. To complement finite element modelling studies of brain injury, a thorough description of an infant’s head motion during shaking is of importance, and is presently lacking in the literature. The head kinematics produced must not be restricted to representative examples but should instead encompass worst-case shaking scenarios. By simulating worst-case scenarios that account for different infant material properties, and different types of shaking inputs, the head kinematics can be compared to, and may help refine, existing injury thresholds, thereby improving the understanding of the injury mechanisms of AHT.

### 2.3 Anatomical considerations

In this study, a biomechanical analysis of AHT was performed using rigid-body computational models describing the head and neck of both lambs and human infants. This section will introduce the terminology and features of an infant’s anatomy which are relevant to this study. The anatomical structures will be contrasted between species, providing evidence that a lamb may be a suitable experimental analogue for investigating shaking biomechanics.
2.3.1 Anatomy of the head and neck

The skeletal structures of the head and neck consist of the bones of the skull and the cervical spine (Figure 2.1). The bones of the adult skull are joined together with rigid sutures that fuse together during development. Although the sutures are flexible, and a large gap between skull bones (fontanelle) exists in infants, the bones of the skull are considered as a single rigid-body for the purposes of this study. The cervical spine consists of seven cervical vertebrae (C1-C7) that are separated by flexible intervertebral joints. Relative motion of the vertebrae can occur as the joints undergo varying amounts of compression, tension, flexion, extension, lateral bending, and torsion. The cervical spine can be subdivided into two upper and five lower vertebrae that are differentiated according to the characteristic morphology that occurs in each region.

Figure 2.1 Skeletal system of the head and neck showing the skull, cervical spine and upper thoracic spine (image adapted from BodyParts3D database).

The morphology of the lower cervical vertebrae (C3-C7) is similar, and each vertebra consists of a roughly cylindrical vertebral body that is attached to a vertebral arch (Figure 2.2). On each end of the vertebral body is an endplate that interfaces with the intervertebral discs, and the arch forms a canal, known as the vertebral foramen, for the

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1 Adapted from freely available images from BodyParts3D, © The Database Center for Life Science licensed under CC Attribution-Share Alike 2.1 Japan
spinal cord to pass through. Each vertebra has two transverse and one spinal process that are landmarks associated with muscle attachment. In addition to the spinous processes, there are a number of articular processes that determine the articulation of adjoining vertebrae and form the facet joints. The articular processes act as geometric constraints to the joint motion but are covered in hyaline cartilage to facilitate sliding of the joint surfaces. The joints can undergo all modes of joint motion but the range of motion in each direction may vary throughout the length of the cervical spine (Penning, 1978).

The C1 vertebra, otherwise known as the atlas, is unique compared to the other vertebrae in that it does not contain a vertebral body. The C1 vertebra forms a joint complex with the skull and the C2 vertebra (known as the axis), and is referred to as the atlantoaxial-occipital joint complex (Eriksen, 2003). The predominant movements permitted by this joint complex are flexion and extension of skull relative to C1 and axial rotation of C1 about the superior process (dens) of C2 (Figure 2.3) (Bonadio, 1993). These movements allow for the “yes” and “no” shake of the head and are largely independent of the remaining cervical spine. A schematic of C1, C2 and C4 (a representative example of the lower cervical spine) is included in Figure 2.2.

![Diagram of C1, C2, C4](image)

**Figure 2.2** Major anatomical features of the cervical vertebrae. The C4 vertebra is included as a representative example of the lower cervical spine (image created using geometry from BodyParts3D database).
The mechanical loading responses of the joints in each direction are determined by the surrounding soft tissue structures. For the lower cervical spine, these tissues are dominated by the intervertebral disc, ligaments spanning between either adjacent vertebrae, or extending throughout the length of the neck, and by the spinal cord. The intervertebral disc is located between vertebral endplates and consists of multiple layers of fibrocartilage tissue. The disc absorbs up to 80% of compressive loads (Bonadio,
1993), and permits a small amount of motion to occur in the joint. Important ligaments in the neck include the anterior and posterior longitudinal ligaments that span the respective surfaces of the vertebral bodies, and the nuchal ligament that spans from the base of the skull along the spinous processes at the back of the neck (Figure 2.4). Other smaller ligaments connect adjacent vertebrae, and determine the localised behaviour of facet joints.

In the upper cervical spine, there are additional ligaments that span across different parts of the atlantoaxial-occipital joint and provide a complex attachment between the neck and the skull. The ligaments are able to constrain joint motion as their material properties exhibit a characteristic “J” stress-strain curve (Yoganandan et al., 2001; Yoganandan et al., 1989). When ligaments are stretched, their stiffness increases and they provide a stabilising force to the vertebral joints. However, as there are many ligaments with many different resting lengths, the lumped loading response of the intervertebral joints will appear graded as different ligaments progressively reach their limiting length (Panjabi et al., 1975).

The loading responses of the intervertebral joints are also determined by the musculature in the neck, which may contribute both passive and active muscle forces across the joints. As infants are known to have weak neck musculature (Bailey, 1952; Bonadio, 1993; Klimo et al., 2007; Roche & Carty, 2001), active muscle forces in the neck were not considered in this study. The passive muscle force contributions from each of the tissues were not considered individually, but their total contribution was combined with the graded loading response of the ligaments to provide a lumped description of joint loading.

2.3.2 Human infant anatomy

There are a number of developmental features that distinguish an infant’s cervical spine from the adult spine described above. The generation of bone tissue (ossification) in the vertebrae is incomplete in infants, and results in cartilaginous connections between
segments of bone. Complete ossification does not occur until a child is up to 12 years old (Bailey, 1952; Fesmire & Luten, 1989; Lustrin et al., 2003). Incomplete bone development results in shallow facet joints and undeveloped bone processes which increases the range of motion of the joints. There is also an increased laxity of infant ligaments resulting in more malleable joints that are better able to tolerate mechanical deformation. The fulcrum of flexion movement is higher in an infant, and is located at the C2-C3 joint compared to C5-C6 in adults (Bonadio, 1993; Klimo et al., 2007; Lustrin et al., 2003; Roche & Carty, 2001). It is clear that there are anatomical differences between infant and adult cervical spines which result in different loading characteristics that must be accounted for in subsequent computational models.

2.3.3 Lamb anatomy

Lambs have been suggested as an appropriate experimental analogue of AHT (Finnie, 2012; Finnie et al., 2012, 2010; Sandoz et al., 2012), and a lamb also has anatomical features in the neck that compare favourably to human infants. Although the sheep’s cervical vertebrae are typically longer in the axial direction than humans (Figure 2.5), they are otherwise anatomically quite similar (Cain & Fraser, 1995; Kandziora et al., 2001; Wilke et al., 1997).

To help support the head during quadrupedal motion, the nuchal ligament in sheep is somewhat larger than in a human which provides an increased passive resistance to flexion movements. The cervical spine also has larger spinous processes that produce larger joint moment arms to assist in supporting the animal’s head (Smit, 2002). The stiffness and range of motion of the cervical joints are also comparable to those of adult humans (Kandziora et al., 2001; Szotek et al., 2004), and the joints demonstrated an exponential loading response during compression and rotation (Clarke et al., 2007; Szotek et al., 2004). Similar developmental characteristics in the joint loading response were also described (Clarke et al., 2007), with lambs being observed to have more compliant joints with a larger range of motion than mature sheep. Although actual loading responses may be different, experimental evidence suggests that the constitutive relation describing the
joint loading is the same as in infants, and supports the use of a lamb as a biomechanical model to investigate infant neck dynamics during shaking.

![Comparison of lamb and infant c4 vertebrae with labelled anatomical landmarks. Sites of incomplete ossification can be observed on the infant vertebra. Anatomy was obtained from CT scans of a 5 day old lamb and a 4.5 months infant.](image)

**Figure 2.5** Comparison of lamb and infant c4 vertebrae with labelled anatomical landmarks. Sites of incomplete ossification can be observed on the infant vertebra. Anatomy was obtained from CT scans of a 5 day old lamb and a 4.5 months infant.

### 2.4 Computational and experimental tools

A number of experimental and computational tools were used in this study. The following section will describe the main technologies used to make experimental measurements, and the computational framework used to interpret these data.

#### 2.4.1 Motion capture

Optical motion capture is a common technique used to record movement in biomechanical research. The technology enables experimental markers on a subject to be quantitatively tracked through time in 3D space, and allows the measurements to be analysed computationally. This study used an OptiTrack™ motion capture system (NaturalPoint Inc, Corvallis, OR, USA) consisting of six Flex13 infrared cameras. Each of the cameras was used to track passive retro-reflective markers simultaneously at 120
frames per second to reproduce their 3D position to a reported marker precision of ±0.1 mm within OptiTrack™’s Tracking Tools software.

![Image](image_url)

Figure 2.6 A rigid-body representation of experimental motion capture markers (a) and a corresponding grey-scale image from one camera (b).

An advantage of the Tracking Tools software was that experimental markers could be clustered together and defined as rigid-bodies (or Trackables). The orientation and position of the rigid-bodies that best matched individual experimental marker positions could be tracked throughout time. If more than three markers were used to describe a rigid-body, redundancy was introduced to allow the position of the rigid-bodies to be described even when some markers were occluded. Interpreting the marker positions together enabled the measurements to directly relate to rigid-bodies within a computational model, reducing the post-processing time. An example of a rigid-body and a corresponding grey-scale image from one camera is illustrated in Figure 2.6.

Optical motion capture techniques can measure the absolute position of markers (and rigid-bodies) during very complex movements, but can be limited by marker occlusion and the frame rate of the cameras. To compensate for these limitations, this study has included additional inertial measurements and a physically based computational model to constrain and interpret the kinematic measurements.
2.4.2 Inertial measurements

Inertial measurement units (IMU) were used to complement the measurements obtained using the OptiTrack™ motion capture system. An IMU was developed and commercialised by Mark Finch from the Auckland Bioengineering Institute and IMeasureU Ltd (Auckland, New Zealand). Each IMU was inductively powered and contained a three-axis accelerometer (ADXL345), a three-axis gyroscope (ITG3200) and a three-axis magnetometer (HMC5843). Each sensor was therefore capable of measuring linear accelerations in three axes, angular velocities about three axes, and magnetic field strength in three axes (Finch et al., 2011). The three axis magnetic field strength measurements provided information about the orientation of the IMU with respect to the earth's magnetic field. The acceleration and angular rate measurements used in this study were sampled at 250 Hz and provided significantly higher temporal resolution than what could be obtained with the OptiTrack™ optical tracking system.

Inertial sensing has been used to measure a number of different biomechanical parameters including joint angles during gait (Yun et al., 2007), arm motion (Luinge et al., 2007; Zhou et al., 2005; Zhou et al., 2006), head motion (Foxlin, 1996), or the motion of generalised body segments (Zhu & Zhou, 2004). A limitation of inertial sensing motion tracking techniques is that integration of accelerations and angular velocities can result in the predicted position estimates drifting from their true values (Giansanti et al., 2003; Luinge et al., 2007; Luinge & Veltink, 2005; Welch & Foxlin, 2002). Many existing studies rely on cyclical motion, such as a foot striking the ground, or use sensor fusion algorithms, such as Kalman Filters, to constrain the estimates. Another technique is to constrain the measurements using a kinematic model that describes the geometry of a system (Zhou et al., 2005). A similar technique was employed in this study whereby a full physics-based dynamic model was used to provide both kinematic and kinetic information to constrain the measurements. A patent application has been filed describing the methodology (Finch et al., 2013), and the method is schematically described in Figure 2.7.
In this methodology, an arbitrary parameterised computational model of an experimental system is used to simulate kinematics. The kinematics can then be converted to measurements that were directly measured with the IMU to allow a direct comparison between the simulated and experimental measurements. Using nonlinear optimisation techniques, an objective function describing the difference between the measured and simulated measurements is minimised by iteratively updating the model parameters. The inertial components and any additional measurements, such as position estimates from motion capture, can be weighted appropriately to obtain a model simulation that best describes all available measurements.

This approach is elegant as it constructs an objective function where the model simulation results are compared directly to the sensor measurements, limiting the accumulation of error. Although the motion cannot be inferred directly from the sensor measurements as an accompanying model is necessary, this methodology allows unknown model parameters which best represent the physical system to be identified. One benefit of this approach is that the physically-verified computational model can be used for subsequent analyses without the need for further experimentation.
2.4.3 OpenSim musculoskeletal modelling

OpenSim is an open-source rigid-body musculoskeletal modelling software package that allows dynamic simulations of movement to be performed (Delp et al., 2007). An OpenSim model is an example of an appropriate dynamic model that can be used to interpret the experimental motion capture and inertial measurements. The methods and tools within OpenSim use the open-source multi-body dynamics libraries of SimTK. OpenSim and SimTK are well described elsewhere (Delp et al., 2007; Seth et al., 2010; Seth et al., 2011; Sherman et al., 2011), but the modelling and computational methods that were relevant to this study will be briefly described in the following section.

2.4.3.1 Model features

An OpenSim model consists of a chain of rigid-bodies that interact with one another. In this chain of rigid-bodies, there are joints separating successive bodies that are denoted as parent and child bodies, respectively. A child body is permitted to move relative to a parent body, which is arbitrarily oriented with respect to a global spatial reference frame (denoted "ground"), by the addition of generalised joint coordinates. The parent and child reference frames are described by $P_0$ and $C_0$, respectively (Figure 2.8). On each body, the location of a joint reference frame (P or C) is described relative its body frame. In the reference configuration, the parent and child joint frames align. Any relative movement of the bodies is described by a rigid-body transformation ($T$) between the fixed parent frame, $P$, and a moving child frame, $C'$. For a custom joint with six DOF, six generalised joint coordinates are necessary to describe the three translations (expressed in the child reference frame) and three Euler angles (1-2-3 body-fixed sequence) that form the transformation. If a joint translates and rotates simultaneously, the rotation occurs about the new translated child joint frame.

OpenSim also allows more complex biomechanical joints to be described using nonlinear mappings of joint DOF (Seth et al., 2010), and allows the addition of constraints to limit the motion of any joints. A set of rigid-bodies can be linked together in a chain using generalised coordinates that describe successive joints. The number of
generalised coordinates in each joint allows the potential kinematics of the model to be described. More information regarding what joints and constraints are permissible in OpenSim can be obtained from the OpenSim documentation (OpenSim, 2013).

2.4.3.2 Equations of motion

Newton’s second law can be used to derive the equations of motion (EOM) describing the relationship of the model’s generalised joint coordinates, \( q \), their velocities, \( \dot{q} \), and their accelerations, \( \ddot{q} \), to any forces and torques in the system (Equation 2.5);

\[
\ddot{q} = M^{-1}(q) \cdot \{ G(q) + C(q, \dot{q}) + R(q) \cdot f_m + F_{ext}(q, \dot{q}) \}
\]

where \( M^{-1}(q) \) is the inverse of the mass matrix, \( G(q) \) is a vector of generalised forces that arise from gravity, \( C(q, \dot{q}) \) is a vector of generalised forces that arise as a result of Coriolis and centripetal forces, \( R(q) \) is a matrix of muscle moment arms, \( f_m \) is a vector of generalised muscle forces, and \( F_{ext}(q, \dot{q}) \) is a vector of generalised forces that result from any external forces acting on the system. OpenSim models use generalised coordinates to

Figure 2.8 Relative motion of a child body with respect to a parent body is described by the transform, \( T \), of the respective joint reference frames. The joint frames are fixed with respect to the reference frames of each body and are included in the model description.
describe the degrees of freedom in each joint. A result of this is that a body’s velocity, and any applied forces (expressed in Euclidean coordinates), are mapped to generalised speeds and forces that relate to each generalised coordinate (Seth et al., 2010; Sherman et al., 2013; Thelen et al., 2003).

In typical musculoskeletal models, the joint torques are dominated by muscle forces acting across the joints and are governed by muscle contraction dynamics. The models described in this study do not include any muscles, but a lumped description of joint loading was used to describe these force contributions. The lumped joint loading description that was used in this study is discussed in Chapters 4-6. A “bushing” (*FunctionBasedBushingForce* in OpenSim) is a force object in OpenSim that allows joint forces and torques to be described as functions of the deviation between joint frames. For this study, the joint loading function was chosen to be an exponential function of the joint coordinates and was customised to represent the combined loading response of the neck. These equations are described further in Chapters 4-6. Linear viscous damping forces and torques that were proportional to \( \dot{q} \) were also included within the bushing definition.

Potential external forces that may be introduced into a model include reaction forces, such as those imposed during contact, and any additional external forces that are applied experimentally. In OpenSim, there are two contact models that can be used to describe contact interactions; the *HuntCrossleyForce* and the *ElasticFoundationForce*. The *HuntCrossleyForce* uses Hertz contact theory (Johnson, 1985) between simple geometric objects, while the *ElasticFoundationForce* calculates contact deformations of an arbitrarily complex meshed geometry using a simplified elastic model (Blankevoort et al., 1991). Both contact models include an energy dissipation model (Hunt & Crossley, 1975) and a model of Strubeck friction (Armstrong-Héouvry, 1991). A more detailed description is provided by Sherman et al. (2011). The *HuntCrossleyForce* contact model was used in this study.
The states of the model describe the generalised joint coordinates and their velocities at a given instant in time. From a user-specified initial state, the forces in the model are computed, and the states at subsequent time steps are calculated by numerically integrating the EOM. OpenSim has many different numerical integration algorithms that can be used for different modelling scenarios and a detailed description of each is included within the documentation (OpenSim, 2013).

There is a large suite of tools available in OpenSim that can be used to generate and analyse dynamic simulations. In this study, standard OpenSim implementations of inverse kinematics and inverse dynamics were used and are described below.

### 2.4.3.3 Inverse kinematics

The purpose of inverse kinematics is to find a set of generalised joint coordinates \( \mathbf{q} \) that optimally match a set of experimental measurements. To achieve this, a weighted least squares problem is iteratively solved (Equation 2.6) with the objective to minimise any marker and coordinate errors.

\[
\min_{\mathbf{q}} \left[ \sum_{i \in \text{markers}} w_i \left( x_i^{\text{exp}} - x_i(\mathbf{q}) \right)^2 + \sum_{j \in \text{unprescribed coords}} \omega_j \left( q_j^{\text{exp}} - q_j \right)^2 \right] \tag{2.6}
\]

A marker error is defined to be the Euclidian distance between a three-dimensional experimental marker position \( x_i^{\text{exp}} \), and a corresponding model marker \( x_i(\mathbf{q}) \) that is rigidly attached to a body within the model. The position of model markers is a function of the generalised joint coordinates. Similarly, a coordinate error is defined as the difference between a joint coordinate within the model \( q_j \), and an external measurement of a joint coordinate \( q_j^{\text{exp}} \). Experimental measurements are often obtained using motion capture, and different measurements can be weighted using a set of weighting functions \( w_i \) and \( \omega_j \), respectively. In an inverse kinematics analysis, the joint coordinates in the model are varied without any forces in the system being considered. Other methods are
necessary to infer the full dynamics of a system such as solving the full equations of motion or performing an inverse dynamics analysis.

### 2.4.3.4 Inverse dynamics

An inverse dynamics analysis is performed to calculate any net joint forces and torques that occur as a result of a prescribed motion. For a known set of generalised joint coordinates (such as those calculated during inverse kinematics), Equation 2.5 can be rearranged to calculate a vector of unknown generalised muscle forces (Equation 2.7). These generalised forces can then be mapped to the physical forces and torques that act on the joints (Sherman et al., 2013).

\[
R^{-1}(q) \cdot \{M(q)\ddot{q} - G(q) - C(q, \dot{q}) - F_{ext}(q, \dot{q})\} = f_m
\]  

Estimating accurate joint forces with an inverse dynamics analysis can be difficult. The method assumes that the prescribed motions are physically realistic and an accurate description of any external forces is necessary.

### 2.4.4 Monte Carlo probabilistic method

In many applications, probabilistic methods have been used to account for parameter uncertainty in computational models. A variety of probabilistic methods have been used in a number of biomechanical applications (Laz & Browne, 2010), but the Monte Carlo method was used in this study. The Monte Carlo method is often referred to as the ‘gold standard’ of probabilistic methods as solution convergence is ensured when the number of simulation trials tends towards infinity (Haldar & Mahadevan, 2000).

When simulating an infant’s head kinematics during shaking, the variation in an infant’s anatomy, material properties, and the type of shaking that they undergo can affect the outcomes. To account for this uncertainty, the Monte Carlo method was used to describe the statistics of the head kinematics (the performance metric) as a function of the probability distributions describing the model parameters. NESSUS, a probabilistic analysis software package (SwRI, San Antonio, TX, USA), was used to perform the
analysis by interfacing with the OpenSim API. Instead of performing a deterministic simulation in OpenSim using a single set of model parameters, the model parameters were sampled from a set of probability density functions (PDF) and NESSUS was used to generate the full cumulative distributions of the performance metrics. The main steps in this Monte Carlo probabilistic analysis are described below:

1. The performance metrics of a parameterised OpenSim computational model were described to be a function of a set of random variables (the model parameters);

2. Each random variable was described by a PDF that accounted for any uncertainty in the model parameters;

3. Each random variable was sampled from its PDF to form N simulation trials;

4. For each trial, the sampled random variables were used to obtain a single instance of the performance metrics in a deterministic model simulation;

5. The statistics of the deterministic results were calculated across the simulation trials using the Probabilistic Analysis tool in NESSUS;

6. Convergence of the statistics was assessed to ensure that consistent probability estimates could be made using the selected number of simulation trials.

The NESSUS software allowed the Monte Carlo method to be implemented quickly by interfacing directly with the OpenSim model. The software allows a range of probability distributions to be prescribed and includes many other probabilistic analysis techniques. Further information can be obtained from the NESSUS website; http://www.nessus.com/.

2.4.5 Summary

OpenSim is a musculoskeletal modelling software, which provided a computational framework for simulating the biomechanics of AHT. The computational framework was validated using experimental measurements obtained with both optical motion capture,
and inertial measurement units. This framework was used to reproduce head kinematics of an animal analogue of AHT during *in vivo* shaking experiments (Chapter 5), and was used to simulate infant head kinematics during shaking (Chapter 6). By combining this computational framework with the Monte Carlo statistical method, worst-case shaking scenarios were able to be investigated (Chapter 6).
Chapter 3

Shaking an experimental phantom

A simple experimental phantom was constructed to validate the computational framework for interpreting head kinematics during *in vivo* animal experiments. A coupled rigid-body computational model of the experimental phantom was developed in OpenSim, and the parameters of the model were identified using nonlinear optimisation techniques. This chapter describes the manual shaking experiments that were performed, and presents the optimisation and simulation results. Validation of the modelling and optimisation frameworks provides confidence when interpreting head kinematics in animal experiments, and supports use of the framework to describe the head kinematics of human infants during AHT.

3.1 Design of the experimental phantom

An articulated, multi-joint experimental phantom was developed to validate the parameter identification methodology used to interpret *in vivo* animal experiments. It was important that the properties of the phantom were well defined to ensure that accurate parameter estimates could be identified. The dimensions of the phantom were chosen to approximate that of a lamb for consistency with the experimental studies in Chapter 5.

A Lurethane® rod (Mulford Engineering Plastics, Auckland, New Zealand) was used to model the articulated neck structure, and to represent the soft tissue between the vertebrae. Lurethane® is a thermoset polyurethane elastomer, and a rod with a 90 Shore A hardness was used to develop the phantom (Figure 3.1). The vertebrae consisted of a number of aluminium collars that were clamped tight against the rod to ensure there was no movement of the rod relative to the collars, and that all rotation occurred in the spaces between the collars. The rod was fixed to an acrylic base plate, representing the torso, and masses were added to the top, using a number of acrylic fixtures, which represented the...
mass of the head. A 200 g and 400 g head mass configuration was used in two different phantoms by using two and four 100 g standard masses, respectively. All parameters for the phantom were well defined. The collar length ($v$) and spacing ($s$) were 20 mm and 6 mm, respectively, and the diameter of the Lurethane® rod was 12 mm. An inertial measurement unit (IMU) was fixed to the head element and retro-reflective motion capture markers were fixed to the torso element to measure the motion during shaking experiments.

![A schematic of the experimental phantom in the 200 g head mass configuration (left) and the constructed phantom in a 400 g head mass configuration (right). Computational models of each configuration were developed in OpenSim.](image)

Each physical phantom was reproduced computationally in the form of an OpenSim model. Bodies in the model included the collars, the 100 g standard masses, the IMU, the base plate, and the other acrylic fixtures. Each joint in the model was restricted to a single rotational degree of freedom (DOF), which was constrained to be within the plane of shaking, and the centre of rotation was defined to be in the centre of the spaces between the collars. The inertial parameters used for each body in the OpenSim model were calculated by assuming that both the collars and standard masses were solid cylinders, and that the acrylic fixtures, and IMU, were rectangular prisms. The torque in each joint,
\( \tau \), was described to be linearly related to the angle between vertebrae \( (\theta) \), where the linear stiffness was characterised by the parameter \( k \) (Equation 3.1). The joint properties were constrained to be same for each joint. A linear damping coefficient \( (c) \) was also used to describe the viscous damping within the joints. The damping force opposed the loading direction and was proportional to the angular velocity of the joint. All parameters were prescribed except for the joint stiffness, \( k \), and the damping coefficient, \( c \), which were identified by fitting model simulations to experimental measurements.

\[
\tau = -k\theta
\]  

3.2 Methods

3.2.1 Manual shaking protocol

Each phantom was shaken manually for twenty seconds in the sagittal plane. The type of shaking was similar to what has been reported to occur during the abusive shaking of infants, and was the same as that used in subsequent experimentation (Chapter 5). During the shaking, the torso element was tracked using the six camera OptiTrack™ motion capture system, and the motion of the torso element was prescribed as a kinematic (displacement) boundary constraint in the OpenSim simulations. The IMU was used to measure the sagittal plane acceleration and angular velocity components that occurred during shaking, and were compared to the simulation results.

Three independent shaking episodes were performed. Each shake was performed once and by the same person. The first shake was performed using a 200 g head mass configuration and was used as a training experiment to identify the unknown model parameters. Two subsequent shaking episodes were performed to test the predictive ability of the model. The first of these validation shakes was performed to reproduce the shaking scenario of the training experiments, and the second was performed using the 400 g head mass configuration. As all shakes were performed manually, the shaking boundary conditions varied between each shaking experiment. In all cases, the imposed
shaking motion was chosen to elicit maximum motion of the phantom head element, and the torso element motion for each shake was prescribed as a boundary constraint in the associated model simulation.

### 3.2.2 Parameter estimation

The unknown stiffness, \( k \), and damping, \( c \), in the model were identified using nonlinear optimisation techniques. The optimisation was performed in MATLAB (The MathWorks Inc, 2013) using the `lsqnonlin` algorithm in MATLAB’s Optimization Toolbox (Coleman & Li, 1994, 1996). The `lsqnonlin` algorithm uses a subspace trust-region method and is based on the interior-reflective Newton method, with each iteration involving the approximate solution of a linear system using the method of preconditioned conjugate gradients. For more details, refer to Coleman and Li (Coleman & Li, 1994, 1996). Model parameters were identified using the methodology described in section 2.4.2, whereby the computational model of the phantom was used to simulate the kinematics measured experimentally with the IMU. The unknown parameters in the model were then iteratively updated to minimise an objective function describing the difference between the simulated and experimental measurements.

![Figure 3.2 Sagittal plane kinematic components measured with the IMU.](image)

The objective function to be minimised, \( \varphi \), was defined as the weighted sum of squared errors between the measured and simulated sagittal plane kinematic components (Equation 3.2). The subscripts `model` and `exp` describe the simulated and experimental
IMU measurements, respectively. The sagittal plane components are described in the IMU coordinate system (Figure 3.2), and consisted of the linear acceleration in the $x$ direction ($a_x$), linear acceleration in the $z$ direction ($a_z$), and angular velocity about the $y$ axis ($\omega_y$). Each component was weighted using the variation of the noise within the respective measurements. The standard deviation of each measurement is represented by $\sigma_{ax}$, $\sigma_{az}$, and $\sigma_{oy}$, and were measured to be 0.1 m·s$^{-2}$, 0.17 m·s$^{-2}$, and 0.005 rad·s$^{-1}$, respectively. These were obtained by measuring the standard deviations of each sensor measurement while the IMU was stationary. The noise standard deviation for each measurement was less than 1% of the acceleration magnitudes measured during shaking.

$$\sum_{i=1}^{n=400} \left( \frac{(a_{x\text{model}} - a_{x\text{exp}})^2}{\sigma_{ax}^2} \right) + \left( \frac{(a_{z\text{model}} - a_{z\text{exp}})^2}{\sigma_{az}^2} \right) + \left( \frac{(\omega_{y\text{model}} - \omega_{y\text{exp}})^2}{\sigma_{oy}^2} \right)$$  

Each objective function evaluation required a single forward simulation of the parameterised computational model to be performed using the OpenSim API. The experimentally-measured boundary constraint was prescribed, and the equations of motion were solved over a time period from the initiation of shaking to the conclusion of a four second representative time sample. The representative time sample was selected after steady-state head motion had been established and was used to form the objective function. The transients associated with the beginning and end of each shake were not used in the objective function evaluation. Each time sample was downsampled from 250 Hz to 100 Hz and consisted of 400 data points. The equations of motion were solved using the implicit CPodes numerical integrator (Shermam et al., 2011) in the OpenSim API.

### 3.2.3 Parameter verification

The loading characteristics of a single joint were measured independently to verify the optimisation results. The same rod was fitted with two collars to create a single joint segment. The base of the joint segment was fixed in place while the top was deflected about a single axis (Figure 3.3).
During loading, the relative motion of each end of the joint segment was measured using the OptiTrack™ motion capture system. The forces and torques necessary to produce the motion were measured using a Nano17Ti six axis force/torque sensor (ATI Industries, Apex, NC, USA). The single plane force and torque components were used to calculate a net torque imposed on the joint. The result was then plotted against the joint angle estimated from the parameter estimation and the results were compared.

![Motion capture markers, Force/Torque sensor, Aluminium collars, Lurethane® rod](image)

**Figure 3.3** Loading verification experimental setup. A force was manually applied and was measured with the Nano17Ti force/torque sensor. The associated deflection was measured with motion capture to describe the loading response of the joint.

### 3.3 Results

#### 3.3.1 Optimisation results

To quantify the results, the root-mean-square error ($RMSE$) between the simulated, and experimentally measured sagittal plane kinematic components, $X_{\text{model}}$, and $X_{\text{exp}}$, respectively, was evaluated (Equation 3.3). The $RMSE$ was normalised by the difference between the maximum and minimum experimental kinematic components ($NRMSE$) (Equation 3.4). The normalised difference between the absolute maximum kinematic components ($dX_{\text{max}}$) predicted by the model, and the experimental measurements, were also evaluated, and are described by Equation 3.5. The sagittal plane kinematic components that were compared were $\omega_y$, $a_x$, and $a_z$. 
The stiffness and damping were identified to be 3.81 N·m·rad\(^{-1}\) and 0.073 N·m·rad\(^{-1}\)·s, respectively. Good agreement was observed between the simulated and sagittal plane kinematic components using the optimal parameter set and the experimental measurements (Figure 3.4). A slight underestimation of the acceleration components was observed with the \(dX_{\max}\) for \(a_x\) and \(a_z\) being measured to be -6.4 % and -3.5 %, respectively (Table 3.1).

![Graph showing comparison between experimental and simulated angular velocity, acceleration, and angular acceleration](image)

Figure 3.4 Optimisation results demonstrating that the optimal model simulation is in good agreement with the experimental measurements.
Table 3.1 The RMSE, and the difference in the maximum and minimum kinematic components, between the optimal simulation results and experimental measurements. The results demonstrate good agreement between the two datasets.

<table>
<thead>
<tr>
<th>Optimisation results</th>
<th>RMSE</th>
<th>NRMSE (%)</th>
<th>$dX_{max}$ (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\omega_y$ (rad·s$^{-1}$)</td>
<td>1.4</td>
<td>4.5</td>
<td>3.6</td>
</tr>
<tr>
<td>$a_x$ (m·s$^{-2}$)</td>
<td>3.5</td>
<td>4.3</td>
<td>-6.4</td>
</tr>
<tr>
<td>$a_z$ (m·s$^{-2}$)</td>
<td>3.9</td>
<td>12.0</td>
<td>-3.5</td>
</tr>
</tbody>
</table>

The NRMSE for the $\omega_y$, $a_x$, and $a_z$, kinematic components were evaluated to be 4.5%, 4.3%, and 12%, respectively. For the dominant $\omega_y$ and $a_x$ sagittal kinematic components, these errors are less than 5%, and the simulations were, therefore, considered appropriate for this application. Although the NRMSE associated with the $a_z$ component was large relative to the other components, the magnitude of $a_z$ was smaller. As larger accelerations are likely to be more important when investigating AHT, and the RMSE of $a_z$ was measured to be similar to $a_x$, a NRMSE of 12% was not considered to be a major limitation.

The optimisation was repeated from a number of initial parameter estimates to ensure that a unique parameter set was attained. The optimal estimates were found to be within 0.1% of 3.81 N·m·rad$^{-1}$ when initial estimates were varied between 2.0 N·m·rad$^{-1}$ to 6.0 N·m·rad$^{-1}$. Similarly, the optimal damping estimates were found to be within 0.3% of 0.073 N·m·rad$^{-1}$·s when the initial estimate was varied between 0.025 N·m·rad$^{-1}$·s and 0.15 N·m·rad$^{-1}$·s.

3.3.2 Predictive validation

The optimal parameters were used to simulate head kinematics during two additional shaking episodes. The first shaking episode was performed using the same phantom configuration (200 g head mass) as the training experiment, and good agreement between the simulated and experimental results was observed (Figure 3.5). The amplitude of each kinematic component is slightly higher than in the training experiment, indicating that the imposed shaking motion may have different characteristics.
Figure 3.5 Independent shaking episode using the optimal parameter set with the 200 g head mass configuration. Good agreement between the simulated and experimental measurements was observed indicating that the model is able to predict head kinematics during independent shaking events.

Figure 3.6 Independent shaking episode using the optimal parameter set with the 400 g head mass configuration. The magnitude of the experimental kinematic components was well reproduced indicating that the model is able to predict head kinematics in independent shaking events under different loading regimes.
The RMSE, and the difference in the maximum and minimum kinematic components, between the simulation results and the experimental measurements. The results are described for the 200 g head mass, and the 400 g head mass phantoms. The results demonstrate that the model is able to reproduce the magnitude of the kinematic components.

<table>
<thead>
<tr>
<th></th>
<th>200 g head mass phantom</th>
<th>400 g head mass phantom</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>RMSE</td>
<td>NRMSE (%)</td>
</tr>
<tr>
<td>( \omega_y ) (rad( \cdot )s(^{-1} ))</td>
<td>3.6</td>
<td>9.5</td>
</tr>
<tr>
<td>( a_x ) (m( \cdot )s(^{-2} ))</td>
<td>7.6</td>
<td>8.3</td>
</tr>
<tr>
<td>( a_z ) (m( \cdot )s(^{-2} ))</td>
<td>9.2</td>
<td>22.1</td>
</tr>
</tbody>
</table>

Although the underestimation of the acceleration components was more pronounced than for the optimisation case (Figure 3.4), the model was able to reproduce the magnitude of the remaining kinematics \( (dX_{\text{max}}) \) to a similar accuracy to the optimisation results (Table 3.2) during independent shaking events.

In the second shaking episode, the phantom configuration was changed to include a 400 g head mass, but the joint loading properties were unchanged. There was modest agreement between the simulated and experimental kinematic components which was demonstrated by larger NRMSE (Figure 3.6, Table 3.2). The shaking frequency necessary to elicit maximal head motion was different for the 400 g head mass phantom configuration, and was approximately half of what was measured in the training experiment. Although the simulated kinematic waveforms showed some differences to the experimental measurements, the magnitude of the simulated kinematic components was observed to overestimate the experimental maxima. The magnitude of \( \omega_y, a_x, \) and \( a_z \), was measured to exceed the experimental measurements by 17.3 %, 2.7 %, and 11 %, respectively. These results indicate that the model has a modest ability to reproduce the magnitude of head kinematics during different experimental configurations and loading regimes.

### 3.3.3 Verification

Further verification was provided by comparing the linear stiffness estimated by the optimisation to the independent loading measurements. The range of angles that individual joints were loaded through was comparable to the individual joint
deformations predicted by the model during shaking. The joint torque predicted using the linear constitutive model agrees well with the experimental loading measurements (Figure 3.7). Some hysteresis in the experimental results was observed, however this could not be reproduced by the model. This may be responsible for the different stiffness estimates being observed in each loading direction.

![Figure 3.7 Independent loading measurements (Exp) show good agreement to the linear constitutive properties predicted by the model (Model). The results validate the description of joint forces within the OpenSim model.](image)

### 3.4 Parameter sensitivity

The sensitivity of the model to different model parameters was evaluated by performing a deterministic sensitivity analysis. Each model parameter was varied individually, and the weighted RMSE between the perturbed result and the optimal result was measured. The weighted RMSE is described in Equation 3.6 where the subscripts model and optimal refer to simulation results obtained from the perturbed and optimal parameter sets, respectively.

\[
RMSE = \sqrt{\frac{1}{3n} \sum_{i=1}^{n} \left( \frac{a_{x model} - a_{x optimal}}{\sigma_{a_x}} \right)^2 + \left( \frac{a_{z model} - a_{z optimal}}{\sigma_{a_z}} \right)^2 + \left( \frac{\omega_{y model} - \omega_{y optimal}}{\sigma_{\omega_y}} \right)^2}
\]  

\[3.6\]
In addition to the stiffness and damping parameters, the mass of the head, the mass of the vertebrae and the joint distances were varied. The mean parameter values ($\mu$) were assumed to be the parameter set that was estimated during the optimisation and, in the absence of appropriate experimental data, the standard deviation ($\sigma$) was arbitrarily assumed to be 10% of each parameter’s respective mean values. These standard deviations were chosen to investigate the relative impact of each model parameter upon the kinematics throughout a relatively narrow range of parameter values. A narrow range of parameters was necessary because the model and imposed shaking boundary conditions were not independent. The limitations of this sensitivity analysis were avoided in Chapter 6, by allowing the input boundary conditions to also be varied. If large changes in parameter values were permitted, the Each parameter value was assumed to be normally distributed; the $\mu$ and $\sigma$ for each parameter are listed in Table 3.3. The weighted RMSE was evaluated as a function of normalised parameter values and was expressed as a standard score ($Z$). The deterministic sensitivity analysis was performed using NESSUS (SwRI, San Antonio, TX, USA), a probabilistic analysis software package. Each model parameter was varied within ±2 standard deviations about its means; the results of the sensitivity analysis are illustrated in Figure 3.8.

<table>
<thead>
<tr>
<th></th>
<th>$\mu$</th>
<th>$\sigma$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stiffness (N⋅m⋅rad$^{-1}$)</td>
<td>3.81</td>
<td>0.38</td>
</tr>
<tr>
<td>Damping (N⋅m⋅rad$^{-1}$⋅s)</td>
<td>0.073</td>
<td>0.0073</td>
</tr>
<tr>
<td>Joint distance (mm)</td>
<td>6</td>
<td>0.6</td>
</tr>
<tr>
<td>Head mass (kg)</td>
<td>0.1</td>
<td>0.01</td>
</tr>
<tr>
<td>Vertebrae mass (kg)</td>
<td>0.013</td>
<td>0.0013</td>
</tr>
</tbody>
</table>

The simulation result was observed to depend most strongly upon the stiffness of the joints and the mass of the head. The damping and joint distance had a relatively modest influence upon the simulation result, but the mass of the vertebrae had very little effect (Figure 3.8). The RMSE measured when the parameters were varied was largely a consequence of a phase difference being introduced between the optimal and simulated results, and was accompanied by only a small change in the magnitude (Figure 3.9).
3.5 Discussion

The experiments performed using the phantom described in this chapter demonstrated that accurate parameter estimates could be obtained during a single sagittal plane shaking
experiment. These results support the use of the computational framework to reproduce, and interpret, head kinematics during subsequent in vivo experiments (Chapter 5).

The design of the phantom used in these experiments was motivated by the anatomy of a lamb and consisted of an articulated neck-like structure. At the end of the neck was a relatively large mass (either 200 g or 400 g) that was representative of the head of a lamb. The phantom was complex with five independent articulating joints, but each had linear joint loading properties. The linear joint properties, and the lack of any additional soft tissue structures, however, resulted in a phantom that did not reproduce all loading characteristics of a lamb’s neck. Although the mechanical properties of the phantom were likely to be different from the properties of a lamb, the parameters estimated in the optimisation showed good agreement to independently measured mechanical properties (Figure 3.7). Reproducing the kinematics with accurate parameter estimates demonstrated that the customised OpenSim modelling framework was applying the joint forces correctly, and the optimisation framework was an appropriate method of estimating the model parameters from a single shaking episode.

The computational modelling framework also demonstrated a predictive ability, and was able to predict head kinematics during independent shaking episodes. Discrepancies in the simulated $a_z$ for the 400 g head mass configuration can be attributed to there being more out-of-plane motion than in the other experiments. Any motion in the $y$ direction would result in axial accelerations that would then be observed in the $a_z$ measurements. Even with some out-of-plane motion being imposed on the phantom, the overall magnitude of the kinematics was still well-reproduced. The reason for the systematic underestimation of the $a_x$ measurements is unclear. A spline was needed to fit to the raw motion capture data so the position and orientation of the torso could be converted to kinematic shaking boundary constraints. It is possible that the underestimation is a result of this smoothing.
The phase difference observed when the parameters were varied from their optimal estimates can be explained by considering the frequency response of the mechanical system. The manual shaking frequency that was chosen to elicit maximum motion of the head was assumed to consist predominantly of the fundamental frequency and its harmonics. This became evident when the head mass of the phantom was increased, which resulted in a lower shaking frequency being imposed upon the phantom (Figure 3.6). This effect is a similar characteristic to that of the resonant frequency of a linear mass-spring-damper system being proportional to $\text{mass}^{-1/2}$ (Meriam & Kraige, 1998). A variation in the parameters of the model will result in a change to the frequency response of the system and is characterised by the phase difference observed in Figure 3.9. On the other hand, the input boundary constraint driving the system was unchanged for all parameter perturbations in the sensitivity analysis, which explains why the magnitude and phase response of the simulated kinematics varied. The results indicated that a ±20% change to the parameter values resulted in only a minor change in the frequency response of the system, with only small changes in the magnitude and phase response being observed (Figure 3.9). An important consequence is that any parameter uncertainty is likely to only have a small influence on the magnitude of head kinematics, and may not be a major limitation in subsequent model simulations. It is likely that an in vivo model will have highly nonlinear mechanical properties, and the model’s frequency response function will be significantly more complex. The impact of model parameters will need to be reassessed in subsequent models, but it is likely that the head kinematics will continue to be relatively insensitive to the vertebral mass. This was demonstrated in the simulation results (Figure 3.8), where it was attributed to the large mass of the head. If the head mass in subsequent models continues to be the dominant mass in the system, inaccurate estimates of vertebral mass may not be a major limitation.

### 3.6 Summary

In this chapter, an experimental phantom was used to validate a customised OpenSim computational framework for interpreting in vivo head kinematics in an animal analogue.
of AHT. It was shown that the joint forces for an articulated experimental phantom could be prescribed, and the computational model parameters that described the joint forces could be identified from a single shaking episode. This framework was used to identify computational model parameters from \textit{in vivo} measurements obtained during a single shaking episode of an experimental analogue of AHT, as described in Chapter 5.
4 OpenSim model development

Aspects of this chapter have been published in Proceedings of the 11th International Symposium, Computer Methods in Biomechanics and Biomedical Engineering (2013); and were presented at the Australian Biomedical Engineering Conference (2012) and were awarded a Bioengineers in Education and Research Award.

This chapter describes the methodology used to develop an OpenSim biomechanical model of a lamb. Customised geometric models were developed using anatomical information obtained from computed tomography (CT) scans. The bones of the head and neck were individually segmented, and the relative positions of the bones were used to define joint descriptions within the OpenSim models. Approximations of the inertial and joint loading properties were identified using quasi-static force experiments, which provided initial parameter estimates for subsequent shaking simulations.

4.1 Anatomical information

Post-mortem CT scans were performed on four 5 day to 8 day old neonatal lambs using a Philips Brilliance 64 channel multi detector CT (Royal Philips, Amsterdam, The Netherlands). The experimental protocol was approved by the University of Auckland Animal Ethics Committee and was performed by Ascot Radiology at Ascot Hospital in Auckland, New Zealand\(^2\). The first two thoracic vertebrae (T1-T2), the cervical vertebrae (C1-C7), the skull, and the remaining torso were individually segmented, and wavefront (.obj) surface meshes were generated with the Surface Rendering tool in OSIRIX (Rosset, et al., 2004), a medical imaging and DICOM viewing software. The relative position and orientation of each bone was used to describe the joint properties, and computational models of each lamb were built in OpenSim (Delp et al., 2007).

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\(^2\) Ascot Radiology. [http://www.ascotrad.co.nz/](http://www.ascotrad.co.nz/)
4.1.1 Average anatomical surface meshes

Physical landmarks on each bone were identified so that anatomical coordinate systems of the vertebrae and skull could be defined. It was necessary to assign an anatomical coordinate system to each body in the OpenSim model, and it allowed any joint deformations to be interpreted in terms of anatomically meaningful joint loading directions. The physical landmarks were identified manually, and were a potential source of error. To prevent inconsistencies, the landmarks were instead selected on an average mesh for each bone, and the associated anatomical coordinate system of the average mesh was mapped onto each of the bones in situ. The steps to create an average mesh for each bone are described below and an example is illustrated in Figure 4.1:

- All bones were rigidly transformed to best align with each other. To do this, an iterative closest point algorithm (Besl & Mckay, 1992) was used where the corresponding data points from the surface meshes were selected using a KD-tree nearest-neighbour search. The Euclidean distances between the two sets of data points were then minimised by calculating the optimal rigid transformation between meshes;
- Once aligned, corresponding points in the lamb-specific meshes were successively identified using the nearest neighbour search and averaged across the four lambs;

![c3 meshes](image1)

![Average c3 mesh](image2)

Figure 4.1 Representative schematic of four lamb-specific c3 surface meshes (a) and the corresponding average mesh that was developed (b).
• Anatomical landmarks on the average mesh were selected using MeshLab\(^3\), and the anatomical coordinate system for each bone was defined.

For each of the average meshes, an anatomic coordinate system was assigned manually in accordance with the International Society of Biomechanics (ISB) recommendations (Wu et al., 2002). For the thoracic and cervical vertebrae (excluding C1) the axes were defined as follows:

- **y axis**: vector through the upper and lower endplates directed superiorly;
- **z axis**: vector directed laterally to the right between the left and right superior articular processes;
- **x axis**: cross product of the y and z axis directed anteriorly;
- **origin**: midpoint between the upper and lower endplates along the y axis.

For the C1 vertebra:

- **y axis**: vector directed superiorly between an inferior and superior landmark;
- **z axis**: vector directed laterally to the right between the left and right superior articular processes;
- **x axis**: cross product of the y and z axis pointing anteriorly;
- **origin**: centroid of the surface mesh.

For the skull:

- **y axis**: cross product of the z and x axis pointing superiorly;
- **z axis**: vector directed laterally to the right between the two occipital condyles;
- **x axis**: vector directed anteriorly between the tip of the nose and the midpoint between the two occipital condyles;
- **origin**: midpoint between the two occipital condyles.

\(^3\) MeshLab v1.3.1 64bit. Visual Computing Lab – ISTI- CNR. Current version available from http://meshlab.sourceforge.net/
As an example, the average mesh for the C4 vertebrae and its anatomical coordinate system are illustrated in Figure 4.2. Once all the average meshes with their anatomical coordinate systems were defined, each mesh was mapped to the lamb-specific bones \textit{in situ}, and the joint descriptions in OpenSim were defined.

![Figure 4.2](image)

\textbf{Figure 4.2} The average mesh for the c4 vertebrae overlaid with the anatomical coordinate system. The x axis points anteriorly, the y axis superiorly and the z axis laterally to the right. The origin was defined to be in the centre of the two vertebral endplates.

Any errors that are associated with the average mesh generation are unlikely to be a significant limitation. A misalignment of the mesh would result in errors in the orientation of the joint’s anatomical coordinate system, but would be mitigated by the model parameter $v_{\text{offset}}$, which accounts for any zero angle offset between the vertebral joints.

\subsection*{4.1.2 Joint descriptions}

To describe the joints in the neck for each lamb, the relative transformations between vertebrae needed to be ascertained. The methodology adopted to calculate these transformations is summarised in Figure 4.3, with the different shapes representing different vertebrae within the neck.

The functional spinal units (FSU) were considered one at a time and were defined to include an inferior parent vertebra (denoted $P$) and a superior child vertebra (denoted $C$). Initially, the FSU was \textit{in situ} and was described in the reference frame of the CT scanner (Figure 4.3(1)). To calculate the relative joint transformation, the position and orientation of the parent vertebra ($P$) was transformed to realign with the anatomical coordinate
system of the parent mesh. The same transformation was applied to the child vertebra ($C$) to maintain the relative transformation between bones in the FSU (Figure 4.3(2)). The realigned parent mesh ($P'$) was replaced by the average child mesh ($C_{av}$), which was then mapped onto the realigned child mesh, $C'$ (Figure 4.3(3)). This transformation, $T$, described the relative rigid-body transformation that existed between bones in the FSU of interest. Replacing the parent vertebra by the average child mesh was necessary so that the accurate mapping could be performed. Without this, any anatomical differences between vertebrae may introduce inaccuracies during the KD-tree nearest-neighbour search and may result in inaccurate joint transformations being calculated.

These joint transformations were calculated for all FSU’s in the neck and the process was applied to the joint between the C1 vertebra and the skull. The three Euler angles and the three translation components of the transformation were used to formulate the OpenSim model joint description. The surface meshes were also transformed to align with the anatomical axes and were included into the model for visualisation purposes.
For each joint, the joint centre was defined to be halfway between the origins of the parent and child vertebrae defined above (Wu et al., 2002). A six degree-of-freedom (DOF) joint was initially chosen to allow deformation in all directions and to maintain generality. Translations of the joint were described in the child coordinate system, and any rotation was described relative to the child coordinate system about the joint centre.

4.1.3 Experimental fixtures

Prior to the CT scans, an experimental fixture was attached to the torso and skull of each lamb for the purposes of optical tracking in subsequent experiments. Similar fixtures were included in the model, and landmarks on the fixtures were registered to the T2 vertebra and the skull by performing an inverse kinematic fit. The model fixtures were positioned and orientated to best match the position of experimental fixture landmarks that were observable in the CT scans.

![Diagram of experimental fixtures and model markers](https://simtk.org/home/lamb_headneck/)

Figure 4.4 Lamb-specific OpenSim computational model. Joints were described using anatomical information obtained from CT scans. The model includes fixtures and markers used for experimental analysis (model available at [https://simtk.org/home/lamb_headneck/](https://simtk.org/home/lamb_headneck/)).

Attached to each experimental fixture were a number of retro-reflective motion capture markers which were used for tracking in subsequent loading experiments. Model
markers were included that corresponded to the experimental positions, and allowed a direct kinematic comparison between the experiments and model simulations to be made. A sagittal view of the model in a neutral pose with model fixtures and markers attached is included in Figure 4.4. The joint coordinate systems associated with the parent vertebrae are also included and can be seen to have a joint-centre located between adjacent vertebral endplates.

4.2 Mass and inertial information

The joints describe the relative kinematics that are permissible between the rigid-bodies, but the dynamics are governed by the mass and inertial properties of each body within the model. Within OpenSim, the mass and the symmetric inertial tensor of each body must be defined. These properties could not be measured with precision in vivo without a post-mortem dissection so, instead, these properties were approximated using the anatomy derived from the CT images.

4.2.1 Skull body

The mass of the skull was measured by weighing the head after it was decapitated near the top of the neck. The mass measurement was imprecise due to the location of the decapitation and the loss of blood, but was estimated to be approximately 0.6 kg across the four animals. To account for this uncertainty, the head mass parameter was not prescribed in the modelling simulations, but was instead identified by fitting simulation results to experimental data. A body’s inertial tensor in OpenSim is defined with respect to the body’s centre of mass (COM). For the skull, the COM was defined to be at the centroid of the surface mesh point cloud (Figure 4.5). This simplifying assumption was made because there was low soft tissue resolution in the CT scans, which prevented an accurate estimate of the COM to be made. This limitation could be addressed in future research by performing an ex vivo analysis to locate the COM, or by making further assumptions regarding the soft tissue distribution within the skull. An approximation of the unit-mass inertia tensor \( I \) was derived from the mass-weighted covariance matrix.
(C) of the skull surface mesh, where $I_3$ is the $3 \times 3$ identity matrix (Balafoutis & Patel, 1991) (Equations 4.1-4.2). The inertia tensor was then scaled by the prescribed head mass in the modelling simulations. It was recognised that it would be more appropriate to calculate the COM of the volume that was described by the respective surface meshes. The approximation of only using the surface mesh information was made as it was difficult to determine the heterogeneous densities of the soft tissues within the skull from the CT images. Small discrepancies in the COM, and inertial properties, may exist, but were not considered to be a major limitation. It will be demonstrated in Chapter 5 and Chapter 6 that the peak shaking kinematics are, relatively, insensitive to the mass and inertial properties of the skull and vertebrae.

$$I = I_3 \text{tr}C - C$$  \hspace{1cm} 4.1

$$C = \sum_n x_n x_n^T$$  \hspace{1cm} 4.2

Figure 4.5 Sagittal view of the average skull mesh with anatomical coordinate system and COM defined.
As the inertia tensor is an approximation, it was calculated using the average surface mesh for the skull and was used in all models. The unit-mass inertial tensor for the skull is described in Equation 4.3.

\[
\mathbf{I}_{\text{skull}} = \begin{bmatrix}
I_{xx} & I_{xy} & I_{xz} \\
I_{yx} & I_{yy} & I_{yz} \\
I_{zx} & I_{zy} & I_{zz}
\end{bmatrix} = 10^{-3} \cdot \begin{bmatrix}
0.723 & 0.213 & 0.025 \\
0.213 & 1.211 & -0.023 \\
0.025 & -0.023 & 1.373
\end{bmatrix} \text{kg} \cdot \text{m}^2
\]

4.2.2 Vertebral bodies

The vertebral bodies in the neck form a relatively small proportion of the total neck volume and, as such, the inertial properties of the bones alone are insufficient to characterise the inertial properties of the neck segment associated with each bone. Because the CT scans had low soft-tissue resolution due to the cartilaginous properties of the lamb’s spine, accurate estimates of inertial properties using Hounsfield units was difficult. Instead, an elliptic cylinder was defined for each vertebra, and the inertial properties of each cylinder were chosen to approximate the inertia of each vertebra and associated soft tissues within a neck segment. A sensitivity analysis (see Chapter 6) indicates that this assumption is unlikely to present a major limitation. The minor and major axis dimensions of the cylinder were chosen to approximate the cross section of the lamb’s neck, and the depth was chosen to approximate the height of the vertebral bodies (Figure 4.6). The inertial tensor was approximated using Equation 4.4 (Wolfram Alpha, 2013).

By inspection of the CT scans, the lengths of the semi minor axis \((a)\), the semi major axis \((b)\), and the depth \((h)\), were approximated to be 42.5 mm, 30 mm and 20 mm, respectively. The mass of the neck segments was estimated by calculating the volume of the elliptic cylinder and assuming that the tissue inside was homogenous and had a density equal to that of water. Using these assumptions, the mass was estimated to be 80 g and the inertial tensor for each vertebra is described in Equation 4.5. Zero off-diagonal
components in the inertial tensor were a result of the symmetries that existed when approximating each neck segment as an elliptic cylinder.

\[
\begin{bmatrix}
I_{xx} & 0 & 0 \\
0 & I_{yy} & 0 \\
0 & 0 & I_{zz}
\end{bmatrix} = \begin{bmatrix}
\frac{1}{12}m(3a^2 + h^2) & 0 & 0 \\
0 & \frac{1}{4}m(a^2 + b^2) & 0 \\
0 & 0 & \frac{1}{12}m(3b^2 + h^2)
\end{bmatrix}
\]

\[
I_{\text{vertebrae}} = 10^{-6} \begin{bmatrix}
20.7 & 0 & 0 \\
0 & 54.1 & 0 \\
0 & 0 & 38.8
\end{bmatrix} \text{kg} \cdot \text{m}^2
\]

It was assumed that the mass of the head would dominate any inertial forces in the system and the vertebrae mass, COM, and the elliptic dimensions were only approximated. Due to these approximations, and the lack of published data on the inertial properties in lambs, the inertial tensor was prescribed homogenously throughout the length of the neck.

### 4.3 Joint loading characteristics

Exponential loading relationships have been demonstrated to be suitable for the paediatric human spine (Dibb et al., 2013; Luck et al., 2008, 2013; Luck, 2012; Nuckley & Ching, 2006; Nuckley et al., 2005; Ouyang et al., 2005), and a similar behaviour was
assumed for the lamb. A lumped description of joint torque during flexion and extension was prescribed using a bushing force (*FunctionBasedBushingForce* in OpenSim). The torque \(\tau\) within each joint was defined to be exponentially related to the joint angle (Equation 4.6) where \(A_f\) and \(A_e\) are gradient coefficients applied to each loading direction, \(k_f\) and \(k_e\) are the coefficients that control the mechanical response during flexion and extension, respectively, and \(\theta\) is the joint angle. The joints were allowed to behave differently in each direction and a spline was used to fit through both loading regimes to ensure derivative continuity when \(\theta\) is zero.

\[
\tau = \begin{cases} 
-A_f\left(e^{-k_f\theta} - 1\right); & \theta \leq 0 \\
A_e\left(e^{k_e\theta} - 1\right); & \theta > 0 
\end{cases} \tag{4.6}
\]

Although the model allows the full six degrees of freedom of joint motion, it was assumed that flexion and extension of the joints would be responsible for most of the head motion during shaking. Because of this, the joints in the model were restricted to a single rotation DOF in the sagittal plane and all other joint DOF’s were removed. The joints in the model were, therefore, restricted to rotation in the flexion and extension directions only. The loading properties were prescribed to be the same for all neck joints in the neck, excluding the atlanto-occipital joint (joint between C1 and the skull), which was described using the same exponential function but with a unique set of parameters (Equation 4.7). This single DOF model is likely to capture most of the head motion during sagittal plane shaking (the likely mode of shaking as suggested by Dr Kelly\(^4\), a child abuse expert), but it will be unable to capture out-of-plane head rotations during impact. This may be significant when investigating the shear loading on the brain and retina, but is beyond the scope of this study.

\[
\tau = \begin{cases} 
-A_f\ c_{15k}\left(e^{-k_f\ c_{15k}\theta} - 1\right); & \theta \leq 0 \\
A_e\ c_{15k}\left(e^{k_e\ c_{15k}\theta} - 1\right); & \theta > 0 
\end{cases} \tag{4.7}
\]

\(^4\) Dr Patrick Kelly, *Paediatrician and Clinical Director, Te Puaruruwha. Starship Children’s Hospital*
4.4 Quasi-static force experiments

A model-based analysis was performed to estimate the rotational properties of the joints within the neck and to provide quasi-static approximations of the model parameters. The head of each lamb was tracked through flexion and extension using optical motion capture whilst simultaneously measuring the forces and torques necessary to produce the motion. The forces and torques were applied as external boundary constraints to the model, and the simulated and experimental kinematics were compared. The following section introduces the experimental methodology and results of quasi-static flexion and extension bending experiments.

4.4.1 Experimental methods

Four anaesthetised neonatal lambs, aged between 4 days and 5 days, were subjected to neck loading in both flexion and extension. The experimental protocol was approved by the University of Auckland Animal Ethics committee. Anaesthesia was induced and maintained with an intravenous dose of propofol. During each loading experiment, the forces and torques necessary to manipulate the head were measured using a Nano17Ti six axis force/torque sensor (ATI Industries, Apex, NC, USA). To register the position and orientation of the head and torso to the forces measured, an OptiTrack™ motion capture system, consisting of six Flex13 infrared cameras, was used. An acrylic plate was screwed directly into each lamb’s skull to provide a rigid attachment for the force sensor (Figure 4.7(a)). A fixture containing 6 retro-reflective markers was mounted to the force sensor to be tracked with the motion capture system. The masses of all experimental fixtures were accounted for in the model to ensure the model parameters were representative of the lambs and not the experimental set up. Loads were applied through the marker fixture to manipulate the head, and deflection of the force sensor itself was negligible. The maximum deflections of the sensor at the rated forces and torques, using the rated stiffness of the sensor, were 12.3 μm, and 0.00143 rad, respectively.
To track the position of the torso, the spine was fixed near the 3rd and 4th thoracic vertebrae. Two cable ties were passed through the skin and under the spine, before being pulled tight and connected to an acrylic plate which was securely mounted to the table to provide a stationary reference point for the manipulations (Figure 4.7(b)). The plate contained four additional markers that were also tracked by the motion capture system. The points of attachment on the head and spine were later identified from CT scans.

![Figure 4.7 Experimental setup used with the head fixture and markers (a), torso fixture (b) and an example of extension loading (c).](image)

Once secure, the head was manipulated through extension and flexion while simultaneously measuring the marker positions, forces, and torques. The protocol consisted of first extending the head and neck backwards into full extension (Figure 4.7(c)), before rotating the lamb onto its back (with the markers on the spine still rigidly attached), and flexing the head and neck forward into full flexion.

### 4.4.2 Model-based parameter estimation

The dynamics of each manipulation were simulated using the lamb-specific OpenSim models. The forces and torques that were measured experimentally were prescribed as external force boundary constraints, and the model parameters were identified using nonlinear optimisation. The simulation methodology is illustrated in Figure 4.8. The experimental forces and torques, expressed in the head fixture frame of reference, were prescribed in the model. The model was initially orientated with the experimental markers using inverse kinematics before a forward simulation was performed using an initial
estimate of the parameters. The model-predicted positions of the head markers were compared to those measured using the motion capture system, and the parameters were updated iteratively to minimise the error between the model prediction and the experimental results.

As the flexion and extension bending manipulations were performed separately for each lamb, the simulations were performed consecutively for each objective function evaluation. This enabled the parameter set that best reproduced both modes of motion to be identified. For each time point, the Euclidian distances \((ED)\) between the experimental markers and the corresponding model-predicted marker positions were calculated. The objective function that was minimised, \(\varphi\), was defined to be the sum of the \(ED\) squared for each marker in the model (Equation 4.8). The indices \(i\) and \(j\) refer to the time index and marker number, respectively, and \(n\) is the number of time samples that were used to form the objective function. The optimisation was performed in MATLAB (The MathWorks Inc, 2013) using the \textit{lsqnonlin} algorithm in MATLAB’s Optimization Toolbox (Coleman & Li, 1994, 1996). This is the same algorithm that was used in Chapter 3 to identify model parameters of the experimental phantom.

![Parameter identification methodology](image)
\[ \varphi = \sum_{j=1}^{6} \left( \sum_{i=1}^{n} (ED_{i,j})^2 \right) \]

The head mass, the joint loading properties, and a viscous damping coefficient \((c)\) were all estimated during the optimisations. For the joint between C1 and the skull, independent \(A\) and \(k\) parameters were prescribed for flexion \((A_{f-c1sk} / k_{f-c1sk})\), and extension \((A_{e-c1sk} / k_{e-c1sk})\), and another set of loading parameters \((A_f, k_f, A_e, and k_e)\) were used to describe all of the remaining joints. Zero angle offsets were also applied to all of the joint torques. An offset for the C1-sk and all remaining joints were described by \(v_{offset-c1sk}\) and \(v_{offset}\), respectively. This offset was necessary as the true unloaded configuration of the neck was not known. The inertial properties of the vertebrae were not included in the optimisation, but were prescribed throughout the simulations.

To aid interpretation of the results, the root-mean-square error \((RMSE)\) between the model-predicted and experimental marker positions was also calculated, and is described by Equation 4.9. As previously, \(n\) refers to the number of time samples, and the coefficient \(6\) corresponds to the number of markers on the head fixture.

\[ RMSE = \sqrt{\frac{\varphi}{6n}} \]

4.4.3 Results and discussion

The optimal parameter estimates are listed in Table 4.1 and representative results are illustrated in Figure 4.9 and Figure 4.10. Only the results for three lambs are included as the skull fixture on Lamb2 was observed to come loose during the experiments. Significant variation in the parameter estimates was observed and this was attributed to the interdependence of model parameters and the limited experimental data used in the analysis. The large variation observed in the zero angle offsets was expected as it accounted for any residual joint loading when each lamb was imaged using CT. Although an unloaded pose was attempted in the scanner, it is likely that this was not achieved and
some residual joint loading was still present in this position. The overall \( \text{RMSE} \) between the model and experimental markers was found to be less than 35 mm for all three cases.

Figure 4.9 and Figure 4.10 provide a representative example of the optimal simulation results for Lamb3 during extension and flexion, respectively. The plots compare the \( y \) and \( z \) coordinates (in the ground reference frame) between one of the model and experimental marker positions. The model images below the plot demonstrate the overall simulation performance, with the model markers in pink and the experimental markers in blue.

<table>
<thead>
<tr>
<th>Name</th>
<th>Description</th>
<th>Lamb1</th>
<th>Lamb3</th>
<th>Lamb4</th>
</tr>
</thead>
<tbody>
<tr>
<td>( m )</td>
<td>Head mass (kg)</td>
<td>0.45</td>
<td>0.35</td>
<td>0.26</td>
</tr>
<tr>
<td>( A_f )</td>
<td>Flexion joint loading coefficient (N( \cdot )m)</td>
<td>0.08</td>
<td>0.05</td>
<td>0.06</td>
</tr>
<tr>
<td>( k_f )</td>
<td>Flexion joint loading coefficient (rad(^{-1}))</td>
<td>16.11</td>
<td>15.97</td>
<td>15.48</td>
</tr>
<tr>
<td>( A_e )</td>
<td>Extension joint loading coefficient (N( \cdot )m)</td>
<td>0.26</td>
<td>0.05</td>
<td>0.11</td>
</tr>
<tr>
<td>( k_e )</td>
<td>Extension joint loading coefficient (rad(^{-1}))</td>
<td>20.01</td>
<td>18.32</td>
<td>10.91</td>
</tr>
<tr>
<td>( A_{f-c1sk} )</td>
<td>Flexion C1-sk joint loading coefficient (N( \cdot )m)</td>
<td>0.05</td>
<td>0.12</td>
<td>0.16</td>
</tr>
<tr>
<td>( k_{f-c1sk} )</td>
<td>Flexion C1-sk joint loading coefficient (rad(^{-1}))</td>
<td>9.53</td>
<td>10.57</td>
<td>3.04</td>
</tr>
<tr>
<td>( A_{e-c1sk} )</td>
<td>Extension C1-sk joint loading coefficient (N( \cdot )m)</td>
<td>0.08</td>
<td>0.05</td>
<td>0.07</td>
</tr>
<tr>
<td>( k_{e-c1sk} )</td>
<td>Extension C1-sk joint loading coefficient (rad(^{-1}))</td>
<td>4.72</td>
<td>5.11</td>
<td>0.70</td>
</tr>
<tr>
<td>( \nu_{offset} )</td>
<td>Zero angle offsets (rad)</td>
<td>-0.00</td>
<td>0.04</td>
<td>-0.03</td>
</tr>
<tr>
<td>( \nu_{offset-c1sk} )</td>
<td>Zero angle offsets for C1-sk joint (rad)</td>
<td>0.03</td>
<td>0.08</td>
<td>-0.62</td>
</tr>
<tr>
<td>( c )</td>
<td>Viscous damping (N( \cdot )m( \cdot )rad(^{-1})( \cdot )s)</td>
<td>2.66</td>
<td>2.30</td>
<td>2.84</td>
</tr>
</tbody>
</table>

\( \text{RMSE} \) (mm)  
26.3  
17.6  
32.5

There was generally good agreement for both modes of loading, but the orientation of the head was not reproduced consistently throughout the entire simulation. When the head was orientated differently to what was observed experimentally, the force vector was also orientated differently, which acted to compound the initial misalignment of the head. It was possible that this may have contributed to inaccuracies in the joint loading parameter estimates.
Good agreement between the marker coordinates was observed but the orientation of the head was not reproduced consistently throughout the simulation.

Good agreement between the marker coordinates was observed but the orientation of the head appeared to overshoot the experimental position before settling near the experimental measurements.
There are number of limitations with this analysis. The model parameters were identified from a restricted dataset using a small range of kinematic measurements. A richer variety of loading regimes would be necessary to improve confidence in the parameter estimates. The range of the force/torque sensor limited the range that the head could be moved through. Perturbing the joint angles throughout their entire range of motion would provide a richer dataset with which to perform the parameter identification with. Perturbing the joints through a larger range of loading rates would also provide more information regarding the viscous damping within the model. Although quasi-static loading was assumed, the head was loaded at a non-zero velocity, which meant that damping in the joints needed to be modelled. It is likely, however, that the viscous forces will be significantly different during shaking events when the loading rate is significantly higher. These differences include increased impact stiffness during the fast decelerations that occur during contact events and stiffening of the vertebral joints during the higher velocity motion. With the limited experimental measurements, the number of identifiable model parameters was limited. A simplifying assumption to alleviate this was to constrain the lower neck joints to have homogenous joint loading properties. More detailed experimental measurements, such as using bone pins to track individual vertebrae, would assist in allowing individual joint properties to be identified.

A large amount of parameter interdependence was observed, which resulted in a number of local minima existing in the objective function. It is likely that this accounted for the head mass parameter being less than expected. One potential local minimum may produce a set of parameters with a small head mass, but with comparably stiffer joint loading properties, or alternatively, a larger head mass, but with more compliant joint loading properties. In the optimisations, the head mass consistently approached the parameters described in Table 4.1 and there were a number of potential errors and assumptions that may have contributed to this underestimation. Errors were introduced into the analysis when the lamb was rotated onto its back. There was potential for movement of the spine and head fixtures during the rotation, which would alter the
direction in which the external force was applied. The inertial properties of the vertebrae were also fixed during the simulations. An over approximation of the vertebral mass could also potentially result in an underestimation of the head mass.

In subsequent lamb shaking experiments, different animals were used and the mode of loading was significantly different from the quasi-static loading described. During shaking, the loading occurred at much larger strain rates, and it is likely that the nonlinearities associated with high joint angles, and high loading rates, may produce different parameter estimates than were observed here. Because of this, and the potential sources of error associated with these experiments, there was little importance placed on the model parameters described. Instead, they were used as initial estimates for the shaking simulations that were performed. This analysis produced parameter estimates that were specific to shaking (Chapter 5), which is the type of loading that is of interest in this study, and is most relevant to the clinical problem.

4.5 Summary

This chapter has described a methodology of developing custom OpenSim computational models of lambs using CT scans. The models were parameterised, and allowed the mass properties, inertial properties, and a lumped description of joint loading to be prescribed. A model parameter identification framework was introduced and quasi-static model parameter estimates were described. The model, parameter identification framework, and quasi-static parameter estimates provided a foundation for modelling shaking dynamics in subsequent chapters. This model development methodology was also used in the development of human infant OpenSim models in Chapter 6.
5

**In vivo** lamb shaking studies

Aspects of this chapter are currently under review in the *Journal of Biomechanics*.

Biomechanical measurements were made during manual shaking experiments to verify the use of a coupled rigid-body computational modelling framework to describe *in vivo* head kinematics during shaking. The shaking experiments were performed on lambs as they have been demonstrated to be a useful experimental model for abusive head trauma (AHT) (Finnie, 2012; Finnie et al., 2012, 2010; Sandoz et al., 2012). To complement the biomechanical measurements, a clinical assessment was performed to look for evidence of brain injury as a result of the shaking episode. This chapter presents the biomechanical and clinical results, and adapts an existing OpenSim computational model (described in Chapter 4) to reproduce the shaking kinematics measured experimentally. The sensitivity of the rigid-body model parameters to the shaking kinematics was investigated, and discussed in the context of reproducing head kinematics in infants during shaking.

5.1 Experimental methods

Six anaesthetised lambs were shaken manually in a manner that has been reported to occur during abusive incidents in human infants (Adamsbaum et al., 2010). All protocols were approved by the Animal Ethics Committee at the University of Auckland and were designed to cause minimal distress to the animals. Anaesthesia was induced with an intravenous dose of propofol, and maintained with inhaled isoflurane following intubation of the airway. The animals were aged between 5 days and 8 days, with mass between 7.0 kg and 8.8 kg.
Prior to any manipulations, the brain of each animal was scanned using magnetic resonance imaging (MRI) to establish a reference for subsequent analysis, and to provide geometric data for the development of a computational biomechanics model. While each animal was under anaesthesia, acrylic fixtures were rigidly attached to the spine and the skull. Four retro-reflective motion capture markers were attached to the torso fixture (Figure 5.1(a)) so that it could be tracked as a rigid-body using the six camera OptiTrack™ motion capture system. An inertial measurement unit (IMU) was attached to the head fixture (Figure 5.1(b)) to record the components of the linear acceleration and angular velocity of the skull during shaking. The positions of both fixtures were later registered to the spine and skull using MRI.

![Figure 5.1 Torso (a) and head (b) fixtures with motion-capture markers and IMU rigidly attached. The sagittal plane axes are shown for the IMU on the head.](image)

Each animal was gripped under each axilla (with the animal facing the shaker) and was shaken manually for twenty seconds in the sagittal plane. The experimental design was motivated by a complimentary clinical study (Bloomfield et al., 2009), and the shaking protocol was recommended by Dr Patrick Kelly⁵, a paediatrician and child abuse expert. After the shaking, each animal was maintained under anaesthesia and the heart

⁵Dr Patrick Kelly, Paediatrician and Clinical Director, Te Puaruruhau, Starship Children’s Hospital
rate, oxygen saturation, expired CO$_2$ concentration, and ventilation rate were recorded for between four and six hours to monitor the stability of the animal. The observation period was designed to allow time for any sustained injuries to develop. This protocol is consistent with those used in previous studies (Finnie, 2012; Finnie et al., 2012, 2010; Sandoz et al., 2012).

After the observation period, the brain and brain stem of each animal was imaged using a MAGNETOM® Skyra 3T MRI scanner (Siemens AG, Erlangen, Germany). Standard clinical sequences were recommended by Dr Andrew Smith, a neuro-radiologist (Auckland City Hospital, New Zealand) with experience with AHT, and were compared to the reference images performed prior to shaking. Image sequences included sagittal T1 and T2 weighted images, axial T2 weighted images, fluid attenuated inversion recovery (FLAIR), diffusion weighted imaging (DWI), and susceptibility weighted imaging (SWI). One additional pulse sequence was used in the post-shake imaging whereby a T2 weighted 3D Turbo Spin Echo (TSE) sequence was used to image the entire cervical spine to register the torso and skull fixtures to the skeleton. For two of the animals, a retinal examination was performed by an ophthalmologist.

All experiments performed in this study were terminal; the animals were euthanised with a lethal dose of pentobarbitone after the post-shake MRI. A post-mortem analysis of each animal was also performed. The animals were perfused through the carotid arteries in the neck with heparinised saline and subsequently with a 10 % neutral buffered formalin solution. After perfusion, the brain, brain-stem, and eyes were dissected, and stored in formalin solution before being mounted for histological analysis. Histological analysis is still ongoing and the results will not be discussed in this thesis.

5.2 Experimental kinematic results

The acceleration, $a$, and the angular velocity, $\omega$, were measured during manual shaking of four lambs using the inertial measurement unit described in 2.4.2. The measurements were sampled at 250 Hz. Unfortunately, during two of the experiments,
there were technical difficulties that prevented the inertial sensors from being used. A summary of the head kinematics during a single shaking event for four lambs (labelled 1, 2, 5, and 6) are described in the following section.

The frequency content of the head kinematics from both the accelerometer and gyroscope were calculated by determining the single-sided amplitude spectrum for the six measurements (3 degrees of freedom (DOF) per sensor). A representative example (Figure 5.2) illustrates that the majority of power in the accelerometer for Lamb5 occurs at a fundamental frequency of approximately 1.4 Hz, with successively less power contained in higher harmonics. The large DC offset observed can be attributed partly to gravity, but was dominated by the low frequency centripetal accelerations (measured in the $z$ direction) that were lumped into the DC frequency bin. There was no DC offset observed in the amplitude spectrum of the gyroscope measurements.

![Figure 5.2 Representative amplitude spectrum describing the frequency content of raw accelerometer measurements (Lamb5).](image)

Different fundamental frequencies were observed between different lambs, but all were found to match the frequency of the shaking motion applied to the torso. The amplitude spectra for the accelerometer and gyroscope measurements in each lamb are summarised in Table 5.1 in addition to the total weight and sex of each lamb. The fundamental frequency ($F_0$) of the kinematics varied between 1.4 Hz and 2.1 Hz, but they do not appear to correlate with the lamb’s weight. For all lambs, the magnitude of
acceleration components at the fundamental frequency \((|a_i|_{f0})\) are largest in the \(x\) and \(z\) directions, and the magnitude of the angular velocity components \((|\omega_i|_{f0})\) is maximal about the \(y\) axis, which is consistent with the shaking motion being predominantly in the sagittal plane. The non-zero out-of-plane components can be explained by the shaking motion not being entirely in plane or due to the sensor not being mounted squarely with respect to the skull.

Table 5.1 Frequency content of experimental head kinematics during shaking indicating that the major components are those within the sagittal plane. Sagittal plane kinematics are described by \(|a_x|_{f0}\), \(|a_z|_{f0}\), and \(|\omega_y|_{f0}\), respectively.

<table>
<thead>
<tr>
<th>No.</th>
<th>Weight/Sex</th>
<th>(F_0) (Hz)</th>
<th>(a) components (m(\cdot)s(^{-2}))</th>
<th>(\omega) components (rad(\cdot)s(^{-1}))</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td>(</td>
<td>a_x</td>
</tr>
<tr>
<td>1</td>
<td>7kg/M</td>
<td>2.08</td>
<td>18.1   12.6   17.4   3.5   10.6   4.7</td>
<td></td>
</tr>
<tr>
<td>2</td>
<td>7kg/F</td>
<td>1.56</td>
<td>12.1   2.9    21.0   2.9   10.0   1.1</td>
<td></td>
</tr>
<tr>
<td>5</td>
<td>7.5kg/F</td>
<td>1.42</td>
<td>15.9   6.2    25.4   4.3   10.4   5.0</td>
<td></td>
</tr>
<tr>
<td>6</td>
<td>8.8kg/M</td>
<td>1.68</td>
<td>8.1    5.3    9.1    1.8   5.3    2.7</td>
<td></td>
</tr>
</tbody>
</table>

To analyse the characteristics of the kinematics, and to provide a dataset for model parameter estimation, a small representative time sample of head kinematics was chosen after steady-state motion had been established. These regions were qualitatively identified, but were chosen to include consistent and maximal accelerations. Transients associated with beginning or end of each shake were not used in this study. An example of the representative time sample for Lamb5 is shown in Figure 5.3, with coloured circles representing maxima and minima for each of the major axes of the sagittal plane motion. Figures illustrating the measurements for the remaining lambs are included in Appendix A.1. The magnitudes of the maxima and minima were quantified for all lambs in Table 5.2 and are described using the mean \((\mu)\) and standard deviation \((\sigma)\) of the maxima and minima throughout the sample.
Table 5.2 Maxima and minima characteristics of each of the sagittal plane kinematic components. Large values for $a_x$ are indicative of contact occurring during full extension ($a_x$ Max) and full flexion ($a_x$ Min).

<table>
<thead>
<tr>
<th>No.</th>
<th>$a_x$ ($\mu \pm \sigma$)</th>
<th>$a_z$ ($\mu \pm \sigma$)</th>
<th>$\omega_y$ ($\mu \pm \sigma$)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Max (m/s$^2$)</td>
<td>Min (m/s$^2$)</td>
<td>Max (m/s$^2$)</td>
</tr>
<tr>
<td>1</td>
<td>140.5±18.6</td>
<td>-125.7±28.8</td>
<td>155.8±4.6</td>
</tr>
<tr>
<td>2</td>
<td>157.0±0</td>
<td>-126.0±32.6</td>
<td>158.1±2.9</td>
</tr>
<tr>
<td>5</td>
<td>107.3±32.0</td>
<td>-122.8±27.5</td>
<td>154.7±3.8</td>
</tr>
<tr>
<td>6</td>
<td>46.2±21.5</td>
<td>-150.1±9.9</td>
<td>124.3±18.5</td>
</tr>
</tbody>
</table>

Important characteristics that were observed in the kinematic waveforms are the maxima and minima that occur during full flexion and during full extension. In Figure 5.3, flexion contact is characterised by the $a_x$ minima being large in magnitude, and the timing of the minima being coincident with the $a_z$ minima. Contact was attributed to the impact occurring between the lamb’s head and the shaker’s hands/arms that was observed experimentally. Contact of the lamb’s head and torso during full extension was characterised by the $a_x$ maxima being large in magnitude.
Flexion and extension contact were not observed to occur in every shaking cycle. This was exemplified by Lamb6 (Figure A.3), which had relatively small mean $a_x$ maxima, and was indicative of full extension contact not being observed. The standard deviations of the acceleration maxima help to describe whether or not contact occurred. A large standard deviation is indicative of the different contact modes not occurring during every shaking cycle. A small standard deviation was observed when the mean acceleration maxima approached 160 m·s$^{-2}$, which was the maximum rating of the accelerometer. This saturation of the accelerometer is responsible for the zero standard deviation observed in Lamb2 (Table 5.2).

Although there was considerable variation in the magnitude of the accelerations, contact occurs when the angular velocity is zero, and this variability was not evident in the gyroscope measurements. The $\omega_y$ maxima and minima are much more consistent with the maxima being between 30.95 rad·s$^{-1}$ and 36.32 rad·s$^{-1}$ and the minima between -19.83 rad·s$^{-1}$ and -25.46 rad·s$^{-1}$.

Another meaningful kinematic metric is the magnitude of the acceleration vector. The acceleration magnitude was calculated by calculating the Euclidian norm of the three acceleration components that were expressed in the IMU coordinate system. The experimentally-measured acceleration magnitudes within the representative time sample are included in Figure 5.4. For all lambs, the maximum acceleration was observed to occur during impact of the animals head with the torso, or impact with the shakers hands, but the anatomical configuration when contact occurred was not consistent throughout the experiments. Contact occurred during both flexion and extension in Lamb1 (Figure 5.4(a)) and Lamb5 (Figure 5.4(c)), exclusively during full extension in Lamb2 (Figure 5.4(b)) and exclusively during full flexion in Lamb6 (Figure 5.4(d)). It is likely that these differences were due to slightly different shaking motions being applied experimentally. When contact did occur, the maximum acceleration magnitude was consistently measured to be between 200 m·s$^{-2}$ and 250 m·s$^{-2}$ and always resulted in the maximum acceleration measured within the representative time sample.
5.3 Model description

5.3.1 Model scaling

The generic OpenSim model, developed in Chapter 4, was used as a basis for interpreting the experimental measurements. The quasi-static experimental fixtures in the generic model were replaced by the fixtures used in the shaking experiments. Anatomical landmarks (Table 5.3 and Figure 5.5), and landmarks on the fixtures, were identified in the 3D TSE MRI, and were used to orient the sensor fixtures in the model to different parts of the skeleton using inverse kinematics. The torso and skull fixture were described with respect to the T2 vertebrae and the skull, respectively, and the process was repeated for Lambs 1, 2, 5 and 6. The models are publically available from https://simtk.org/home/lamb_headneck/.
The generic model was then scaled to match each experimental lamb. To do this, each joint was allowed to stretch axially by uniform amounts to scale the model to match the T2 and skull anatomical landmarks. After the scaling, joint degrees of freedom were removed such that the spine could not stretch axially, thereby locking the model into the scaled configuration. All axial displacements were between 1 \text{ mm} and 3 \text{ mm per joint}.

<table>
<thead>
<tr>
<th>T2</th>
<th>Skull</th>
</tr>
</thead>
<tbody>
<tr>
<td>Left articular process$^1$</td>
<td>Left occipital condyle$^5$</td>
</tr>
<tr>
<td>Right articular process$^2$</td>
<td>Right occipital condyle$^6$</td>
</tr>
<tr>
<td>Superior vertebral endplate (centre)$^3$</td>
<td>Anterior tip of mandible$^7$</td>
</tr>
<tr>
<td>Inferior vertebral endplate (centre)$^4$</td>
<td></td>
</tr>
</tbody>
</table>

Table 5.3 Anatomical landmarks used to register torso and skull fixtures. The numbers correspond to those in Figure 5.5.

Figure 5.5 Illustration of anatomical landmarks with labels corresponding to Table 5.3.

5.3.2 Model parameterisation

After scaling, the nine vertebral joints within the model were each restricted to a single rotational DOF in the sagittal plane. Instead of including individual soft tissue elements within the model, a lumped-parameter description of joint torque was defined to represent the passive torque contributions of the muscles, ligaments, and other soft tissues for each vertebral joint within the neck. The same joint loading relationship that was described in section 4.3 was used in the model but a simplified parameterisation was
performed (Equation 5.1). The torque ($\tau$) within each joint was defined to be exponentially related to the joint angle (Equation 5.1), where $A$ is a gradient coefficient applied to each loading direction, $k_f$ and $k_e$ are the coefficients that control the mechanical response during flexion and extension, respectively, and $\theta$ is the joint angle. Rotation during extension was defined as a positive joint angle within the model.

$$\tau = \begin{cases} -A(e^{-k_f\theta} - 1); & \theta \leq 0 \\ A(e^{k_e\theta} - 1); & \theta > 0 \end{cases}$$

The joints were allowed to exhibit different loading characteristics in each rotation direction and a spline was used to fit through both loading regimes to ensure derivative continuity when $\theta$ is zero. To simplify further, the properties of each joint were assumed to be homogenous meaning that the mechanical properties of the entire neck could be described by the three parameters, $A$, $k_f$, and $k_e$. In addition, a linear viscous damping coefficient, $c$, and the mass of the head, $m$, were included in the parameterised model. The generic inertial properties of the model were fixed for the simulations.

Hunt-Crossley contact spheres (Delp et al., 2007; Sherman et al., 2011) were included in the model to describe the contact interactions between the lambs’ heads and torsos (Figure 5.6). The contact stiffness ($K_{cont}$) and location of the contact spheres during flexion ($Contact_{flex}$) and extension ($Contact_{ext}$) were identified by fitting model simulations to experimental measurements. $Contact_{flex}$ is a parameter that defines the rotation of the flexion contact sphere about T2’s anatomical coordinate system with a zero angle corresponding to alignment with the $x$ axis ($x_{T2}$). $Contact_{ext}$ describes the translation of the extension contact sphere along $x_{T2}$. The radius of curvature of the flexion contact, the $y$ axis ($y_{T2}$) offset of the extension contact, and the location of contact spheres on the skull were estimated and fixed in the simulations. A schematic describing the features and parameters of the model is included in Figure 5.6.
5.3.3 Boundary constraints

The experimentally-measured torso motion was prescribed as a kinematic (displacement) boundary constraint in the simulations. Inverse kinematics was used to best match the location of model markers on the torso fixture to those measured experimentally with motion capture. In order to do this, the three translational and three rotational model coordinates describing the position and orientation of the T2 vertebrae with respect to the ground reference frame were adjusted to track the motion capture markers throughout time. After the inverse kinematics analysis, a spline was fit through each coordinate and it was used to prescribe each coordinate in subsequent simulations. A representative example of T2 Euler angles (1-2-3 body-fixed sequence) and positions that were described with respect to the ground reference frame are illustrated for Lamb6 (Figure 5.7). The remaining boundary constraints that were used for the other lambs are included in Appendix A.2.

Figure 5.6 Generic OpenSim computational model of a lamb. The model was developed using anatomy from CT scans and included a lumped description of joint loading within the neck. Contact was modelled to describe impact of the head with the torso and fixtures were used to register the model with experimental shaking (model available at https://simtk.org/home/lamb_headneck/).
5.4 Model parameter estimation

5.4.1 Optimisation procedure

The model parameters describing the joint loading, contact, and inertial properties were identified using nonlinear optimisation techniques. The optimisation was performed in MATLAB (The MathWorks Inc, 2013) using the *lsqnonlin* algorithm in MATLAB’s Optimization Toolbox (Coleman & Li, 1994, 1996). A similar methodology was introduced in Chapter 4 (section 4.1.2). Within each function evaluation in MATLAB’s optimisation procedure, a single forward simulation was performed of the parameterised OpenSim model using the OpenSim API. The equations of motion were solved over a time period which began at the initiation of shaking and ended at the conclusion of the representative time sample using the implicit CPodes numerical integrator (Sherman et al., 2011). To improve the optimisation performance, the finite differencing steps used to approximate the Jacobian matrix in the optimisation were implemented in parallel.

Figure 5.7 The Euler angles (θ) and absolute positional coordinated of the T2 vertebra that were prescribed as kinematic (displacement) boundary constraints for Lamb5.
The objective function to be minimised, $\varphi$, was defined as the sum of squared errors in the acceleration magnitude simulated by the model, $|a_m|$, and the acceleration magnitude measured experimentally, $|a_{imu}|$, with the IMU (Equation 5.2). The acceleration magnitude was used in the objective function instead of the angular acceleration, which is the injury metric used in subsequent infant simulations (Chapter 6). This was done because the acceleration components were measured directly by the IMU and the accumulation of errors associated with differentiating the angular velocities measurements was avoided.

The acceleration magnitude was calculated by calculating the Euclidian norm of the three acceleration components expressed in the IMU coordinate system. The angular velocities were not included in the objective function as they resulted in the kinematics being weighted more towards the motion that occurred away from contact events. Using the acceleration magnitude allowed kinematics during contact events, characterised by peaks in the acceleration, to be favoured. It was these contact events that were considered the most important when investigating AHT.

$$\varphi = \sum_{i=1}^{n} (|a_m|_i - |a_{imu}|_i)^2$$  \hspace{1cm} 5.2

The objective function was evaluated using a single shaking period and the measured kinematics were sampled at 250 Hz. The number of measurements within a single shake varied between experiments but the objective function was normalised by the number of data points, $n$, within the shaking period. The period with the largest magnitude of acceleration within the representative time sample was used in the objective function evaluation.

During the optimisations, parameters were constrained to lie between upper and lower bounds (boxed constraints, identified by inspection) that spanned all realistic values. Parameter values were qualitatively considered unrealistic when, during loading, the head flexed/extended through the torso, or when different head kinematic patterns (such as
higher harmonics being excited) than were observed experimentally were produced. Identifying robust parameter ranges for the lamb experimental model is a limitation of this study. The initial parameter estimates were obtained by manual adjustment about the quasi-static experimental estimates for Lamb 4 described in Chapter 4. These provided a representative example of model parameter values, but because the form of loading is significantly different to that described in Chapter 4, different parameters were required. After parameter estimates were obtained using the single shaking period, optimal simulation results were qualitatively compared across the representative steady-state time sample to assess the performance of the fitted model.

Additional metrics were used to assess the identifiability of the model parameters. Within the neighbourhood of the optimal parameter set (the indifference region), the Hessian curvature matrix \( (H) \) was estimated and the D-optimality criterion \( (det(H)) \), condition number \( (cond(H)) \) and M-optimality criterion \( (det(\bar{H})) \) were evaluated. The D-optimality criterion is a metric that is related to the volume of the indifference region and is ideally maximised for a well behaved objective function. The condition number is the ratio of the shortest and longest lengths of the indifference region’s eigenvectors. A value of 0 indicates that the objective function is ill-conditioned, and that the objective function is anisotropic in different parameter directions. The M-optimality criterion is the determinant of the normalised Hessian matrix (Equation 5.3), and is a number less than or equal to 1. The \( det(\bar{H}) \) describes any parameter interaction, and a value of 1 indicates that there are no interactions between model parameters (Gamage et al., 2011; Lanir et al., 1996; Nathanson & Saidel, 1985).

\[
\bar{H} = \frac{H_{ij}}{\sqrt{H_{ii}H_{jj}}} \quad (i, j, \text{not summed})
\]

5.4.2 Optimisation results

The experimentally-measured acceleration magnitudes were compared to those predicted by the model (Figure 5.8-Figure 5.11). Good agreement was observed in the
fitted regions for Lamb5 (Figure 5.10), and Lamb6 (Figure 5.11), and the accelerations were well matched beyond the fitted region. Although there were small changes in the timing of the acceleration peaks between the experimentally-measured and model-predicted results, the magnitude of the peak accelerations were well reproduced. Modest agreement was observed for Lamb2 (Figure 5.9), but the peak accelerations during extension contact were consistently underestimated throughout the steady-state time sample. Lamb1 showed modest agreement (Figure 5.8) in the fitted region, but poor agreement in the remaining time sample, although the results showed similar characteristics to the experimental data. Although only modest agreement was achieved between the model-predicted and experimental acceleration magnitudes, reasonable agreement between the model-predicted and experimental sagittal plane kinematic components for Lamb1 was still achieved (Figure 5.12). Good agreement between the sagittal plane kinematics for the other lambs was also observed (Appendix A.3)

![Graph](image-url)

**Figure 5.8** Comparison of model simulated acceleration magnitude to experimental measurements for Lamb1. The data in the shaded region was used to fit the model parameters, and the same parameter set used to simulate the kinematics beyond the fitted region. There is modest agreement in the fitted region but poor agreement in the rest of the representative time sample.
Figure 5.9 Comparison of model simulated acceleration magnitude to experimental measurements for Lamb2. The data in the shaded region was used to fit the model parameters, and the same parameter set used to simulate the kinematics beyond the fitted region. Modest agreement was observed throughout the time sample but the magnitude of the acceleration peaks are poorly reproduced in the model simulation.

Figure 5.10 Comparison of model simulated acceleration magnitude to experimental measurements for Lamb5. The data in the shaded region was used to fit the model parameters, and the same parameter set used to simulate the kinematics beyond the fitted region. Good agreement was observed throughout the representative time sample.
Significant variation was observed in the estimated model parameters (Table 5.4) and joint loading properties (Figure 5.13). This may be attributed to interactions occurring
between model parameters, as indicated by the M-optimality criterion being close to zero. The optimisation problem was also ill-conditioned, as indicated by the condition number of the Hessian matrix being small. Figure 5.13 demonstrates that the joint loading properties were stiffer in flexion than extension for all animals and the joints were measured to rotate further during extension than flexion during shaking (Figure 5.14). The D-optimality condition was large in each optimisation, which indicated that the objective function was sensitive to the model parameter values. Note that the magnitude of the D-optimality condition is problem specific and the relative values should be considered. The \( \text{det}(H) \) was largest for Lamb5 and Lamb6 which indicated a better behaved objective function when compared to the other cases. This may explain why the simulations for Lamb5 and Lamb6 demonstrated better agreement to the experimental data.

The contact positions (\( \text{Contact}_{\text{flex}} \) and \( \text{Contact}_{\text{ext}} \)) were defined with respect to the torso mesh used in the model. These positions were used in the optimisation, but do not consider the position of the shaker’s hands, which was observed to vary slightly between experiments. The large variation in the \( \text{Contact}_{\text{flex}} \) parameter may have been because of this hand position variation. The variation in the contact positions was also because flexion and extension contact was not observed in each experiment (Figure 5.4). For those experiments for which contact did not occur, the value of the contact parameters cannot be identified during the optimisation.

<table>
<thead>
<tr>
<th>Name</th>
<th>Description</th>
<th>Lamb1</th>
<th>Lamb2</th>
<th>Lamb5</th>
<th>Lamb6</th>
</tr>
</thead>
<tbody>
<tr>
<td>( c )</td>
<td>Viscous damping (N( \cdot )m( \cdot )rad(^{-1})( \cdot )s)</td>
<td>0.18</td>
<td>0.18</td>
<td>0.26</td>
<td>0.20</td>
</tr>
<tr>
<td>( m )</td>
<td>Head mass (kg)</td>
<td>0.92</td>
<td>0.60</td>
<td>0.59</td>
<td>0.71</td>
</tr>
<tr>
<td>( A )</td>
<td>Loading gradient coefficient (N( \cdot )m)</td>
<td>0.29</td>
<td>0.06</td>
<td>0.26</td>
<td>0.24</td>
</tr>
<tr>
<td>( k_f )</td>
<td>Flexion joint loading coefficient (rad(^{-1}))</td>
<td>19.72</td>
<td>18.75</td>
<td>15.79</td>
<td>15.26</td>
</tr>
<tr>
<td>( k_e )</td>
<td>Extension joint loading coefficient (rad(^{-1}))</td>
<td>13.71</td>
<td>13.69</td>
<td>5.88</td>
<td>9.73</td>
</tr>
<tr>
<td>( K_{\text{cont}} )</td>
<td>Contact stiffness (MPa)</td>
<td>0.21</td>
<td>0.20</td>
<td>0.20</td>
<td>0.15</td>
</tr>
<tr>
<td>( \text{Contact}_{\text{flex}} )</td>
<td>Flexion contact position (rad)</td>
<td>-0.18</td>
<td>-0.41</td>
<td>-0.11</td>
<td>0.47</td>
</tr>
<tr>
<td>( \text{Contact}_{\text{ext}} )</td>
<td>Extension contact position (m)</td>
<td>-0.027</td>
<td>-0.026</td>
<td>-0.032</td>
<td>-0.024</td>
</tr>
<tr>
<td>( \text{det}(H) )</td>
<td>D-optimality criterion</td>
<td>3.9x10(^{-6})</td>
<td>6.5x10(^{-4})</td>
<td>1.8x10(^{-7})</td>
<td>6.1x10(^{-7})</td>
</tr>
<tr>
<td>( \text{cond}(H) )</td>
<td>Condition number</td>
<td>3.7x10(^{-9})</td>
<td>1.5x10(^{-9})</td>
<td>1.1x10(^{-9})</td>
<td>2.1x10(^{-9})</td>
</tr>
<tr>
<td>( \text{det}(H) )</td>
<td>M-optimality criterion</td>
<td>1.0x10(^{-7})</td>
<td>3.8x10(^{-9})</td>
<td>1.2x10(^{-5})</td>
<td>3.6x10(^{-3})</td>
</tr>
</tbody>
</table>
There were noticeable differences from the quasi-static parameter estimates described in Chapter 4, with much higher estimates of the head mass and viscous damping being observed. The mode of loading during shaking is highly dynamic and it is likely that the damping is nonlinearly dependent on strain rate, which may account for the difference in the parameter estimates obtained from the quasi-static and shaking experiments. This may be explained by the model parameters changing to compensate for the viscous effects. For example, increased impact stiffness due to high contact decelerations could be modelled.

Figure 5.13 Joint loading properties estimated for each animal. In all cases, the joints were observed to be stiffer during flexion than extension, and each joint demonstrated different loading properties.

Figure 5.14 Representative joint angles from Lamb6 during shaking. Joint angles between T1 and T2 vertebrae, c5 and c6 vertebrae, and C2 and c3 vertebrae are illustrated. More rotation was observed in extension (positive joint angle) than flexion (negative joint angle).
by a higher contact stiffness parameter, and the stiffer vertebral joints during high velocity motion could be described by different joint loading parameters. Other viscous factors such as stress relaxation and creep may also contribute to the loading during the quasi-static experiments, but were unlikely to influence the measured kinematics during shaking. It was concluded that the mass estimated in the shaking experiments was likely to be a better estimate of the true head mass than the quasi-static estimates, since the model-predicted positions of the lambs’ heads did not align well with the measurements during the quasi-static simulations.

The identifiability of the model describes how easily a unique set of model parameters can be identified from the experiments. These results demonstrated poor parameter identifiability (small condition number) but the simulated head kinematics appeared realistic. This indicates that there may be multiple parameter estimates that can produce the same kinematics, but as the parameters are specific to the lamb, it is more important to understand how the different parameters (within the rigid-body modelling framework) influence the kinematics, rather than how the different parameters influence the quality of the fit. This sensitivity is investigated and discussed in the following section.

5.5 Model sensitivity

The model parameters estimated using these experiments are representative of lambs and may not be useful in describing the shaking kinematics in a human infant. The parameters do, however, provide realistic estimates of a lamb’s head motion, and can be used as a reference for a perturbation analysis to investigate the influence of the rigid-body model parameters on the shaking kinematics. It was assumed that the acceleration peaks are of the most importance when investigating AHT, and the influence of different parameters on the peak flexion and extension acceleration magnitudes during contact was investigated. The sensitivity analysis was performed by varying one parameter at a time, whilst prescribing all other parameters to be their predicted values. A combined sensitivity analysis was not performed for the lamb experiments because the model parameter estimates were not used in any predictive capacity for the human infant models.
(Chapter 6), and it was unlikely to provide information to further address the clinical problem.

For each lamb model, the mass, damping, and joint loading parameters, were varied individually by ±20% of each parameter’s respective optimal value. The effect of each parameter upon the head acceleration is illustrated in Figure 5.15 and Figure 5.16, where the optimal simulation result is depicted by the black line. The simulation results using higher parameter values are in red, where parameter values that are 10% and 20% above the optimum are depicted by light and dark red, respectively. Similarly, the simulation results using lower parameter values are in blue, where parameter values that are 10% and 20% below the optimum are depicted by light and dark blue, respectively. The parameters describing the positions of the contact regions were varied throughout a range of realistic positions and the contact stiffness was varied to span across the elastic moduli ranges proposed for an infant’s chest (Tsai, et al., 2012). The influence of these parameters is illustrated in Figure 5.17 where, once again, the optimum simulation is in black, increasing parameter values in light and dark red, and decreasing parameter values in light and dark blue. The following figures are representative, and demonstrate the sensitivity of the head acceleration to changes in the model parameters for Lamb6 within the shaking cycle, which is indicated by the fitted region in the optimisations. Sensitivity results for the other lambs, and for shaking cycles beyond the fitted regions, are included in Appendix B.

Varying the head mass and the damping in the model had different effects. Figure 5.15 illustrates that the magnitude and timing of the flexion peaks changed by approximately 80 m·s⁻² and 0.05 seconds, respectively, when the damping was varied by ±20% (Figure 5.15(a)). However, when the mass alone was changed by ±20%, only a small change in the acceleration magnitude (approximately 80 m·s⁻²) was observed (Figure 5.15(b)). In flexion, the peaks increase with an increase in both damping and head mass. Increasing the mass also began to produce an increase in the extension peak magnitude which was not clearly evident when the damping was varied. Similar results
were observed in a shaking cycle beyond the fitted region for Lamb6 (Figure B.19). Lamb2 and Lamb5 also demonstrated similar trends for shaking cycles within and beyond the fitted regions (Figures B.7-B.8 and Figures B.13-B.14, respectively), although contact peaks during flexion were not always observed for Lamb5. The sensitivity results for Lamb1 were inconsistent with those observed for the other lambs (Figures B.1-B.2). This may be due to the accelerations not being well matched in the optimisations and the actual kinematics not being well reproduced.

![Image of acceleration magnitude simulated during perturbation analysis while varying damping (a) and mass (b) in the model. Refer to text above for description of the parameter variation associated with each line colour. Results illustrate that the damping parameter influences the magnitude and timing of acceleration peaks and the mass influences only the magnitude.](image)

Figure 5.15 Acceleration magnitude simulated during perturbation analysis while varying damping (a) and mass (b) in the model. Refer to text above for description of the parameter variation associated with each line colour. Results illustrate that the damping parameter influences the magnitude and timing of acceleration peaks and the mass influences only the magnitude.

The simulated accelerations were also dependent on the loading characteristics of the joints in the model (Figure 5.16). It was found that the kinematics were influenced more by $k_e$, implying a stronger dependence on the joint properties during extension (Figure 5.16(b)). Varying $k_f$ affected the magnitude of the flexion peak, but had little influence on the remaining kinematics (Figure 5.16(a)). Varying $k_e$, however, resulted in magnitude and timing changes of both the flexion and extension peaks, in addition to significant changes to the non-peak accelerations. A decrease in $k_f$ and $k_e$ both result in larger accelerations. Similar results were also observed in shaking cycles beyond the
fitted region for Lamb6 (Figure B.20). For Lamb5, very similar trends were observed (Figures B.15-B.16), with the accelerations depending more strongly upon $k_e$, and this is discussed in Appendix B. The accelerations for Lamb2 demonstrated a similar dependence on the joint properties during flexion and extension (Figures B.9-B.10), but again, the accelerations were observed to increase as $k_f$ and $k_e$ was decreased. The sensitivity of the accelerations to the joint loading properties for Lamb1 were also inconsistent with the other lambs (Figures B.3-B.4).

The other model parameters that were considered were those describing the contact mechanics in the model (Figure 5.17). The model appears to be sensitive to the contact stiffness with large changes in the peak acceleration occurring when $K_{cont}$ was varied from 20 kPa to 5 MPa (Figure 5.17(c)). The model was also highly sensitive to the location of the contact surfaces. Moving the flexion contact position towards the head (i.e. to limit the range of flexion) increased the flexion acceleration peak considerably and had a small influence on the timing of the peaks (Figure 5.17(a)). Adjusting the extension contact location also had a marked effect on the peaks. Large increases in both peaks.
were observed when the contact position was moved anteriorly from the spine (i.e. to limit the range of extension) (Figure 5.17(b)). As the extension position was moved anteriorly, contact of the head during extension was initiated, which was not observed in the optimal fit. When contact was initiated, the magnitude of the acceleration peaks was comparable to those occurring in flexion. Similar results were observed beyond the fitted region for Lamb6 (Figure B.21) and for Lamb2 (Figures B.11-B.12). The sensitivity trends were also similar for Lamb5, however the dominant peaks were observed to occur
during extension, rather than flexion (Figures B.17-B.18). Once again the sensitivity results appear different for Lamb1 (Figures B.5-B.6).

5.6 Discussion

This study has described a computational modelling framework that is suitable for describing head kinematics during in vivo shaking of an animal model of AHT. The experimental identification of model parameters reproduced animal head kinematics and allowed a model-based interpretation of the experimental results. The computational model was shown to be able to reproduce features of the head accelerations that occur during contact of the head with the torso and demonstrated that these were the dominant accelerations of the animals head during shaking. The ability to reproduce head kinematics that include the high acceleration contact justifies using the coupled rigid-body modelling framework to reproduce in vivo shaking kinematics in human infants.

The experimental constraints imposed by the clinical study (a single manual shake) necessitated a number of modelling assumptions, and compromised some of the potential modelling inferences that could have been made if more rigorous loading regimes were used. This model-based investigation of head kinematics required a rigid-body model of a lamb to be developed with any model parameters being assumed, or inferred from fitting model simulations to experimental data. As the measured animal shaking motion was predominantly within the sagittal plane, there was limited information available to determine any out-of-plane mechanical properties. A result of this was that single rotational DOF joints were used, and it was assumed that joint flexion and extension would be sufficient to reproduce the majority of the experimental head kinematics. It was also assumed that additional DOFs would have a negligible effect on the head kinematics in a rigid-body model and, as spinal column injuries are atypical in AHT, any axial stretch was ignored. Upper spinal cord and brainstem injuries have been observed (Shannon et al., 1998), and have previously been attributed to spinal hyperflexion and extension, but these injuries cannot be explained by this model.
Although the model provides no information about any out-of-plane motion, the shaking inputs simulated in this study are consistent with those thought to occur in AHT (Cory & Jones, 2003; Duhaime et al., 1987; Prange et al., 2003), and indicate that major kinematic components were considered. A further simplification was that each joint was prescribed the same joint loading properties. The atlantoaxial-occipital joint complex at the top of the cervical spine is anatomically and functionally very different to the rest of the cervical spine but, with measurements being made only on the head and torso, there was no way to differentiate the motion of individual joints and the joints were treated homogenously. A further limitation is the use of a scaled generic model to create the computational models for each lamb. Scaling is a common technique used in musculoskeletal modelling to account for subject variability but it cannot account for specific oddities that a particular subject may have. It was necessary to scale a generic model rather than develop a model specific to the experiments because there was insufficient resolution in the experimental MRI scans to accurately describe the joints. Despite these limitations, the model was able to reproduce the head acceleration magnitudes and sagittal plane shaking kinematics reasonably well and therefore substantiates the suitability of using similar models to describe shaking kinematics in human infants.

Of the four lambs, Lamb1 showed the least agreement between the simulated and experimental acceleration magnitude. One reason for this was that there were larger out-of-plane acceleration components than the other experiments (Table 5.1), and accelerations were measured that exceeded the 160 m·s\(^{-2}\) saturation level of the accelerometer. Considering the limitations of this data, modest agreements to the sagittal plane kinematic components were still observed (Figure 5.12). For other animals, the quality of the experimental data was higher and very good agreement between the simulated and experimental planar kinematic components was observed (Appendix A.3).

It was identified that the large acceleration peaks were a result of contact of the head with the back of the lamb or with the shaker’s hands. These contact accelerations could
not be described using the joint properties alone but needed contact elements to be included in the OpenSim model. OpenSim provides two choices for contact modelling (Sherman et al., 2011); the Hunt-Crossley model and the Elastic Foundation model. The simpler Hunt-Crossley model was used in this study as there was insufficient experimental information to warrant the choice of a more sophisticated model. The contact stiffness parameter, $K_{\text{cont}}$, describes the combined composite elastic modulus of two interacting spheres. The parameter estimate of $K_{\text{cont}}$ was estimated to be 0.2 MPa, and corresponds to a Young’s modulus of 0.17 MPa. This was low, but within the bounds suggested by Tsai et al. (2012), who identified a potential range of Young’s moduli for an infant’s chest and ribcage based on the values in the literature. The low parameter value can be attributed to the redundancy of the contact model parameters. For example, there is an interaction between the contact stiffness parameter and the location of the contact spheres. The spheres could be positioned nearer to the contact position to compensate for a lower contact stiffness parameter or, for a high contact stiffness parameter, the contact spheres could be positioned further from the contact position to compensate. The quality of the initial estimate was limited by contact not being observed in each experiment, and this may be improved by a richer set of experimental motions being applied. As a result, more accurate estimates of the apparent contact stiffness are likely to be obtained.

Another contact parameter was the effective dissipation coefficient within the contact model. This is defined to be the slope of the coefficient of restitution versus impact velocity curve and was defined to be 0.5 s$^{-1}$ using estimates described by Hunt & Crossley (1975). The contact position, $K_{\text{cont}}$, and the dissipation coefficient are all related, and a richer variety of loading regimes would be necessary to improve parameter identifiability. Although more information is necessary, features of the contact accelerations were able to be reproduced and the results support the use of a Hunt-

\[ E^* = E(1-\nu)^2 \]

where $E^*$ is the plane strain modulus, $E$ is the Young’s modulus and $\nu$ is Poisson’s ratio. In this study, the Poisson’s ratio was assumed to be 0.4 (Tsai et al., 2012).
Crossley contact model to describe contact accelerations during the shaking of human infants.

The position of contact was observed to change slightly with each shaking cycle, and was indicated by the temporal changes of the flexion peaks in Figure 5.11. This all-or-nothing contact response resulted in discontinuities in the objective function that the optimiser was unable to deal with using gradient based optimisation techniques. In addition to the contact position complications, there were other factors that made the optimisation difficult. The problem was numerically stiff, which was caused, in part, by the contact, but was predominantly due to the small mass and inertia of the vertebrae in the neck. This resulted in the Runge-Kutta-Merson integrator in OpenSim performing poorly, and instead the CPodes implicit integrator was used (Sherman et al., 2011). There were also a large number of local minima in the objective function due to the parameter interactions that caused the optimisation results to be highly dependent on the initial parameter estimates. Because of these complications, it was decided to fit to a single shake period of the representative time sample (to remove the problem of contact positions changing), and to use a coarse finite differencing step size in the optimisation (to overcome any deficiencies in integration convergence). These choices were made at the expense of confidence in the parameter estimates as ideally all of the available data would be used to form the objective function. As the parameters in these models are specific to a lamb, and were not used in any predictive capacity in subsequent infant models, inaccurate optimisation results may not be a significant limitation. It was deemed more important to investigate the influence of the model parameters on the peak angular acceleration rather than determine the effect of each parameter on the quality of the fit. By perturbing the model parameters about their optimal estimates, the sensitivity of parameters within the coupled rigid-body modelling framework were described.

The results of this study demonstrated that variations in the model parameters influenced both the timing and magnitude of the contact acceleration peaks (Figure 5.15-Figure 5.17). A limitation of the sensitivity analysis in this chapter was that the lamb
models and the input shaking boundary conditions were not independent. When the parameters in the model were varied, the model’s frequency response also changed. It is likely that this would result in a different input shaking motion being imposed on the lamb. Because the shaking boundary conditions in this model were measured and prescribed, varying the model parameters alone may not provide a realistic reflection of the model parameter sensitivities. This limitation was avoided in the sensitivity analysis performed in Chapter 6, by allowing the input boundary conditions to also be varied. Nevertheless, by varying the model parameters within a narrow range, some insights into the modelling framework can be attained.

The most sensitive parameters were observed to be those describing the extension joint loading properties and those describing the contact conditions between the lamb’s head and its torso. The importance of the extension properties can be attributed to the joints rotating further in extension than flexion (Figure 5.14). As the joints are rotated further, the joint properties move to a stiffer region on the exponential loading curve, and contribute more to the mechanical stiffness of the neck. The maximum angle that the joints can achieve is influenced by the anatomy. The lambs’ necks are relatively long and they can extend substantially, while flexion is limited by contact of the head with the chest (or shaker’s hands). As the flexion angle of the joints is limited to a more compliant region of the joint loading curve, the influence of the $k_f$ parameter is significantly less than that of $k_e$ for Lamb5 and Lamb6, but it is likely that this behaviour is specific to the unique anatomy of the lambs. A similar dependence on $k_f$ and $k_e$ for Lamb2 may be explained by the way the animal was held, which may have altered the magnitude of joint rotation in each direction. The different sensitivity results observed for Lamb1 may be a result of joint loading properties in the model being stiffer than in the other lambs (Figure 5.13). A stiffer neck may produce joint forces that prevent full contact between the head and torso from occurring. This is likely to change how the model behaves in response to the parameter perturbations. Investigating the range of motion of an infant’s neck may
inform whether flexion or extension parameters are likely to be more important when describing infant head kinematics in AHT.

The acceleration caused during full flexion and full extension is very sensitive to the position where contact occurs. Although contact was not always achieved in the simulations, an 8 cm adjustment of the extension contact position allowed acceleration peaks to be modulated from no contact (no contact acceleration peak was observed), to large peaks that exceeded 200 m·s⁻², which were greater than the magnitudes measured experimentally. A similar response was observed when varying the flexion contact position. Once again, the effect on the peak timing was more pronounced when the extension contact position was varied as the joint angles where this occurs were being varied in a stiffer region of the joint loading curve. The contact stiffness, $K_{cont}$, had a large influence on the peak amplitude with an 450 m·s⁻² increase in the acceleration magnitude when the stiffness was increased from 0.02 MPa and 5 MPa. However, there was little influence on the between-peak kinematics and similar kinematics could be produced with different values of $K_{cont}$, by also varying the contact position. When simulating shaking kinematics in an infant, it will be important to prescribe the location of any contact surfaces carefully because of their substantial influence on the acceleration peaks, which are thought to contribute to injury.

Varying the head mass in the model had a relatively small influence on the kinematics in comparison to the other model parameters. In Lamb 6, for example, the peak acceleration changed by only 40 m·s⁻² for a 300 g change in head mass, with no discernable phase difference observed. It was expected that varying the head mass would have a similar influence as the damping on the frequency response of the model. It was unclear why only a small increase in the peak accelerations was observed when the mass was increased, but there was no noticeable change in the timing of the peaks. The increase in peak magnitude was attributed to the inertial forces being sufficient to overcome the joint stiffness so that contact of the head and torso could be initiated. In an infant, the head mass is significantly greater than that of a lamb. This is likely to produce
larger inertial forces during shaking. The inertial forces and the joint loading forces throughout shaking will determine whether contact of a human infant’s head and torso is initiated, and thus whether large contact accelerations will occur.

Another important feature to recognise is that the influences of different model parameters are interdependent as indicated by the M-optimality criterion being close to 0. An example of this is the parameters $A$ and $k_f/k_e$, which are related through the exponential joint loading relationship (Equation 5.1). In order to approximate a similar joint loading function, one could increase $A$ and at the same time decrease $k_f/k_e$. The magnitude and timing of the acceleration peaks are also dependent on a combination of parameters. For instance, the joint loading parameters describe the between-peak kinematics and it is this motion that determines whether the joints rotate enough for contact to occur. Once contact is initiated, the contact parameters are critical in describing the maximum accelerations that occur. These coupled and interacting effects are one of the reasons that it was difficult to identify a unique set of model parameters using the experimental measurements that were available. A limitation of the sensitivity analysis that was performed was that no coupled effects were considered, and only the individual parameter sensitivities were assessed.

In addition to the model-based interpretation of the head kinematics, the experiments included a clinical assessment of brain injuries in the animals. Existing studies have shown that lambs exhibit signs of traumatic brain injury when exposed to repeated shaking events (Finnie, 2012; Finnie et al., 2012, 2010; Sandoz et al., 2012), but it was of interest whether the single twenty second shaking event could cause the large macroscopic injuries clinically observed in infants. To assess these injuries, post shake MRI scans were performed between four and six hours after the shaking incident and were then reviewed by a neuro-radiologist. No significant parenchymal changes or extra-axial haemorrhages were observed in any of the lambs (n=6). Although each animal appeared stable throughout the observational period, five out of six animals died, or had to be prematurely euthanized, between four and six hours after the shaking episode. The
the heaviest lamb (Lamb6 at 8.8 kg) was the one animal to remain stable until the conclusion of the post-shake MRI (MRI for other animals was performed post-mortem). Similar outcomes were reported by (Finnie et al., 2012) where a number of the lighter lambs died in similar circumstances after shaking with similar accelerations to those measured in this study (Sandoz et al., 2012). Our in vivo results appear to corroborate the observation of (Finnie et al., 2012) that shaking alone can be fatal for a lamb, although other possible contributing factors (such as the anaesthetic protocol used) cannot be excluded, and subsequent histological analyses will provide more evidence to address this. As neither brain damage nor retinal haemorrhage was observed, there is doubt about the validity of the lamb being used as an animal model of AHT. Nevertheless, the in vivo shaking experiments allowed the validation of the biomechanics modelling framework to be performed. The framework described in this study is independent of the experimental model used and has the potential to be targeted towards more appropriate animal models to complement subsequent research into AHT.

5.7 Summary

This chapter has presented a coupled rigid-body computational model of a lamb that is able to reproduce head kinematics during in vivo shaking experiments, and is able to describe the complexities associated with head and torso impact. The modelling framework was validated using biomechanical measurements of the head kinematics that were analysed using computational model-based techniques. It was concluded that the large acceleration peaks measured were attributed to contact occurring between the lamb’s head and torso, but locations of contact are highly specific to the animal model. Although a clinical assessment of the brain provided no evidence of macroscopic bleeding in lambs after a shaking episode, the computational results support approaches to apply the modelling framework to describe infant head kinematics during shaking, and will complement subsequent research investigating the mechanisms of injury in AHT.
Infant shaking simulations

Aspects of this chapter are currently under review in the Journal of Biomechanics.

In previous chapters, the use of a coupled rigid-body modelling framework was shown to be able to reproduce shaking kinematics in a lamb, an animal experimental analogue of AHT. To apply this methodology to the clinical problem, it is important to create an equivalent computational model of the human infant. To develop such a model, anatomical data were obtained using computed tomography (CT) scans, material properties were taken from the literature, and shaking boundary conditions were estimated using existing computational models and anthropometric test dummy (ATD) experimental data. Using this computational model, a probabilistic analysis was performed from which ranges of infant head kinematics were produced, and the probability of exceeding current injury thresholds was analysed. The shaking kinematics were constrained by realistic muscle forces and worst-case shaking scenarios were investigated for different types of shaking motion, infant ages, and during shakes of different strengths.

6.1 Model development

6.1.1 Anatomy from CT

A computational model of a 4.5 months human infant was developed using anonymised CT scans that were obtained from an existing clinical database at Starship Children’s Health in Auckland, New Zealand. The collection and use of images 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 wavefront (.obj) surface meshes were generated with the Surface
Rendering tool in OSIRIX (Rosset, et al., 2004), a medical imaging and DICOM viewing software. The relative positions and orientations of the each bone’s anatomical coordinate system were used to build the computational in OpenSim (Delp et al., 2007). The models are publically available at https://simtk.org/home/infant_headneck/.

Anatomical coordinate systems were assigned to each of the bones using landmarks, and the relative location and orientation of successive bones in the neck were identified using the methodology described in Chapter 4. The anatomical coordinate systems of all vertebrae excluding C1 were defined according to Wu et al. (2002) (Figure 6.1(b)):

- **y axis**: vector through the upper and lower endplates directed superiorly;
- **z axis**: vector directed laterally to the right between the left and right superior articular processes;
- **x axis**: cross product of the y and z axis directed anteriorly;
- **origin**: midpoint between the upper and lower endplates along the y axis.

For the C1 vertebra (Figure 6.1(a)):

- **y axis**: cross product of the z and x axis directed superiorly;
- **z axis**: vector directed laterally to the right between the left and right superior articular processes;
- **x axis**: vector directed anteriorly from the mean of the two posterior tips of the unfused vertebral arch towards the origin;
- **origin**: centroid of the surface mesh.

For the skull (Figure 6.1(c)):

- **y axis**: cross product of the z and x axis directed superiorly;
- **z axis**: vector between the two occipital condyles directed to the right;
- **x axis**: vector parallel to the Frankfort horizontal plane directed anteriorly;
- **origin**: centroid of the two occipital condyles.
The convention chosen for the head is the same as was proposed by Loyd et al. (2010), but with the axis labels changed to match the directions used for the vertebrae. A schematic describing these anatomical coordinate systems is included in Figure 6.1.

![Figure 6.1 Anatomical coordinate system for C1 (a), remaining vertebrae (b) and skull (c). The y axis can be seen to point superiorly, the x axis anteriorly and the z axis is shown to point laterally to the right.](image)

The joint frames were approximated using biomechanical convention (Wu et al., 2002), but were fixed with respect to the parent and child vertebrae. This was a simplifying assumption that neglected any joint centre movement during vertebral motion. The joint centre was defined to be halfway between the anatomical origins of the parent and child vertebrae along the y axes of the inferior parent bone, when in a neutral pose (Figure 6.2).

![Figure 6.2 Inferior parent vertebrae (c5) and the superior child vertebrae (c4) with the joint frame in between. The joint centre is halfway between the anatomical origin of the parent and child vertebrae.](image)
During motion, the child vertebra and its associated joint frame move relative to the parent joint frame. For pure flexion and extension, this movement occurs about the $z$ axis of the parent joint frame (Figure 6.3). The relative motion of the parent and child joint frames allowed joint forces, which are proportional to the deviation between joint frames, to be included in the model.

![Figure 6.3 OpenSim model in 3 representative configurations. Each joint in 2° flexion (a), neutral (b) and each joint in 8° extension (c). The parent joint frames can be seen in each model (model available at https://simtk.org/home/infant_headneck/).](image)

### 6.1.2 Material properties

#### 6.1.2.1 Head mass and inertia

The CT scans provided anatomical information about the joint configurations, but only limited information regarding the mass and inertial properties of each bone. To describe the mass and moment of inertia (MOI) of an infant’s head, logarithmic age-dependent regression models of the head mass ($m_{sk}$), the head centre of gravity, and the sagittal plane inertial component ($I_{zz}$) were used (Equations 6.2-6.3) (Loyd et al., 2010). These regression models were calculated using CT scans of 14 fresh-frozen un-embalmed post-mortem human subjects (PMHS) aged between 31 weeks-gestation and 16 years. Each regression model described the nonlinear relationship necessary to scale $m_{sk}$ and $I_{zz}$ of an average adult to an infant using the characteristic length ($CL$). The $CL$ was defined to be the sum of the age-dependent parameters, head circumference ($C$), head length...
\( CL = C + HL + W \)  \hspace{1cm} (Equation 6.1)

The \( CL \) calculated by Loyd et al. (2010) for a 4.5 months infant, and the corresponding \( m_{sk} \) and \( I_{zz} \), are listed in Table 6.1. The centre of gravity of the head in the model was prescribed to be 10 mm anterior, and 40 mm superior, of the anatomical origin, which is the same as was measured for a 5 months infant (Loyd et al., 2010).

\[
\begin{align*}
  m_{sk} &= 4.54 \left( \frac{CL}{0.923} \right)^3 \text{ kg} \\
  I_{zz} &= 0.1537CL^2 - 0.1652CL + 0.04516 \text{ kg} \cdot \text{m}^2
\end{align*}
\]

Table 6.1 Characteristic length (CL), head mass (\( m_{sk} \)) and moment of inertia (\( I_{zz} \)) calculated for a 4.5 months infant using nonlinear age regression relations (Loyd et al., 2010).

<table>
<thead>
<tr>
<th>( CL ) (m)</th>
<th>( m_{sk} ) (kg)</th>
<th>( I_{zz} ) (kg\cdotm(^2))</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.735±0.019</td>
<td>2.29±0.18</td>
<td>0.0068±0.0053</td>
</tr>
</tbody>
</table>

### 6.1.2.2 Vertebral mass and inertia

Logarithmic age regression models describing the mass and MOI of the paediatric cervical spine were described by Dibb (2011). In Dibb’s study, 14 PMHS with ages ranging from 24 days to 17 years were imaged with CT, and the mass and MOI of the C5 vertebra was measured in each. The age-dependent regression models for the C5 vertebral mass (\( m_{c5} \)) and inertial properties (\( I_{xx} \), \( I_{yy} \), and \( I_{zz} \), respectively) (Equations 6.4-6.7) provide scaling between the properties of the C5 vertebra in adults to children, where the parameter \( A \) describes the age of the infant in years.

\[
m_{c5} = 3.52A^{0.552} \text{ kg} \hspace{1cm} (Equation 6.4)
\]

\[
l_{xx} = 216.8A + 127.8 \text{ kg} \cdot \text{m}^2 \hspace{1cm} (Equation 6.5)
\]

\[
l_{yy} = 104.4A + 58.7 \text{ kg} \cdot \text{m}^2 \hspace{1cm} (Equation 6.6)
\]

\[
l_{zz} = 275.2 \cdot A + 212.9 \text{ kg} \cdot \text{m}^2 \hspace{1cm} (Equation 6.7)
\]
The scaling factors (SF) calculated for the C5 vertebra were then used to scale the adult properties (Dibb, 2011) for all the remaining vertebrae. The scale factors and the scaled properties for a 4.5 months infant are listed in Table 6.2 and Table 6.3.

Table 6.2 Scaling factors (SF) calculated from Dibb (2011) to scale the C5 properties of an adult to a 4.5 months infant. These scale factors were used to scale other vertebrae.

<table>
<thead>
<tr>
<th></th>
<th>( m_{c5} ) (kg)</th>
<th>Mass SF</th>
<th>C5 MOI (1x10^{-6} kg\cdot m^2)</th>
<th>MOI SF</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td>( I_{xx} )</td>
<td>( I_{yy} )</td>
</tr>
<tr>
<td>4.5 months</td>
<td>0.002</td>
<td>0.11</td>
<td>0.21</td>
<td>0.10</td>
</tr>
<tr>
<td>Adult</td>
<td>0.02</td>
<td>1.0</td>
<td>4.2</td>
<td>2.5</td>
</tr>
</tbody>
</table>

Table 6.3 Mass and MOI of the cervical spine scaled for a 4.5 months infant using the scale factors in Table 6.2.

<table>
<thead>
<tr>
<th></th>
<th>Mass (kg)</th>
<th>MOI (1x10^{-6} kg\cdot m^2)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>( I_{xx} )</td>
<td>( I_{yy} )</td>
</tr>
<tr>
<td>T2(^a)</td>
<td>0.003</td>
<td>0.35</td>
</tr>
<tr>
<td>T1(^a)</td>
<td>0.003</td>
<td>0.35</td>
</tr>
<tr>
<td>C7</td>
<td>0.003</td>
<td>0.35</td>
</tr>
<tr>
<td>C6</td>
<td>0.003</td>
<td>0.30</td>
</tr>
<tr>
<td>C5</td>
<td>0.002</td>
<td>0.21</td>
</tr>
<tr>
<td>C4</td>
<td>0.002</td>
<td>0.23</td>
</tr>
<tr>
<td>C3</td>
<td>0.003</td>
<td>0.28</td>
</tr>
<tr>
<td>C1/C2(^b)</td>
<td>0.006</td>
<td>1.03</td>
</tr>
</tbody>
</table>

\(^a\) Mass and inertial properties not described and have assumed to be equal to C7
\(^b\) C1 and C2 were calculated as a single rigid-body

These properties were calculated using the vertebrae in isolation, and therefore do not account for the surrounding soft tissues in the neck. It is expected that the actual mass and MOI of each vertebra would be larger than these values, and it was decided to scale the age-dependent estimates of the vertebral properties by an order of magnitude to account for the entire intact neck. The order of magnitude scaling was calculated by estimating the volume of the neck cross section and assuming the tissue had the same density as water.

A similar approach was described in section 4.2.2. To mitigate the issues associated with inaccurate vertebral properties, the inertial parameters were allowed to vary across an order of magnitude to ensure any variation is described.

Existing cadaveric studies (Luck et al., 2008; Nightingale et al., 1998, 2002) have described joint loading properties of the atlantoaxial-occipital joint complex as a single functional unit, and as a result, C1 and C2 were described together as a rigid-body in the model. The upper thoracic vertebrae (T1 and T2) were not included in this study, and to account for this, the properties of T1 and T2 were prescribed to be the same as C7. It was
not clear what frame of reference was used to describe the inertial tensor in this study, but as the inertial forces in the model are likely to be dominated by the mass of the head, a representative estimate from these results was chosen.

### 6.1.2.3 Joint loading characteristics

Due to the difficulty in obtaining paediatric cadaveric specimens, there are limited data concerning the mechanical properties of an infant’s cervical spine. There is some literature regarding the tensile properties (Dibb et al., 2013; Luck, 2012; Luck et al., 2008, 2013; Nuckley et al., 2005, 2013; Nuckley & Ching, 2006; Ouyang et al., 2005), but very little investigating the rotational properties (Luck, 2012; Nuckley et al., 2013; Ouyang et al., 2005). The variation between studies was partly associated with the preparation of the samples, but general trends indicate more tensile compliance between the skull and C2 than in the lower cervical spine, and that the loading response can be modelled using an exponential relationship. The tensile stiffness (a linear approximation of the joint loading curve) for a 4.5 months infant can be estimated to be between 0.01 kN·m⁻¹ and 0.02 kN·m⁻¹ for the upper cervical spine, and between 0.05 kN·m⁻¹ and 0.1 kN·m⁻¹ for the lower cervical spine (Luck et al., 2008).

Exponential loading relationships have been demonstrated to be suitable for the paediatric human spine (Luck, 2012; Nuckley et al., 2013; Ouyang et al., 2005). These experiments exhibited different loading responses during flexion and extension. Therefore, the loading response was described using different constitutive parameters for flexion versus extension. The torque (τ) within each joint was therefore defined to be exponentially related to the joint angle (Equation 6.8), where the constants $A_f$ and $A_e$ are gradient coefficients applied to each loading direction, $k_f$ and $k_e$ are the coefficients that control the mechanical response during flexion and extension, respectively, and $\theta$ is the joint angle.
It was observed that the spine was more compliant in flexion than extension in all of the joints, and that the joints become successively stiffer with age (Luck, 2012). The loading parameters measured for a 5 months infant during flexion and extension, and the estimated ranges of motion ($ROM_f$ and $ROM_e$, respectively) for low loads (±0.1 N·m), are presented in Table 6.4. The results are listed for the joint between the base of the skull and the C2 vertebra (C0-C2), for the joint between the C4 and C5 vertebrae (C4-C5), and for the joint between the C6 and C7 vertebrae (C6-C7). It was difficult to calculate age-dependent descriptions of the constants due to the interdependence of the parameters. Because of this, the experimental parameters for a 5 months infant were used in the model and only provide a representative example of potential loading parameters.

Table 6.4 Loading characteristics of a cadaveric neck for a 5 months infant in three regions of the cervical spine (adapted from Luck (2012)). Results demonstrate that the neck tends to be more compliant during flexion and in the upper C0-C2 joint as indicated by a larger range of motion.

<table>
<thead>
<tr>
<th></th>
<th>C0-C2</th>
<th></th>
<th>C4-C5</th>
<th></th>
<th>C6-C7</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>$k_x$ (rad$^{-1}$)</td>
<td>$A_e$ (N·m)</td>
<td>$ROM_e$ (°)</td>
<td>$k_f$ (rad$^{-1}$)</td>
<td>$A_f$ (N·m)</td>
<td>$ROM_f$ (°)</td>
<td></td>
</tr>
<tr>
<td>14.22</td>
<td>0.0004</td>
<td>22.2</td>
<td>6.3171</td>
<td>0.0036</td>
<td>30.6</td>
<td></td>
</tr>
<tr>
<td></td>
<td>14.08</td>
<td>6.8</td>
<td>23.20</td>
<td>0.0024</td>
<td>9.3</td>
<td></td>
</tr>
<tr>
<td>12.73</td>
<td>0.0277</td>
<td>6.9</td>
<td>23.87</td>
<td>0.0032</td>
<td>8.3</td>
<td></td>
</tr>
</tbody>
</table>

*The direction of flexion and extension has not been specified to remain independent of modelling convention

The low loading condition was necessary to ensure non-destructive tests could occur indicating that the neck in young infants is very compliant as significant joint rotation was still achieved. Additional studies (Nuckley et al., 2013; Ouyang et al., 2005) have exhibited similar loading characteristics, and with larger torques, but with subject ages outside those expected for AHT.

6.1.2.4 Loading rate dependence

The above loading characteristics were measured predominantly during quasi-static experiments and thus may not be appropriate when the rate of loading in the neck is very...
high, such as during shaking. In the model, a lumped joint loading response was used to
describe the individual effects of muscles, ligaments, intervertebral discs and other
surrounding soft tissues. It is well accepted that these soft tissues exhibit a viscoelastic
response whereby the force opposing movement of the materials increases with an
increased loading rate. This phenomenon has been demonstrated in whole cervical spine
functional units and in isolated ligaments that encapsulate the intervertebral discs
(Nuckley et al., 2005; Chang et al., 1992; Mattucci et al., 2012; Wang et al., 2000;
Yoganandan et al., 1989, 2001). The stiffness was shown to increase by a factor of two to
four when tensile loading was increased from 10 mm·s$^{-1}$ to 2500 mm·s$^{-1}$. As there is
significant variation in the joint loading properties between subjects in the sparse data
available, it was decided to not consider loading rate dependence in the model. It was
assumed that the inter-subject joint stiffness variability would be of a similar magnitude
to the effect associated with the loading rate. Any loading rate dependence would also be
offset by changes in the existing model parameters and thus be indirectly captured by the
model. For example, a high velocity impact event would be able to be described by the
contact stiffness model parameter being higher. Similarly, the high velocity stiffening of
the vertebral joints could be offset by different joint loading parameters. Nevertheless, not
including loading rate dependence within the model remains a limitation that should be
investigated in subsequent research.

### 6.1.2.5 Contact stiffness

It has been shown in the previous *in vivo* lamb shaking experiments (Chapter 5) that
contact of the head with the torso is responsible for large peaks in acceleration. Estimates
of the Young’s modulus that describe the contact mechanics within the model were
obtained using paediatric measurements of cranial bone, ribs and cartilage. Although
there was insufficient data to draw statistical conclusions, Margulies & Thibault (2000)
reported the Young’s modulus of cranial bone to be between 71.6 MPa and 3582.2 MPa,
for infants of ages ranging from 25 weeks gestation to 6 months. Similar measurements
were performed on paediatric ribs and cartilage (Mizuno et al., 2005; Pfefferle et al.,
2007) and were combined in a finite element modelling analysis performed by Tsai et al. (2012). The Young’s modulus of the ribs was estimated to be 463 MPa for a 4.5 months infant and the Young’s modulus of cartilaginous structures was estimated to be 2.03 MPa. Significant variation was reported in these measurements and it was important to ensure a range of stiffness properties were simulated. The apparent impact stiffness would vary depending on the location of impact and may be due to the flexing of the rib cage. This is different to the direct loading that was used to estimate the Young’s moduli and because of this, it was difficult to determine an accurate estimate of the contact stiffness. It was assumed that the contact stiffness is dominated by the most compliant structure. Thus, the properties used in the model were chosen to be representative of the cartilage measurements, but were allowed to vary across orders of magnitude to account for the large uncertainty and to describe the apparent contact stiffness of different contact mechanisms.

6.1.2.6 Summary of model parameters

As was the case when modelling the lamb shaking kinematics, the model was restricted to motion within the sagittal plane by including vertebral joints that had a single rotational degree of freedom. It was assumed that maximal shaking kinematics would occur during this mode of shaking. In the lamb, the largest accelerations were observed during contact of the head with the torso, and it is these sagittal plane impacts that are likely to elicit the most damage.

One limitation of this study is that the vertebral joints could not rotate out of the sagittal plane or extend in the axial direction. This means that the model is unable to capture any out-of-plane motion during contact, which may contribute to the shear loading on the brain and eyes, and is unable to describe changes to the contact position that may occur as a result of neck elongation. As the head kinematics were compared to simplified injury metrics (section 6.2), this was not deemed to be a significant limitation, but increasing the joint ranges of motion should be considered in future studies.
The joint between the skull and the C2 vertebra (C0-C2 joint) was fixed, and the atlantoaxial-occipital joint complex was modelled as a single functional unit (Nightingale et al., 1998, 2002). The joint loading was modelled using the exponential constitutive relation described by Luck (2012) for the C0-C2 joint. As there was little change in the behaviour in the rest of the neck, all remaining joints were modelled similarly and were assigned the properties described for the C6-C7 joint.

The mass and MOI of the head were modelled using data directly adapted from Loyd et al. (2010), with the MOI reported being used for each of $I_{xx}$, $I_{yy}$, and $I_{zz}$. Owing to a lack of clarity regarding the coordinate system used by Dibb (2011), representative mass and inertial values were chosen for both the C1/C2 body as well as the remaining vertebral bodies. These representative values were chosen to be the same order of magnitude as those described by Dibb (2011), but were then scaled by an order of magnitude to account for the intact neck. This scaling factor was calculated by estimating the volume of the neck cross section and assuming the tissue had the same density as water. A similar approach was described in section 4.2.2. To simplify the model, all vertebral bodies (excluding the C1/C2 body) were prescribed the same mass and inertial values. In subsequent analyses (section 6.3), inaccurate estimation of vertebral mass and inertial properties was found not to be a significant limitation. A summary of the inertial and joint loading model properties is listed in Table 6.5.

There is large variation between the biomechanical properties of infants and the properties described are a best estimate for a 4.5 months infant using the literature available. These properties provided a starting point for a biomechanical analysis to investigate an infant’s head kinematics during shaking.
Table 6.5 Inertial and joint loading properties used for the 4.5 months OpenSim model. These parameters are a representative example for the age range and will provide a basis for subsequent parametric analysis.

<table>
<thead>
<tr>
<th>Inertial properties</th>
<th>(I_{xx}) (kg)</th>
<th>(I_{yy}) (kg(\cdot)m(^2))</th>
<th>(I_{zz}) (kg(\cdot)m(^2))</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head</td>
<td>2.29</td>
<td>6800</td>
<td>6800</td>
</tr>
<tr>
<td>C1-C2</td>
<td>0.06</td>
<td>20.0</td>
<td>20.0</td>
</tr>
<tr>
<td>C3</td>
<td>0.03</td>
<td>7.0</td>
<td>7.0</td>
</tr>
<tr>
<td>C4</td>
<td>0.02</td>
<td>7.0</td>
<td>7.0</td>
</tr>
<tr>
<td>C5</td>
<td>0.02</td>
<td>7.0</td>
<td>7.0</td>
</tr>
<tr>
<td>C6</td>
<td>0.03</td>
<td>7.0</td>
<td>7.0</td>
</tr>
<tr>
<td>C7</td>
<td>0.03</td>
<td>7.0</td>
<td>7.0</td>
</tr>
<tr>
<td>T1</td>
<td>0.03</td>
<td>7.0</td>
<td>7.0</td>
</tr>
<tr>
<td>T2</td>
<td>0.03</td>
<td>7.0</td>
<td>7.0</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Joint loading properties</th>
<th>(k_e) (rad(^{-1}))</th>
<th>(A_e) (N(\cdot)m)</th>
<th>(k_f) (rad(^{-1}))</th>
<th>(A_f) (N(\cdot)m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>C0-C2</td>
<td>14.22</td>
<td>0.0004</td>
<td>6.3171</td>
<td>0.0036</td>
</tr>
<tr>
<td>C2-C3 to T1-T2</td>
<td>12.73</td>
<td>0.0277</td>
<td>23.87</td>
<td>0.0032</td>
</tr>
</tbody>
</table>

6.1.3 Shaking boundary conditions

A computational model of a 50th percentile adult male upper-body (Holzbaur et al., 2005; Steele et al., 2013) was included with the infant model to prescribe kinematic (displacement) boundary constraints on the infant. The boundary constraints produced realistic shaking motions in the sagittal plane by varying the elevation of the shoulder joint \(\theta_{elev}\), flexion of the elbow \(\theta_{flex}\), and the deviation of the wrist \(\theta_{dev}\). All joints were perturbed sinusoidally about the configuration in Figure 6.4 (Equations 6.9-6.11) where \(\theta_{elev}\), \(\theta_{flex}\) and \(\theta_{dev}\) denote the amplitude of the sinusoids that describe each joint angle, respectively. Other parameters used to describe the shake 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. Sinusoidal loading was used as an approximation of manual shaking events as similar input boundary conditions were observed during the lamb experiments (Figure 5.7). This parameterisation, however, would be unable to reproduce complex multi-modal shaking inputs, but the impacts of these shaking inputs were not considered in this study.
By varying $\Phi$, different types of shaking motions were able to be produced. When $\Phi = 0$ radians, any shoulder elevation is in phase with elbow flexion and motion occurs along an arc. Conversely, when $\Phi = \pi$ radians, the shoulder elevation and elbow flexion are out of phase and a more linear shaking profile is produced. 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.

\[
\theta_{elev} = \theta_{elev} \sin(2\pi f) \quad 6.9
\]

\[
\theta_{flex} = \theta_{flex} \sin(2\pi f + \Phi) \quad 6.10
\]

\[
\theta_{dev} = \theta_{dev} \sin(2\pi f + \Phi) \quad 6.11
\]

Appropriate shaking frequencies were taken from previous studies that involved the manual shaking of a 1.5 months infant surrogate (Coats & Margulies, 2008). The shaking frequency from five independent shaking experiments varied between 1.6 Hz and 2.5 Hz.

Figure 6.4 A sagittal plane OpenSim model of a 4.5 months human infant used to simulate head flexion and extension kinematics during AHT. The model contained a lumped description of joint torques to describe loading within the neck joints. An existing upper-body model (Holzbaur et al., 2005) was used to prescribe sinusoidal shaking motions that were constrained to be anatomically realistic (model available at https://simtk.org/home/infant_headneck/).
with an average of 2 Hz (Coats, 2011). The infant model was rigidly attached to the hands of the 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), and the peak angular acceleration and peak angular velocity of the infant’s head was measured.

### 6.2 Deterministic model of infant head kinematics

An infant model with the mean model parameter values listed in Table 6.6 was shaken using a representative set of shaking boundary conditions. The infant model was shaken at 2 Hz and the amplitudes of the limb segments were selected to produce two mid-range motions; one along an arc and the other that followed a more linear shaking profile.

The simulation results were compared with published injury criteria that has been used to assess the biomechanics of AHT, and to results from other surrogate and computational modelling experiments (Bondy et al., 2012; Cory & Jones, 2003; Duhaime et al., 1987; Wolfson et al., 2005). The injury criteria described the kinematics necessary to produce concussion and subdural haemorrhage (SDH). These were scaled from the injury criteria described by Duhaime et al. (1987) to the estimated brain mass of the infants used in this study (Appendix C).

The peak angular velocity and acceleration predicted by the model when shaken along an arc were 42 rad·s$^{-1}$ and 0.6 rad·s$^{-2}$, respectively (Figure 6.5). Similarly, the peak angular velocity and acceleration predicted by the model when shaken linearly were 27 rad·s$^{-1}$ and 0.38 rad·s$^{-2}$, respectively. The kinematics predicted by the arc shake were comparable to those measured by Cory & Jones (2003), but the angular acceleration was approximately 0.15 rad·s$^{-2}$ less than Cory & Jones’ minimum angular acceleration measurement. The kinematics predicted by the linear shake were comparable to the simulation results of Wolfson et al (2005), but with a slightly higher (approximately 0.1 rad·s$^{-2}$) peak angular acceleration.
Shaking along an arc resulted in a much larger peak angular velocity ($\omega_{\text{peak}}$) and a small increase in the peak angular acceleration ($\alpha_{\text{peak}}$) than when shaken linearly (Figure 6.5). Neither of the simulations produced kinematics that exceeded any of the concussion nor SDH injury thresholds. Large differences in the simulated kinematics were caused by only varying the shaking input, which suggests that further investigation into the effects of different shaking motions is necessary. It is important to understand the range of kinematics that can be produced and to evaluate whether worst-case shaking scenarios can produce kinematics that exceed any of the published injury thresholds.

Figure 6.5 The kinematics produced during a linear and an arc shake are consistent with others published in the literature. Results indicate that higher kinematics are produced during an arc shake and that there is significant variation in an infants head kinematics when the type of shaking motion is varied.
6.3 Probabilistic model of infant head kinematics

The simulated kinematics has been shown to vary significantly during different types of shaking, but there is also uncertainty in the model parameters that describe an infant’s mechanical properties (Dibb, 2011; Loyd et al., 2010; Nuckley et al., 2013; Ouyang et al., 2005; Tsai et al., 2012). To account for this variability, probabilistic methods were used to analyse the infant head kinematics during shaking by combining the OpenSim API with NESSUS (SwRI, San Antonio, TX, USA), a probabilistic analysis software package (Figure 6.6). The OpenSim forward simulations were initiated by NESSUS by prescribing a deterministic set of model parameters and calling the executable developed in the OpenSim API. The model parameters were sampled from a set of probability density functions (PDF) to obtain cumulative distribution functions (CDF) describing the peak angular acceleration and the peak angular velocity of the infant’s head during different shaking scenarios.

![Figure 6.6 Schematic describing the use of NESSUS to estimate the cumulative distribution function (CDF) of the output kinematics using a Monte Carlo analysis. Model parameters were sampled from estimated probability distributions.](image)

PDFs were prescribed for all parameters including those describing the mass properties, inertial properties, joint loading properties, contact properties, and the shaking
boundary conditions (Table 6.6). 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 chosen for the shaking frequency with the mean and standard deviation estimated from the average and the variance in the surrogate shaking experimental dataset (Coats, 2011). The lognormal 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 an order of magnitude and to ensure that the MOI was greater than zero. Uniform probability distributions between specified upper and lower limits were assumed for the shaking amplitudes and phase difference as there was no experimental justification for the use of any other PDFs.

The Monte Carlo probabilistic method was applied in this study as it ensures solution convergence as the number of simulation trials tends towards infinity (Haldar & Mahadevan, 2000). 3000 simulation trials were found to provide representative results for this study (Appendix D). In this method, each parameter was randomly sampled according to its PDF, and statistics describing the output kinematics were evaluated. 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 & 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.
Table 6.6 Model parameter probability distributions used in the probabilistic analysis. Parameter means ($\mu$) and standard deviations ($\sigma$) are listed for the different distribution functions.

<table>
<thead>
<tr>
<th>Name</th>
<th>Description</th>
<th>$\mu$</th>
<th>$\sigma$</th>
<th>Distribution</th>
</tr>
</thead>
<tbody>
<tr>
<td>$c$</td>
<td>Viscous damping (N·m·rad$^{-1}$·s$^a$)</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>$A_{f-C0-C2}$</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>$k_{f-C0-C2}$</td>
<td>Flexion C0-C2 joint loading coefficient (rad$^{-1}$)</td>
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<td>0.637</td>
<td>Normal</td>
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<tr>
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<td>Extension C0-C2 joint loading coefficient (N·m)</td>
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<td>0.000041</td>
<td>Normal</td>
</tr>
<tr>
<td>$k_{e-C0-C2}$</td>
<td>Extension C0-C2 joint loading coefficient (rad$^{-1}$)</td>
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</tr>
<tr>
<td>$K_{cont}$</td>
<td>Contact stiffness (MPa)</td>
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<td>13</td>
<td>Lognormal</td>
</tr>
<tr>
<td>$\alpha_{diss}$</td>
<td>Contact dissipation (s·m$^{-1}$)</td>
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<td>0.1</td>
<td>Normal</td>
</tr>
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<td>Head mass (kg)</td>
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<tr>
<td>$I_{head}$</td>
<td>Head MOI - sagittal plane (kg·m$^{-3}$)</td>
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<td>0.0014</td>
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</tr>
<tr>
<td>$m_{vert}$</td>
<td>Vertebral mass (kg)</td>
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<td>0.0003</td>
<td>Normal</td>
</tr>
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<td>$I_{vert}$</td>
<td>Vertebral MOI - sagittal plane (kg·m$^{-3}$)</td>
<td>7x$10^{-6}$</td>
<td>3.5x$10^{-6}$</td>
<td>Lognormal</td>
</tr>
<tr>
<td>$m_{vert-C0-C2}$</td>
<td>C0-C2 Vertebral mass (kg)</td>
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<td>0.006</td>
<td>Normal</td>
</tr>
<tr>
<td>$I_{vert-C0-C2}$</td>
<td>C0-C2 Vertebral MOI - sagittal plane (kg·m$^{-3}$)</td>
<td>20x$10^{-6}$</td>
<td>10x$10^{-6}$</td>
<td>Lognormal</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></td>
<td>Deterministic</td>
</tr>
</tbody>
</table>

* $a$ Estimated using lamb shaking experimental results (Lintern et al., 2013)
* $b$ Representative joint loading properties for a 5 months infant (Luck, 2012)
* $c$ Estimated using elastic moduli ranges for infant rib cage (Tsai et al., 2012)
* $d$ Estimated using ranges proposed by Hunt & Crossley (1975)
* $e$ Head mass and MOI for 4.5 months using age-regression model (Loyd et al., 2010)
* $f$ Vertebral mass and MOI for 4.5 months using age-regression model (Dibb, 2011)
* $g$ Anthropometric dummy shaking data (Coats, 2011)
* $h$ Torso mass obtained from 50th percentile male child growth charts (World Health Organization, 2006)

### 6.3.1 Variation of infant model and shaking parameters

The results of three different Monte Carlo simulations, each with 3000 trials ($N = 3000$), are shown in Figure 6.7. The model parameters describing the input shaking boundary conditions (referred to hereafter as the "shaking parameters") were sampled first from their probability distributions with the model parameters describing the infant material properties fixed to their mean values. Subsequently, two Monte Carlo simulations were performed whereby the shaking parameters were fixed to produce (i) a linear shake, and (ii) an arc shake (described in Figure 6.5), while the infant model
parameters were sampled from their probability distributions. Linear and arc shakes were produced by prescribing the arm segments to be out-of-phase ($\Phi = \pi$ radians), and in-phase ($\Phi = 0$ radians), respectively.

![Figure 6.7 Output kinematics produced from 3 Monte Carlo simulations (N = 3000) where the shaking and infant model parameters are varied independently. There is a large variation in the peak angular velocity and acceleration when the shaking parameters were varied (green) but only localised kinematics were observed when the infant model parameters were varied during a fixed linear (blue) or arc (red) shake. The scaled injury thresholds and mean results during the linear and arc shakes are as described in Figure 6.5.](image)

A large variation in the output kinematics was observed when the shaking parameters were varied, with the maximal kinematics observed to exceed the scaled SDH thresholds. Conversely, the output kinematics produced from varying the infant model parameters were very localised about their mean values. The CDFs for the angular acceleration and velocity (Figure 6.8) also demonstrate a higher sensitivity to the shaking input than to the infant model parameters. The CDFs describing the output kinematics when the shaking parameters were varied are broad, indicating that output kinematics span a wide range. The CDFs obtained when the infant model parameters were varied are much steeper, indicating that less variation was observed in the output kinematics. The lower mean
values observed during the linear shake are illustrated by a left shift of the corresponding CDFs.

![Figure 6.8 CDF’s for peak angular acceleration (a) and peak angular velocity (b) when input parameters are randomly sampled from their probability distributions. Variation in the shaking parameters resulted in a wide variation of the output kinematics, whereas varying the infant model parameters during a fixed shake resulted in more localised kinematics. The acceleration and velocity are both higher during an arc shake than during a linear shake.]

**6.3.2 Combined variation of all model parameters**

The results of an additional Monte Carlo simulation (N = 3000), where all parameters (infant model and shaking parameters) were sampled from their probability distributions, are illustrated in Figure 6.9. The envelope of output kinematics is wider than in Figure 6.7, and can be seen to encompass the variation of both sets of parameters. Although there is a difference between the peak angular velocity observed in Figure 6.7 and Figure 6.9, this would likely not be the case if more simulation trials were performed and the simulated kinematics were more densely sampled from the cumulative distribution functions of the models. Once again, the maximal kinematics exceed SDH injury thresholds, but the proportions of cases that exceeded the threshold values for angular acceleration and angular velocity were small (0.01 and 0.05, respectively) (Figure 6.10).
The sensitivity factors demonstrated that the variability of the output kinematics was affected most strongly by the variability in the shaking frequency and the contact stiffness (Figure 6.11). The sensitivities of the peak angular acceleration and velocity to the shaking frequency \( f \) were similar (0.651 and 0.654, respectively), but only the peak acceleration demonstrated any sensitivity to the contact stiffness (0.32). The amplitude of shaking \( \theta_{elev}, \theta_{flex}, \theta_{dev} \), and the head inertia \( l_{head} \), also had a modest impact upon the output kinematics. The relatively high sensitivity to the shaking parameters was also evident when Monte Carlo simulations were performed while independently varying parameters describing the mechanical properties of the infant and the shaking parameters (Figure 6.7). These results demonstrate that the output kinematics are highly sensitive to the type of shaking motion applied but are relatively insensitive to the model parameters describing the infant’s properties.

![Figure 6.9 Output kinematics produced from a Monte Carlo simulation (N = 3000) where all parameters were randomly sampled from their probability distributions. The maximal kinematics were observed to exceed injury thresholds described in Figure 6.5.](image)
Subdividing the results in Figure 6.9 according to frequency ($f$) and contact stiffness ($K_{cont}$) (Figure 6.12(a) and Figure 6.12(b)) also demonstrates the sensitivities of these parameters. An increase in shaking frequency resulted in an upwards shift through the output kinematics envelope, increasing both the peak angular acceleration and velocity produced by the model. The maximal kinematics, when the SDH injury threshold was exceeded, occurred when the shaking frequency exceeded 4.5 Hz, which was well outside any experimentally observed shaking frequency. Conversely, an increase in the contact stiffness yielded an increase in the peak angular acceleration but there was little change to
the peak angular velocity range. These results are expected as the peak angular acceleration has been shown during \textit{in vivo} lamb shaking experiments to occur when the head strikes the torso, and it is likely that similar contact occurs in the human infant. Although increasing $K_{\text{cont}}$ resulted in higher angular accelerations, there was not a large difference in the velocity distribution (Figure 6.13), indicating that variation in another parameter (likely to be shaking frequency) is needed to elicit the high peak angular velocities necessary to exceed published injury thresholds.

The amplitudes of the shaking sinusoids were also important with the simulated angular accelerations and velocities increasing as the shaking amplitude were increased. A reason for this is that the simulated angular accelerations and velocities appeared to be bimodal with the determining factor being whether or not the head passed through the neutral pose in phase with the shaking motion. At low shaking frequencies and amplitudes, the head had insufficient momentum to transition from full flexion to full extension (or vice versa) and lower peak angular accelerations and velocities were observed. When the shaking frequency and amplitude were increased, the peak angular velocity of the head was larger. The head also proceeded to impact the front and back of the torso periodically, producing much larger peak angular accelerations. During this mode, the magnitude of the output kinematics increased as the frequency and amplitude were increased. This characteristic is also likely to be responsible for the non-Gaussian shape of the CDF describing the peak angular acceleration, which was skewed towards high accelerations (Figure 6.10). Shaking along an arc was found to promote the periodic mode of head kinematics as the rotation of the arms and hands acted to assist the transition of the head through the neutral pose. This also acted to exacerbate the impact that occurred at full extension while the peak accelerations during a linear shake had similar magnitudes in both flexion and extension. Whether the peak kinematics occurred at full flexion or full flexion was not differentiated in this study, but it is likely that the loading response on the brain will be different depending on when the peak kinematics occur. The probability distributions used in the simulations resulted in head motion that
produced contact between the head and the torso, and resulted in the high accelerations observed. Those simulations that did not involve head/torso impact resulted in head kinematics that did not exceed the injury thresholds. Because of this, it is likely that head/torso impact events will be necessary for kinematics to exceed current estimates of injury thresholds in AHT.

Figure 6.12 Output kinematics from Figure 6.9 differentiated according to shaking frequency (f) and contact stiffness (K_{cont}). Higher shaking frequencies produced higher angular velocities and a right shift up through the injury criteria. An increase in contact stiffness demonstrates higher angular acceleration but no change to the angular velocity.

Figure 6.13 CDFs for peak angular acceleration (a) and peak angular velocity (b) calculated when K_{cont} is set to 15 MPa and 5 MPa. All other input parameters are randomly sampled from their probability distributions. Higher K_{cont} produced higher angular accelerations indicated by a right shift of the CDF but had little influence on the CDF describing the angular velocity.
6.3.3 Kinetic constraints on the shaking motion

It has been shown that head kinematics exceeding SDH injury thresholds can be produced during shakes that are kinematically constrained to be within an average adult’s natural range of motion, and that the head kinematics are highly dependent on the shaking frequency. When the frequency of a shake increases, the accelerations imposed upon the infant increases, and are proportional to the square of the frequency. To accelerate the mass of an infant at a higher frequency, the muscles in the arms of the shaker need to generate larger forces resulting in larger torques across the shoulder, elbow and wrist joints. Estimated joint torques for a given shaking motion can be compared to maximal joint torques to provide a realistic kinetic constraint upon the shaking motion.

The shoulder, elbow and wrist joint torques were estimated using inverse dynamics and were calculated to balance the inertial forces necessary to accelerate the infant and the shakers arms. In the model, the infant is rigidly attached to the shakers hands and the shaking motion is kinematically constrained. This results in any contact between the infant’s head and torso producing large torque spikes that propagate through the limb segments, and may exceed the maximum torques that can be generated by the muscles. In the physical situation, these torque spikes would be attenuated as the kinematic constraints are not applied. Instead, there are only kinetic loading conditions (via the shaker’s muscles) applied to the joints. This would prevent the oscillatory motion from occurring when the muscles can no longer produce the necessary force to continue the movement. Rather than adopting the computationally demanding approach of modelling the full activation dynamics and muscle activity necessary to produce the shaking motion, some simplifying assumptions were made. The inverse dynamics analysis was simplified to only consider 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. It is likely that this, and the smooth shaking motion that is prescribed, underestimates the peak torques that would occur at the joints.
The peak torques estimated using inverse dynamics were compared to the maximum isometric joint torques that were predicted by Holzbaur et al.'s (2005) upper body model. These joint torques demonstrated good agreement to experimental cadaveric data (Amis et al., 1980; Buchanan et al., 1998; Delp et al., 1996; Garner & Pandy, 2001; Otis et al., 1990; Winters & Kleweno, 1993). The maximum isometric joint torque provides an upper limit due to the inverse relationship between shortening velocity and force production in skeletal muscle. Although skeletal muscle can generate a larger force during an eccentric contraction than when contracting isometrically, it has been shown that the joint torque across the elbow, shoulder and other joints does not increase significantly above the peak isometric force (Amis et al., 1980; Babault et al., 2001; Hortobágyi & Katch, 1990; Westing et al., 1988). The maximum isometric joint torque limits that were used are listed in Table 6.7.

<table>
<thead>
<tr>
<th>Joint</th>
<th>Moment (N⋅m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shoulder flexion</td>
<td>80</td>
</tr>
<tr>
<td>Shoulder extension</td>
<td>120</td>
</tr>
<tr>
<td>Elbow flexion</td>
<td>100</td>
</tr>
<tr>
<td>Elbow extension</td>
<td>40</td>
</tr>
<tr>
<td>Wrist deviation (radial)</td>
<td>12.5</td>
</tr>
<tr>
<td>Wrist deviation (ulnar)</td>
<td>7</td>
</tr>
</tbody>
</table>

The peak isometric torque constraints were applied to three different Monte Carlo simulations that were targeted to different infant ages. In addition to the 4.5 months infant model described, the mass, inertial, and joint loading properties were scaled to a newborn and a 12 months infant (Table 6.8). The mass and inertial parameters were scaled using the same age-dependent models (Dibb, 2011; Loyd et al., 2010), and the loading properties estimated from representative experimental measurements (Luck, 2012). The flexion loading parameters were not scaled for the 12 months infant as the experimental data was observed to be within the joint loading variation prescribed for the 4.5 months infant. The torso mass \( m_{\text{torso}} \) was prescribed as a deterministic model parameter in these simulations and was obtained from human growth charts (World Health
The injury thresholds were also scaled to the respective age ranges (Appendix C).

Table 6.8 Age-scaling of probabilistic model parameters described in Table 6.6. The variable $m_{\text{torso}}$ was prescribed as a deterministic variable in these models. All parameters not listed were not scaled and were prescribed to be the same as for the 4.5 months infant.

<table>
<thead>
<tr>
<th>Name</th>
<th>Newborn</th>
<th>12 months</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>$k_{f-coc2}$ (rad$^{-1}$)</td>
<td>0.25</td>
<td>0.76</td>
</tr>
<tr>
<td>$k_{e-coc2}$ (rad$^{-1}$)</td>
<td>5.73</td>
<td>14.32</td>
</tr>
<tr>
<td>$m_{\text{head}}$ (kg)</td>
<td>1.0</td>
<td>2.85</td>
</tr>
<tr>
<td>$I_{\text{head}}$ (kg$\cdot$m$^2$)</td>
<td>0.00068</td>
<td>0.01</td>
</tr>
<tr>
<td>$m_{\text{vert}}$ (kg)</td>
<td>0.003</td>
<td>0.06</td>
</tr>
<tr>
<td>$I_{\text{vert}}$ (kg$\cdot$m$^2$)</td>
<td>4.67x10$^{-6}$</td>
<td>10.5x10$^{-6}$</td>
</tr>
<tr>
<td>$m_{\text{vert-coc2}}$ (kg)</td>
<td>0.006</td>
<td>0.12</td>
</tr>
<tr>
<td>$I_{\text{vert-coc2}}$ (kg$\cdot$m$^2$)</td>
<td>13.3x10$^{-6}$</td>
<td>30x10$^{-6}$</td>
</tr>
<tr>
<td>$m_{\text{torso}}$ (kg)</td>
<td>2.4</td>
<td>6.75</td>
</tr>
</tbody>
</table>

Figure 6.14 Peak angular accelerations ($\alpha_{\text{peak}}$) and angular velocities ($\omega_{\text{peak}}$) for the kinetically constrained shaking simulations for a newborn infant (a), a 4.5 months (b) and a 12 months (c). Significantly higher peak angular accelerations were observed for the newborn infant but little difference was observed between the 4.5 months and 12 months cases. Only small changes in the peak angular velocity were observed and no results exceeded scaled SDH injury thresholds.

The results using these kinetic constraints are presented in Figure 6.14. Many of the simulations in Figure 6.9 that produced the higher output kinematics required joint torques that exceeded what the muscles could generate under isometric conditions. The maximum peak angular velocity was limited to approximately 60 rad$\cdot$s$^{-1}$, 50 rad$\cdot$s$^{-1}$ and 45 rad$\cdot$s$^{-1}$ for the newborn, 4.5 months and 12 months infants, respectively, and was observed to scale with the joint torque limits (Appendix E). The change in the kinematics.
envelope was more pronounced for the newborn infant where the simulations produced much higher accelerations (up to 35000 rad s\(^{-2}\)), but the peak angular velocity was insufficient to exceed the scaled SDH injury thresholds. The results were similar for the other aged infants with no simulations producing kinematics that exceeded the SDH injury thresholds.

The total mass of the 4.5 months infant in these simulations was approximately 7.5 kg which was of a similar mass to the lambs that were shaken experimentally (Chapter 5). The peak angular velocities in the lamb experiments were measured between 36 rad s\(^{-1}\) and 40 rad s\(^{-1}\). This is consistent with the kinetically constrained simulation results. As illustrated in Figure 6.5, the finite element modelling results reported by Bondy et al. (2012) exceeded those simulated with the model of the 4.5 months infant model. This difference can be attributed to the fact that the material properties in the model were targeted to a newborn infant. The simulation results for the model of the newborn presented in Figure 6.14(a) encompass those described by Bondy et al. (2012), which indicates that the parameter scaling was appropriate.

A large increase in the peak angular acceleration was observed when the infant age was reduced, and can be attributed to the logarithmic scaling of the mass and inertial properties (Dibb, 2011; Loyd et al., 2010; World Health Organistation, 2006). A reduction in head mass resulted in smaller inertial forces, which increased the magnitude of deceleration of the head during impact. The importance of head inertia on the peak angular acceleration was also demonstrated by a relatively high sensitivity factor (Figure 6.11). The infant’s age, however, was observed to only have a small effect on the peak angular velocity, which is likely to be more strongly affected by the strength of the shaker.

For stronger shakers, it is likely that they have larger muscles, and a greater ability to generate force, which will increase the peak isometric joint torque limits. The torque constraint could be scaled appropriately as the peak isometric joint torque in the arms has
been shown to scale linearly with muscle volume (Holzbaur et al., 2007), and the variance in muscle volume was shown to account for up to 95% of the inter-subject variance of peak joint torques. Although limb length had little influence on a shaker’s strength, it is likely that it would influence the kinematics imposed upon the infant. To account for this, there are existing tools in OpenSim to scale limb segments using anatomical landmarks. If custom limb lengths were combined with custom muscle volumes, for example obtained using MRI (Holzbaur et al., 2007a), then kinetic constraints that are highly specific to a potential shaker could be estimated. It is worth noting a recent clinical series that concluded that the severity of the injuries was worse when the perpetrator was male, and hypothesised that the size and strength of male perpetrators is one possible explanation (Esernio-Jenssen, et al., 2011). By combining these customised joint torque constraints with an infant’s specific anatomical and inertial properties, a range of head kinematics that are highly incident specific could be simulated, allowing worst-case shaking scenarios to be investigated.

6.4 Summary

This chapter has described a rigid-body computational model of the human infant that was developed using anatomical data obtained from CT, and experimentally derived material properties from the literature. The model was used to simulate the peak head kinematics that occur during shaking. To account for the uncertainty in the material properties of the model, a probabilistic analysis was performed to predict worst-case shaking scenarios. These results demonstrated that the peak head kinematics are strongly sensitive to the type of shaking imposed upon the infant, and relatively insensitive to changes in the infant’s material properties. The shaking motions applied to the model were constrained by the peak isometric joint torques of the shaker’s arms. It was observed that for higher joint torque limits (representing a stronger shaker), greater head kinematics were produced. These motions were unable to produce head kinematics that exceeded published SDH injury thresholds. The results of this chapter 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, or that the current estimates of injury thresholds are incorrect. The limitations of this research will be described in the following chapter (section 7.3) and areas for future research in the field will be suggested.
Conclusions and perspectives

The objective of this thesis was to develop and validate a computational modelling framework to investigate the biomechanics of AHT. The framework, developed in OpenSim, was used to reproduce head kinematics during shaking of an animal experimental model, and to simulate an infant’s head kinematics during shaking, which were compared with published injury thresholds. This chapter summarises the conclusions of the experimental studies and the infant shaking simulations, and describes the major limitations and implications of this research.

7.1 Animal shaking experiments

A coupled rigid-body computational modelling framework was used to reproduce head kinematics during manual shaking experiments. The in vivo experiments, using the lamb as an animal analogue of AHT, necessitated the development of an OpenSim rigid-body computational model, the implementation of a novel model parameter identification framework, and required experimental head kinematics to be measured. The model-based interpretation of the experiments, and the ensuing clinical assessment of shaking injuries, led to the following conclusions:

1. The peak accelerations of the lamb’s heads during shaking were consistently between 200 m·s⁻² and 250 m·s⁻². The peak accelerations consistently occurred during impact of the head with the torso, or impact of the head with the shaker’s hands. The observation that head impact was responsible for the maximal accelerations is consistent with the suggestion that some form of impact is necessary to elicit the injuries of AHT (Duhaime et al., 1987; Prange et al., 2003).
2. The rigid-body computational modelling framework was able to reproduce the head kinematics of the lambs during manual shaking. The results support using a sagittal plane OpenSim model, with an exponential vertebral joint loading relationship, and Hunt-Crossley contact forces, to describe infant head kinematics in AHT.

3. The peak acceleration of each lamb’s head was most sensitive to the model parameters describing the contact stiffness, and the location of head impact. These sensitivities are consistent with the peak accelerations occurring during impact of the head with the torso. The model also demonstrated different sensitivities to model parameters that described the loading characteristics during flexion and extension. A higher sensitivity to the extension loading parameters was attributed to the specific anatomy of a lamb.

4. A clinical assessment of the lamb’s brains revealed no macroscopic signs of brain injury as a result of the single shaking episode protocol adopted in this study. There was no evidence to suggest that lambs exhibit the same brain injury mechanisms that have been proposed in human infants. The results, therefore, indicate that the lamb may not be an appropriate experimental analogue of AHT. Others have reported similar macroscopic results, although histological evidence of injury was observed (Finnie et al., 2012; Finnie et al., 2010). Histological analyses of animals used in this study are presently being performed (results are not described in this thesis) the results of which will be used to investigate the use of the lamb experimental model further.

7.2 Infant shaking simulations

A series of coupled rigid-body OpenSim computational models describing human infants were developed and were used to simulate an infant’s head kinematics during manual shaking. The infant models were coupled to an existing upper-body musculoskeletal model to prescribe realistic shaking boundary constraints, and a probabilistic analysis was performed to compare worst-case shaking scenarios to
published injury thresholds, and to previous experimental and computational results. The
human infant shaking simulations that were performed in this study led to the following
conclusions:

1. Impact of the infant’s head with the torso was observed to occur during worst-case
shaking scenarios. The results are consistent with the animal experimental
observations, and the suggestion that head-torso impact is a likely injury mechanism
in AHT (Cory & Jones, 2003).

2. Infant head kinematics were highly sensitive to the type of shaking motion that was
applied, but were relatively insensitive to the model parameters describing the infant’s
mechanical properties. The sensitivity results suggest that future research be directed
at investigating the shaking inputs rather than the material properties of the infant
when describing head kinematics during shaking. When shaking was applied along an
arc, the periodic transition between full flexion and full extension was promoted,
which resulted in larger peak kinematics than when compared to more linear shaking
motions.

3. The peak angular acceleration and peak angular velocity of an infant’s head was
found to depend most strongly on the frequency of shaking. The peak angular
acceleration was also observed to be strongly dependent on the contact stiffness
between the head and the torso, but the peak angular velocity was comparably
insensitive to the contact properties.

4. Shaking motions that were kinematically prescribed using the upper-body
musculoskeletal model were able to produce peak kinematics that exceeded published
injury thresholds. When the kinematics were subsequently constrained using the peak
isometric joint torque of a 50th percentile male, no simulated head kinematics were
observed to exceed scaled SDH thresholds. This was observed for the newborn, the
4.5 months, and the 12 months infant models. These results were similar when the
torque constraints were scaled to represent shakers of different strengths. The results
of this study provide no biomechanical evidence to demonstrate that shaking alone can cause the injuries observed in AHT. The implications are that either (i) additional factors, such as impact, are required, and/or (ii) that the current estimates of injury thresholds should be interpreted with caution.

5. The peak isometric joint torque constraint acted to limit the peak angular velocity that a shaking motion can produce. The peak angular velocity limit was observed to be inversely proportional to an infant’s age, and proportional to the strength of the shaker.

The computational modelling framework was used to simulate the statistical distributions of an infant’s peak head kinematics. These distributions were shown to encompass published experimental and computational results (Bondy et al., 2012; Cory & Jones, 2003; Duhaime et al., 1987; Wolfson et al., 2005). The framework was also able to be scaled to different infant ages and to shakers of different strengths. The framework is a versatile tool that can complement other modelling techniques in future AHT research.

7.3 Limitations and future work

The computational framework developed in this thesis was used to investigate an infant’s head kinematics during manual shaking in Chapter 6. When using this framework to address the clinical problem, there are a number of limitations that must be acknowledged. The model of the infant was restricted by the paucity of experimental data available and the mean parameter values are a representative example rather than being calculated from across a population. This is also true for the probability density functions used to describe the input parameters in the statistical analysis. The output CDF’s are limited by the quality of the input approximations, but as appropriate experimental data become available, the confidence in these probability distributions can be improved. The lack of data describing the material properties of an infant’s neck may, however, not be a significant limitation. The probabilistic results indicated that the material properties of the neck had little influence on the peak head kinematics, and therefore suggest that
increasing joint complexity may not substantially improve predictions of the overall head
ingematics. It may, however, be worth adding localised regions of complexity, possibly
using finite element modelling, to investigate specific biomechanical questions. These
could include investigating the potential for injury during flexion and extension of the
atlantoaxial-occipital joint complex, as has been proposed as a potential injury
mechanism in AHT (Finnie et al., 2010; Geddes, Hackshaw, et al., 2001; Geddes et al.,
2003; Geddes, Vowles, et al., 2001; Shannon et al., 1998).

The model consisted of joints that were restricted to a single rotational DOF in the
sagittal plane. This simplification limits the biofidelity of the model, but in vivo shaking
kinematics have been reproduced well using a similar computational model during
sagittal plane shaking (Chapter 5). Although the model provides no information about any
out-of-plane motion, the shaking inputs simulated in this study are consistent with those
thought to occur in AHT (Cory & Jones, 2003; Duhaime et al., 1987; Prange et al., 2003).
Other loading mechanisms, such as coronal head rotations (Gennarelli et al., 1982), have
been thought to be significant but are outside the scope of this study. The joints in the
model were also described using a representative geometrical description of the infant’s
neck that was not scaled 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.

There were also a number of limitations in the way that the shaking motion was
applied in the model. 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 that, at
times, the joints were forced through motions that the muscle forces could not support.
An improvement upon this would be to prescribe the muscle activations in a full
musculoskeletal model, such as described by Holzbaur et al., (2005), which would result
in a torque boundary condition acting upon the joints. Although possible, there would be
significant difficulties in estimating appropriate activation patterns for the complex array of shaking motions that could potentially occur. Alternatively, an extensive population based experimental study could be performed where the shaking patterns of individuals could be investigated. In addition to providing more realistic shaking parameters, the results would improve the accuracy of the PDFs of the shaking parameters and would provide more confidence in the statistics of the output kinematics. Without these, the joint torques were estimated by excluding the additional inertial and contact forces associated with the head motion and head impact. This will likely underestimate the joint torques necessary to produce the shaking. How the shaking motions are applied, and how the kinetic constraint are imposed, is an area that may be improved in subsequent research. Improvements may include adding shaking motions that are not restricted to the sagittal plane and/or introducing degrees of freedom to the model that allow the neck to stretch axially (see section 6.1.2.6 for further discussion). Other improvements include the application of a joint torque boundary condition, the contribution of the torso motion on the shaking forces, and an investigation into the positioning of the infant in the hands. Although further investigation into an infant’s material properties would improve the biofidelity of the model, it is likely that there would be a greater benefit in analysing the input biomechanics when investigating an infant’s head kinematics during AHT.

Another major limitation is with the sensitive nature of the subject matter. Head kinematics during violent shaking are highly unlikely to be obtained in vivo and it is unlikely that cadaveric experiments will be performed due to the paucity of infant specimens. As a result, there is limited validation that can be performed. Computational results can be compared to animal and anthropometric test dummy experimental results, and have shown good agreement, but these comparisons are made across tests with different properties and under different shaking conditions. This study is not limited to single shaking events but allows a range of head kinematics to be produced, which allows worst case scenarios to be simulated. These results are interpreted with respect to published concussion and SDH injury thresholds, which are also limited by the lack of
experimental data. As detailed experimental investigations of brain injury thresholds in human infants are unlikely, the field must turn to computational modelling studies to identify more appropriate biomechanical injury metrics. The rigid-body shaking simulation results will complement this, providing realistic worst-case boundary conditions for subsequent analysis. Computational studies such as these may produce results with forensic applications. The ability to estimate brain loading during shaking and to compare these kinematic data to accurate injury thresholds would allow the injuries of AHT to be investigated with confidence, which may help address the legal contention that exists.
Appendix A

In vivo kinematic results

During the in vivo shaking experiments, the head and torso kinematics were measured with IMUs and motion capture, respectively. The raw inertial data were used to quantify the head kinematics for each lamb, and motion capture was used to measure the positions of the lambs’ torsos, and these were prescribed as kinematic (displacement) boundary constraints in the simulations. The sagittal plane inertial measurements, the boundary constraints, and the optimal sagittal plane kinematic components reproduced by the model that were not described in sections 5.2, 5.3.3 and 5.4.2, respectively, are illustrated below.

A.1 Sagittal plane inertial measurements

Figure A.1 Acceleration ($a$) and angular velocity ($\omega$) measured for Lamb1 with maxima and minima on the sagittal plane axes marked with circles. Minima for $a_x$ and $a_z$ correspond to contact during full flexion and maxima for $a_x$ correspond to contact during full extension.
Figure A.2 Acceleration ($a$) and angular velocity ($\omega$) measured for Lamb2 with maxima and minima on the sagittal plane axes marked with circles. Minima for $a_x$ and $a_z$ correspond to contact during full flexion and maxima for $a_x$ correspond to contact during full extension.

Figure A.3 Acceleration ($a$) and angular velocity ($\omega$) measured for Lamb6 with maxima and minima on the sagittal plane axes marked with circles. Minima for $a_x$ and $a_z$ correspond to contact during full flexion and maxima for $a_x$ correspond to contact during full extension.
A.2 Kinematic boundary constraints

Figure A.4 The Euler angles ($\theta_i$) and absolute positional coordinated of the T2 vertebra that were prescribed as kinematic (displacement) boundary constraints for Lamb1.

Figure A.5 The Euler angles ($\theta_i$) and absolute positional coordinated of the T2 vertebra that were prescribed as kinematic (displacement) boundary constraints for Lamb2.
Appendix A

A.3 Predicted sagittal plane motion following parameter estimation

Figure A.7 Optimal sagittal plane kinematic components (blue) that are compared to the experimental measurements for Lamb2.
Figure A.8 Optimal sagittal plane kinematic components (blue) that are compared to the experimental measurements for Lamb5.

Figure A.9 Optimal sagittal plane kinematic components (blue) that are compared to the experimental measurements for Lamb6.
Appendix B

Sensitivity of lamb model

The sensitivity of each lamb’s head accelerations to perturbations of different model parameters was investigated. The following figures (Figure B.1-B.21) illustrate the sensitivity of the head accelerations for each lamb, and they complement the results described in section 5.5. The sensitivities were analysed for a shaking cycle corresponding to the fitted region in the optimisations, and for an additional shaking cycle that was not included in the optimisation. For each lamb model, the mass, damping, and joint loading parameters, were varied individually by ±20% of each parameter’s respective optimal value (Table 5.4) and the changes to the head acceleration were observed. In each graph, the optimal simulation result is depicted by the black line and the simulation results using higher parameter values are in red, where parameter values that are 10% and 20% above the optimum are depicted by light and dark red, respectively. Similarly, the simulation results using lower parameter values are in blue, where parameter values that are 10% and 20% below the optimum are depicted by light and dark blue, respectively. The parameters describing the positions of the contact regions and the contact stiffness were varied throughout a range of realistic values and are listed on each figure. For these parameters, the optimum simulation is in black, increasing parameter values in light and dark red, and decreasing parameter values in light and dark blue. The results demonstrated consistent parameter sensitivities for Lamb2, Lamb5, and Lamb6, which are described in section 5.5. Lamb1, however, demonstrated parameter sensitivities that were inconsistent with the other models. These results were attributed to the optimal model-predicted simulations not reproducing the experimental accelerations (Figure 5.8). In not reproducing the accelerations, it is possible that the physics of the shaking dynamics predicted by the Lamb1 model were different to the physical case, resulting in the different model behaviour.
B.1 Lamb1 model sensitivity

B.1.1 Damping and mass sensitivity

Figure B.1 Acceleration magnitude simulated during perturbation analysis while varying damping (a) and mass (b) using the model of Lamb1 in a shaking cycle within the fitted region. Results demonstrate large changes in acceleration when the damping is decreased, which is inconsistent with the observations using the other lamb models.

Figure B.2 Acceleration magnitude simulated during perturbation analysis while varying damping (a) and mass (b) using the model of Lamb1 in a shaking cycle outside of the fitted region. Similar behaviour to the shaking cycle within the fitted region was observed.
B.1.2 Joint loading sensitivity

Figure B.3 Acceleration magnitude simulated during perturbation analysis while varying $k_f$ (a) and $k_e$ (b) using the model of Lamb1 in a shaking cycle within the fitted region. Large changes in the acceleration were observed when the joint loading parameters were varied. This behaviour is inconsistent with that observed for the other lamb models.

Figure B.4 Acceleration magnitude simulated during perturbation analysis while varying $k_f$ (a) and $k_e$ (b) using the model of Lamb1 in a shaking cycle outside of the fitted region. Similar behaviour to the shaking cycle within the fitted region was observed.
B.1.3 Contact parameters sensitivity

Figure B.5 Acceleration magnitude simulated while varying $\text{Contact}_{\text{flex}}$ (a), $\text{Contact}_{\text{ext}}$ (b), and $K_{\text{cont}}$ (c) using the model of Lamb1 in a shaking cycle within the fitted region. The acceleration was observed to increase during decreases in all contact parameters. This behaviour is inconsistent with that observed for the other lamb models.
Figure B.6 Acceleration magnitude simulated while varying Contact\textsubscript{flex} (a), Contact\textsubscript{ext} (b), and $K_{\text{cont}}$ (c) using the model of Lamb1 in a shaking cycle outside of the fitted region. Similar behaviour to the shaking cycle within the fitted region was observed.
B.2 Lamb2 model sensitivity

B.2.1 Damping and mass sensitivity

Figure B.7 Acceleration magnitude simulated during perturbation analysis while varying damping (a) and mass (b) using the model of Lamb2 in a shaking cycle within the fitted region. Results illustrate that the damping and mass parameter perturbations resulted in small changes to the magnitude and timing of the acceleration peaks. These results are consistent with those described in Chapter 5.

Figure B.8 Acceleration magnitude simulated during perturbation analysis while varying damping (a) and mass (b) using the model of Lamb2 in a shaking cycle outside of the fitted region. Similar behaviour to the shaking cycle within the fitted region was observed.
B.2.2 Joint loading sensitivity

Figure B.9 Acceleration magnitude simulated during perturbation analysis while varying $k_f$ (a) and $k_e$ (b) using the model of Lamb2 in a shaking cycle within the fitted region. Results indicate that the acceleration increases when the joint loading parameters are decreased. These results are consistent with those described in Chapter 5, although the dependence on $k_f$ and $k_e$ appears to more similar than that observed for Lamb6 (Figure 5.16).

Figure B.10 Acceleration magnitude simulated during perturbation analysis while varying $k_f$ (a) and $k_e$ (b) using the model of Lamb2 in a shaking cycle outside of the fitted region. Similar behaviour to the shaking cycle within the fitted region was observed.
B.2.3 Contact parameters sensitivity

Figure B.11 Acceleration magnitude simulated while varying Contactflex (a), Contactext (b), and \( K_{\text{cont}} \) (c) using the model of Lamb2 in a shaking cycle within the fitted region. All parameter values are described in the accompanying legend in the lower right. Results indicate that the acceleration can be modulated by varying both the contact position and \( K_{\text{cont}} \). These results are consistent with those described in Chapter 5.
Figure B.12 Acceleration magnitude simulated while varying Contact_flex (a), Contact_ext (b), and $K_{cont}$ (c) using the model of Lamb2 in a shaking cycle outside of the fitted region. Similar behaviour to the shaking cycle within the fitted region was observed.
B.3 Lamb5 model sensitivity

B.3.1 Damping and mass sensitivity

Figure B.13 Acceleration magnitude simulated during perturbation analysis while varying damping (a) and mass (b) using the model of Lamb5 in a shaking cycle within the fitted region. Results illustrate that the damping and mass parameter perturbations resulted in small changes to the magnitude and timing of the acceleration peaks. These results are consistent with those described in Chapter 5, although peak kinematics occur during extension rather than flexion.

Figure B.14 Acceleration magnitude simulated during perturbation analysis while varying damping (a) and mass (b) using the model of Lamb5 in a shaking cycle outside of the fitted region. Similar behaviour to the shaking cycle within the fitted region was observed.
B.3.2 Joint loading sensitivity

Figure B.15 Acceleration magnitude simulated during perturbation analysis while varying $k_f$ (a) and $k_e$ (b) using the model of Lamb5 in a shaking cycle within the fitted region. Results indicate that $k_e$ has a greater influence than $k_f$ on the kinematics and that the peak accelerations increase as the joint loading parameters are decreased. These results are consistent with those described in Chapter 5.

Figure B.16 Acceleration magnitude simulated during perturbation analysis while varying $k_f$ (a) and $k_e$ (b) using the model of Lamb5 in a shaking cycle outside of the fitted region. Results demonstrate that decreasing joint loading parameters can result in impact between the head and torso being initiated. These results are slightly different to those in Figure B.15, but were attributed to the shaking cycle not producing impact during full flexion. Unlike in Figure B.15, decreasing $k_e$ was insufficient to initiate impact during flexion.
B.3.3 Contact parameters sensitivity

Figure B.17 Acceleration magnitude simulated while varying $\text{Contact}_{\text{flex}}$ (a), $\text{Contact}_{\text{ext}}$ (b), and $K_{\text{cont}}$ (c) using the model of Lamb5 in a shaking cycle within the fitted region. All parameter values are described in the accompanying legend in the lower right. Results indicate that the acceleration can be modulated by varying both the contact position and $K_{\text{cont}}$. These results are consistent with those described in Chapter 5, although the peak accelerations were observed to occur during full extension and not flexion.
Figure B.18 Acceleration magnitude simulated while varying $\text{Contact}_\text{flex}$ (a), $\text{Contact}_\text{ext}$ (b), and $K_{\text{cont}}$ (c) using the model of Lamb5 in a shaking cycle outside of the fitted region. Similar behaviour to the shaking cycle within the fitted region was observed.
B.4 Lamb6 model sensitivity

B.4.1 Damping and mass sensitivity

Figure B.19 Acceleration magnitude simulated during perturbation analysis while varying damping (a) and mass (b) using the model of Lamb6 in a shaking cycle outside of the fitted region. Results illustrate that the damping parameter influences the magnitude and timing of acceleration peaks and the mass influences only the magnitude. Similar behaviour to the shaking cycle within the fitted region was observed (Figure 5.15).

B.4.2 Joint loading sensitivity

Figure B.20 Acceleration magnitude simulated during perturbation analysis while varying $k_f$ (a) and $k_s$ (b) using the model of Lamb6 in a shaking cycle outside of the fitted region. Results indicate that $k_s$ has a greater influence than $k_f$ on the kinematics. Similar behaviour to the shaking cycle within the fitted region was observed (Figure 5.16).
B.4.3 Contact parameters sensitivity

Figure B.21 Acceleration magnitude simulated while varying $\text{Contact}_{\text{flex}}$ (a), $\text{Contact}_{\text{ext}}$ (b), and $K_{\text{cont}}$ (c) using the model of Lamb6 in a shaking cycle outside of the fitted region. All parameter values are described in the accompanying legend in the lower right. Results indicate that the acceleration can be modulated by varying both the contact position and $K_{\text{cont}}$. Similar behaviour to the shaking cycle within the fitted region was observed (Figure 5.17).
Scaling of published injury criteria

As described in section 6.2, the concussion and subdural haemorrhage (SDH) injury thresholds used in this study were scaled from those described by Duhaime et al. (1987). The thresholds were obtained by scaling according to brain mass (Margulies et al., 1990) and the material properties of an infant’s brain (Thibault & Margulies, 1998). The scaling relationships for the peak angular acceleration ($\ddot{\theta}$) and the peak angular velocity ($\dot{\theta}$) injury thresholds are described in Equations C.1-C.2. The subscript “Duhaime” and “Infant” correspond to measurements used by Duhaime et al. and those used to describe the infants in this study, respectively. $M$ denotes the brain mass of each infant and $G'_{infant}$ and $G'_{adult}$ were defined as the elastic portion of the complex shear modulus for infant and adult brains, respectively. The ratio $G'_{infant}/G'_{adult}$ was found to be 0.667 in porcine brains (Thibault & Margulies, 1998) and was used in this study.

$$\ddot{\theta}_{Infant} = \ddot{\theta}_{Duhaime} \left( \frac{M_{Duhaime}}{M_{Infant}} \right)^{2/3} \left( \frac{G'_{Infant}}{G'_{Adult}} \right)$$  \hspace{1cm} \text{C.1}$$

$$\dot{\theta}_{Infant} = \dot{\theta}_{Duhaime} \left( \frac{M_{Duhaime}}{M_{Infant}} \right)^{1/3} \left( \frac{G'_{Infant}}{G'_{Adult}} \right)$$  \hspace{1cm} \text{C.2}$$

The results of the scaling are described in Table C.1 and use estimates of brain mass for each of the three infant ages. The brain mass was estimated as a function of body weight (obtained from average growth charts (World Health Organization, 2006)) using results described by Dobbing & Sands (1973). It should be noted that these injury thresholds are scaled from primate experiments investigating whiplash during car accidents and are not considered to be definitive injury metrics.
Table C.1 Injury thresholds that were scaled from Duhaime et al. (1987) to the age ranges used in this study.

<table>
<thead>
<tr>
<th></th>
<th>Duhaime et al.</th>
<th>Newborn</th>
<th>4.5 months</th>
<th>12 months</th>
</tr>
</thead>
<tbody>
<tr>
<td>Body weight (kg)</td>
<td>-</td>
<td>3.4</td>
<td>7.5</td>
<td>9.6</td>
</tr>
<tr>
<td>Brain mass (kg)</td>
<td>0.5</td>
<td>0.4</td>
<td>0.9</td>
<td>1.2</td>
</tr>
<tr>
<td>Concussion $\dot{\theta}$ (rad$\cdot$s$^{-2}$)</td>
<td>10000</td>
<td>7740</td>
<td>4508</td>
<td>3721</td>
</tr>
<tr>
<td>SDH $\dot{\theta}$ (rad$\cdot$s$^{-2}$)</td>
<td>35000</td>
<td>27089</td>
<td>15777</td>
<td>13023</td>
</tr>
<tr>
<td>Concussion $\dot{\theta}$ (rad$\cdot$s$^{-1}$)</td>
<td>86</td>
<td>61.8</td>
<td>47.2</td>
<td>42.8</td>
</tr>
<tr>
<td>SDH $\dot{\theta}$ (rad$\cdot$s$^{-1}$)</td>
<td>105</td>
<td>75.4</td>
<td>57.6</td>
<td>52.3</td>
</tr>
</tbody>
</table>
Monte Carlo convergence

The Monte Carlo method is often referred to as the ‘gold standard’ of probabilistic methods as convergence is ensured when the number of simulation trials tends towards infinity (Haldar & Mahadevan, 2000). The applicability of the method is only limited by the tractability of the problem. A convergence analysis was performed to estimate the number of simulation trials necessary for the analysis in Chapter 6. The performance metrics used in the probabilistic analysis were the peak angular acceleration and the peak angular velocity of an infant’s head during shaking.

Convergence was assessed by discretising the CDF for each performance metric into twenty even probability levels, and measuring the probability for the third, tenth, and eighteenth probability levels. For 3000 simulation trials, the probability estimated at each level had converged to within 0.002 of the probability that was measured for 2500 simulation trials, indicating that sufficient convergence had been attained (Figure D.1).

Figure D.1 Monte Carlo analyses that were performed to estimate the CDF for the peak angular acceleration (a) and the peak angular velocity (b) that occurred during shaking of a 4.5 months infant. The red, blue and green traces show the third, tenth, and eighteenth probability levels, respectively.
Appendix E

Scaling of joint torque constraints

The simulated head kinematics of a human infant during shaking were kinetically constrained by restricting the prescribed shaking motions that were applied to the infant models. The peak isometric joint torques of the shoulder, elbow, and wrist joints were used to identify what shaking motions were realistic. In Chapter 6 (section 6.3.3), the simulation results for a newborn, 4.5 months, and a 12 months infant were constrained using the peak isometric joint torques that were predicted by a 50\textsuperscript{th} percentile male upper body model (Holzbaur et al., 2005). In this appendix, the joint torque constraints were halved and doubled to investigate the effect of varying the strength of the shaker upon the simulated head kinematics. These results are illustrated in Figure E.2 and Figure E.3, respectively, and for comparison purposes the results for the 50\textsuperscript{th} percentile male have been reproduced in Figure E.1. The peak angular velocities ($\omega_{\text{peak}}$) were observed to scale with the joint torque constraints for all infant ages. The peak angular acceleration ($\alpha_{\text{peak}}$) also scaled with the joint torque constraints, but the effect was less pronounced. It is unlikely that the achievable isometric joint torques can double between shakers of different strengths, so these results indicate that even for a shaker that is much stronger than average, producing head kinematics that exceed the SDH injury thresholds is unlikely, because even when using these unrealistic joint torque constraints, very few simulations exceeded the SDH injury thresholds.
Figure E.1 Peak angular accelerations ($\alpha_{\text{peak}}$) and angular velocities ($\omega_{\text{peak}}$) for the kinetically constrained shaking simulations for a newborn infant (a), a 4.5 months infant (b) and a 12 months infant (c). Significantly higher peak angular accelerations were observed for the newborn infant but little difference was observed between the 4.5 months and 12 months cases. Only small changes in the peak angular velocity were observed and no results exceeded scaled SDH injury thresholds.

Figure E.2 Peak angular accelerations ($\alpha_{\text{peak}}$) and angular velocities ($\omega_{\text{peak}}$) for the kinetically constrained shaking simulations for a newborn infant (a), a 4.5 months infant (b) and a 12 months infant (c). Each peak isometric joint torque constraint was half of that described in section 6.3.3. Results demonstrate that the peak $\omega_{\text{peak}}$ is less than that predicted using a 50th percentile male upper body model.

Figure E.3 Peak angular accelerations ($\alpha_{\text{peak}}$) and angular velocities ($\omega_{\text{peak}}$) for the kinetically constrained shaking simulations for a newborn infant (a), a 4.5 months infant (b) and a 12 months infant (c). Each peak isometric joint torque constraint was double that described in section 6.3.3. Results demonstrate that the peak $\omega_{\text{peak}}$ is greater than that predicted using a 50th percentile male upper body model and a small proportion of simulations were observed to exceed the SDH injury thresholds.
List of References


Holbourn, A. H. S. (1956). Private communication to Dr. Sabina Strich.


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