Infant restraint systems in air transport: 
Injury risk and prevention

A thesis submitted in fulfilment of the requirements for the 
degree of Doctor of Philosophy

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July 2014
Declaration

I certify that except where due acknowledgement has been made, the work is that of the author alone; the work has not been submitted previously, in whole or in part, to qualify for any other academic award; the content of the thesis is the result of work which has been carried out since the official commencement date of the approved research program; any editorial work, paid or unpaid, carried out by a third party is acknowledged; and ethics procedures and guidelines have been followed.

Adam J. Shrimpton

22 July 2014
Acknowledgements

I am grateful to my supervisors Professor Graham Clark and Associate Professor Cees Bil for their guidance and patience over the course of my candidature. I would also like to thank the Civil Aviation Safety Authority and the Australian government for funding this research project.

This work was made possible by prior research in the fields of aviation safety and biomechanics; in particular, I would like to acknowledge Mr Mark Bathie of CASA. I am indebted to Mr Rick DeWeese and Mr David Moorcroft of the FAA Civil Aeromedical Institute and Mr Ryan Adams of ADVEA Engineering for their generous technical guidance and encouragement.

This effort would have been impossible for me if not for the love, support, patience and encouragement of my family and friends. In particular I would like to thank my wife Diana, my parents Barry and Elizabeth, my sisters Rachael and Nicole and my dear friends Megan, Michael, Jason, Daniel and the Ippel family.

Adam J. Shrimpton

22 July 2014
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## Nomenclature

- $b(t)$: Benchmark signal for input to Sprague and Geers error metric
- $c(t)$: Candidate signal for input to Sprague and Geers error metric
- $C$: Sprague and Geers comprehensive error factor
- $g$: Acceleration due to gravity
- $HIC$: Head injury criterion
- $HIC_{15}$: Head injury criterion, 15 ms maximum time window
- $HIC_{36}$: Head injury criterion, 36 ms maximum time window
- $M_{neg}$: Magnitude error factor, signal negative peak
- $M_{pos}$: Magnitude error factor, signal positive peak
- $M_{SG}$: Sprague and Geers magnitude error factor
- $N_{i}$: Neck injury criterion
- $N_{i}^*$: Neck injury criterion calculated using non-standard critical values
- $P$: Sprague and Geers phase error factor
- $W_{neg}$: Weighting of signal negative peak
- $W_{pos}$: Weighting of signal positive peak
Summary

Aviation safety regulations in Australia and overseas permit the carriage of an infant passenger on the lap of an adult; however, this practice is known to be unsafe in an emergency landing scenario. While automotive child restraint systems (CRS) have been found to provide adequate protection to infant passengers, safety issues associated with their use have been identified. Principally, they may interfere with the dynamic behaviour of the aircraft seat, increasing the risk of injury to an adult seated directly behind. In particular, the behaviour of CRS installed using the novel LATCH and ISOFIX methods has previously been poorly understood.

This study compared and evaluated the safety of seating arrangements involving infant air transport passengers under emergency landing conditions. Two arrangements were considered: the infant was either held on the lap of an adult or seated in a forward-facing CRS. A partially validated numerical model was used to study seat behaviour and measure injury potential. Results were analysed to compare the safety of different configurations, to evaluate safety in terms of an occupant protection standard and to predict the occurrence and severity of a particular injury using existing biomechanical relationships.

One of two practices applies to lap-held infants depending on the jurisdiction a flight is operating under: the infant is either unrestrained or restrained by a supplementary loop belt. This study presents the first comparison of the safety of these restraint conditions and an evaluation of the risk of abdominal injury associated with the supplementary loop belt. Analysis demonstrated that, regardless of restraint condition, the lap-held arrangement does not represent a means of protecting an infant from injury in an emergency landing scenario. During the impact sequence an unrestrained lap-held infant is likely to be projected through the aircraft cabin, while a lap-held infant restrained by a supplementary loop belt is at risk of serious abdominal injuries.

Though the adoption of an emergency brace position by an adult carrying a lap-held infant was found to be beneficial, infant injury potential remained high. Of the three brace positions studied, a modified position proposed by this study was the most effective in reducing head injury potential for the infant.

Three CRS installation methods - the aircraft seat lap belt, ISOFIX and LATCH - were analysed in terms of CRS performance and behaviour. Each installation method was found to
adequately protect the infant CRS occupant. The ISOFIX method resulted in a small increase in head injury potential for an adult seated directly aft, while in the lap belt and LATCH cases a significant increase was observed. The absence of a CRS occupant was found to slightly mitigate head injury to the aft-seated adult in the ISOFIX case but resulted in a further significant increase in the lap belt and LATCH cases. The presence of a CRS in the seat directly forward of an adult passenger was found to mitigate the risk of neck injury in the adult regardless of whether the CRS is occupied.

Some airlines specifically prohibit the use of Australian-certified CRS, presumably due to the requirement that these CRS be installed using a ‘top tether’ strap. This study found that the use of a top tether does not present any benefit in terms of reduced CRS motion or decreased injury potential for the CRS occupant. The prevention of aircraft seat-back rotation did not improve the efficacy of the top tether; in fact, this served to promote infant head and neck injury.

This study makes the following four recommendations:

1. The practice of holding an infant passenger on the lap should be prohibited. Until this is the case, advice on emergency brace positions involving lap-held infants should be revised to include a lateral offset between the head of the infant and that of the adult.
2. The use of automotive child restraints to restrain infant passengers should be mandated.
3. Any requirement for the use of a top tether in aircraft CRS installations should be removed.
4. Child restraint systems should be installed in outboard seats directly forward of a bulkhead, unoccupied space or another CRS.
1 Introduction

1.1 Background and significance

In 1962 Dr Stanley R. Mohler, the director of what is now the US Federal Aviation Administration Civil Aeromedical Institute, wrote:

“A particular need exists for proper infant protective equipment (in many cases at present the mother must hold her infant during the take-off and landing of an airliner)...”[1]

Fifty years later it remains common practice for an infant to be held on the lap of an adult during air travel despite several studies in the last two decades finding that this is unsafe in an emergency landing[2-5]. Automotive child restraint system (CRS) are known to improve the safety of infant aircraft passengers[2, 3, 6-8], though their use in aircraft is complicated by issues such as geometrical incompatibility with the aircraft seat, poorly controlled CRS motion under emergency landing conditions and, in Australia, the mandatory use of a ‘top tether’ strap[3-6, 9, 10]. Preliminary evaluations[2, 10] of novel CRS attachment methods such as ISOFIX and LATCH suggest they overcome the issue of poorly controlled CRS motion but may restrict the energy-absorbing function of the aircraft seat back and promote injury to a passenger seated directly aft.

From a regulatory viewpoint, the issue as it stands today presents several problems:

• Aviation safety regulations[11, 12] permitting the carriage of infants on the lap are incongruous with advisory material[8] and research[4] published by the same regulatory bodies declaring this practice to be unsafe.

• At the international level, aviation safety regulators disagree on whether a lap-held infant is best protected when restrained by a supplementary belt or by only the arms of the adult passenger[11, 13].

• Emergency brace positions are proven to provide significant protection to adults during emergency landings[14]. None have been evaluated for lap-held infants.

• The use of CRS in aircraft is widely permitted by law; however, in Australia, individual airlines may prohibit their use.

A possible reason for this lack of coherence is that studies on the topic to date have been disparate in their method and scope, producing little or no data describing injury risk for the
This situation is in turn a result of the experimental method employed by those studies[2-5], i.e. physical deceleration sled tests. This type of experiment is costly and does not permit the measurement of the impact loads on the infant; these loads must be fully understood before an accurate prediction of injury potential for the infant can be made.

This study applies a numerical modelling approach to the problem of how to most safely carry infants aboard transport aircraft. In so doing it forms the first detailed evaluation and comparison of injury potential across a range of configurations involving infants in air transport. This information is used to arrive at a series of recommendations aimed at improving the safety of infant airline passengers.

1.2 Research questions

The objectives of this study were to evaluate the injury potential for a lap-held infant and to determine the safety risks associated with the use of automotive child restraint systems, particularly those installed by the novel LATCH and ISOFIX methods. A review of global aviation safety regulations, air accident reports and past studies related to infant safety in air transport gave rise to the following research questions:

1. What are the safety risks for the lap-held infant?
   i. How is the risk of injury for the lap-held infant affected by the use or non-use of a supplementary loop belt?
      This study presents the first comparison of injury potential for the lap-held infant with and without the use of a supplementary loop belt.

   ii. What are the characteristics of the forces acting on the head of the lap-held infant?
      This study is the first to predict the forces acting on the infant’s head during impact. This information will be useful in the future as infant head injury models are refined, enabling a detailed analysis of infant head injury potential.

   iii. Are the recommended emergency brace positions involving lap-held infants effective in preventing injury to the infant?
      This study is the first to evaluate the efficacy of emergency brace positions involving lap-held infants. A comparison of the performance of two brace
positions recommended by safety regulators is presented and a modified position which results in improved safety outcomes is proposed.

2. What are the safety risks associated with the use of automotive child restraints in air transport, particularly those installed by novel methods?
   
i. How do novel CRS installation methods affect the dynamic performance of the CRS and what is the corresponding effect on injury potential for the infant?
   
   Novel CRS installation methods have been identified as a possible solution to poorly controlled CRS motion in aircraft. This study is the first to undertake a detailed analysis of the dynamic performance of CRS installed using novel methods with a focus on occupant injury as well as CRS kinematics.

   ii. How do novel CRS installation methods alter the behaviour of the aircraft seat and what is the corresponding effect on the safety of an adult seated directly aft?
   
   This study is the first to investigate in detail the potential negative effects on the safety of an adult seated directly aft of a CRS installed by novel methods. Through a parametric analysis taking into account seat pitch, CRS installation method and CRS occupant size the mechanisms of these effects are identified and discussed.

   iii. Are current practices of top tether use effective in controlling CRS motion and what is the corresponding effect on infant injury potential?
   
   The regulatory requirement in Australia for the use of a top tether in CRS installation may discourage airlines from allowing the use of CRS. This study is the first to investigate the efficacy of the top tether in a typical airline installation including a situation where the aircraft seat has been modified as proposed by a previous study.

1.3 Scope

This study considered the safety of seating arrangements involving infants under emergency landing conditions only; other loading conditions such as turbulence were not considered.
1.4 Method overview

Several series of experiments, detailed in chapters six to nine, were designed and carried out with the aim of answering the research questions posed in section 1.2 above. To facilitate these experiments a numerical model of a typical air transport seating arrangement was developed for use as an experimental platform. This method was chosen over further physical experiments for its infinite adaptability, low cost per experiment and its ability to measure important parameters such as contact force[15]. The model development process is detailed in chapter three.

To enable validation of the model its design replicated that of physical experiments carried out by the Australian Civil Aviation Safety Authority (CASA)[2]. In the validation process, eight of the experiments carried out by the CASA study were replicated using the numerical model and the results were compared with high frame-rate video footage and sensor signal data obtained from CASA. The details of model verification, validation and implementation are presented in chapter five.

This study was developed and conducted within the bounds of several limitations which are detailed in section 10.3.
2 Infant air transport passengers: a review

2.1 Introduction

This chapter presents a review of Australian and international transport safety regulations, air transport accident reports and past studies relevant to the safety of infant air transport passengers under emergency landing conditions.

2.2 Regulatory and operational aspects

2.2.1 Australian aviation regulations

Australian aviation regulations [11] require that all passengers, regardless of age, be restrained by a harness or seat belt during:

- takeoff,
- landing,
- instrument approach,
- flight less than 1000 feet above terrain, and
- turbulence.

During these times, all passengers must occupy a seat. An exception is made for infants, defined as passengers aged less than three years, who may alternatively be:

- held on the lap of an adult passenger,
- carried in a suitably-restrained bassinet, or
- seated in an approved and suitably-restrained ‘infant seat’.

This provision does not except any passenger from the requirement that all passengers be restrained. CASA advisory material[11] states that a supplementary loop belt (see section 2.3.1) is the only known device that provides acceptable restraint for an infant held on the lap of an adult. Currently, no bassinet devices are approved for use as a restraint in Australian transport category aircraft.

CASA advisory material[11] also states that during a severe but survivable crash a lap-held infant restrained by a supplementary loop belt is not provided a level of safety equivalent to a separately-seated adult.
2.2.2 International aviation regulations

Aviation regulations in Europe and the United Kingdom allow infants up to the age of 24 months to be held on the lap of an adult passenger; in this situation the use of a supplementary loop belt is mandatory. Infants may also be seated in an approved CRS.

Aviation regulations in the United States and Canada allow infants up to the age of 24 months to be held on the lap of an adult passenger. In these jurisdictions the use of a supplementary loop belt is prohibited and the infant must be restrained only by the arms of the adult passenger. Infants may also be seated in an approved CRS.

Aviation regulations in the United States are unique in their requirement that airlines must allow the use of aviation-approved CRS when a ticket has been purchased for the seat in which the CRS will be installed[13]. Such a requirement does not exist in Australia or other countries, leading to a situation where some airlines prohibit the use of CRS.

2.2.3 Australian operational environment

In Australia individual airlines set their own policies regarding the carriage of infants within the scope of CASA regulations. Of the four major domestic carriers, each allows infants up to the age of 24 months to be restrained on the lap of an adult[16-19]. Two carriers allow the use of CRS meeting Australian and overseas design standards[16, 17]. One carrier specifically prohibits the use of CRS meeting Australian design standards but allows those from overseas[19] while the fourth prohibits the use of CRS entirely[18]. Of the two domestic carriers that allow the use of Australian CRS, one carrier’s policy is that rear-facing restraints will only be approved for use if a member of the infant’s travelling party occupies the seat forward of the CRS due to potential difficulties in reclining that seat[16].

2.3 Safety of the lap-held infant

Lap-held infants account for approximately 1% of air transport passengers in the US[20]. Worldwide, the regulations of individual jurisdictions mandate one of two opposing restraint conditions for lap-held infants:

1. The infant must be restrained by a supplementary loop belt, as is the case in jurisdictions including Australia, the United Kingdom and Europe.
2. The infant is unrestrained, as is the case in jurisdictions including the United States and Canada. The use of a supplementary loop belt is prohibited.
2.3.1 The supplementary loop belt

The supplementary loop belt is essentially a lap belt that is fastened around the infant and connected to the adult’s belt by a webbing loop. It consists of a length of 50 mm wide polyester or nylon webbing of the type used in the aircraft seat lap belt. At each end are mating halves of a typical aircraft seat belt buckle; stitched at the centre is a small connecting loop of webbing which the adult’s lap belt is passed through.

![Image of the supplementary loop belt](image)

Figure 2-1 The supplementary loop belt[7].

Restraints suitable for infants should spread restraint forces over a large portion of the body [21]; however, the supplementary loop belt applies restraint forces over only a small portion of the body and in an area with little underlying structure. In adults, a lap belt is positioned over the bony structure of the iliac crest in order to most effectively transmit the arresting force into the body. In infants, however, the iliac crest is not fully developed and this results in the supplementary loop belt being positioned entirely over the abdomen[7].

Several studies[2-5, 7] have considered the safety of lap-held infants. However, each study considered only one of two possible restraint conditions for the infant: the infant was either unrestrained or restrained by a supplementary loop belt. No previous study has compared the effect of supplementary loop belt use or non-use on the loads acting on the lap-held infant.
Lap-held infants are vulnerable to impact with the forward seat and the adult passenger on whose lap they are seated regardless of restraint condition[2-5, 7]. The two primary safety-related differences between the two restraint conditions are that an unrestrained infant is at risk of being thrown about the aircraft cabin if the adult is unable to restrain them[5, 20, 22, 23] while an infant restrained by a supplementary loop belt is at risk of the so-called ‘seat belt syndrome’[28] (see section 3.4.4).

### 2.3.2 Injury potential

Several studies have made qualitative assessments of the safety of an infant carried on the lap of an adult aircraft passenger during rapid forward deceleration[2-5, 7]. In each of these studies, aircraft seats were mounted on a deceleration sled and occupied by anthropomorphic test devices (ATDs, or ‘crash test dummies’). Differences in ATD type, applied acceleration pulse and aircraft seat type and configuration preclude a direct comparison of results; however, the conclusion of each study was similar: the lap-held infant is at risk of injury under emergency landing conditions. No study has thus far attempted to quantify infant injury potential.

- A 1990 study[5] by Hardy assessed the case of an unrestrained lap-held infant, finding that this method “is likely to promote fatalities and injuries to these children during impact situations.”
- A 1994 report[4] by Gowdy and DeWeese at the US Federal Aviation Administration (FAA) Civil Aeromedical Institute assessed the case of a lap-held infant ATD restrained by a supplementary loop belt. The authors reported that dynamic testing confirmed “[t]he impossibility of protecting a small child, by any means, sitting on the lap of an adult restrained by seat belts...”
- A 2006 report[3] by Gibson, Thai and Lumley detailed the results of dynamic testing of a lap-held infant ATD restrained by a supplementary loop belt. It was found that “the forward motion of the adult dummy in a lap belt trapped and crushed the infant in the space between the front row seat back, the head and torso and the knees of the adult.”
- A 2008 report[7] by the TUV Rheinland organisation concluded that a lap-held infant is at risk of “extremely serious to fatal injuries” due to crushing by the adult and also abdominal loading by the supplementary loop belt.
- A 2009[2] report by Bathie of CASA concluded that for the case of an infant restrained by a supplementary loop belt on the lap of an adult, infant ATD injury
measurements indicated “excessive head and neck trauma” (see Figure 2-2 below). The author noted that there was no available means of measuring pertinent infant injury parameters such as compression of the head, chest and abdomen.

![Figure 2-2: Frame of high-speed footage from CASA study[2]. The head of the 50th percentile adult ATD impacts the head of the eighteen-month-old infant ATD, which impacts the forward seat-back.](image)

2.3.3 Emergency brace positions

An emergency brace position is intended to mitigate injury risk during an emergency landing by minimizing flailing of the occupant’s body[24, 25]. In the 1989 Kegworth incident (see section 2.5.2.2 below), passengers received little warning of the impending impact; some passengers adopted an emergency brace position while others did not. Using this incident as a case study, White, Firth and Rowles[14] found that the adoption of an emergency brace position provides significant protection from injury during emergency landings.

Aviation safety regulators give specific advice regarding emergency brace positions for adults carrying lap-held infants. The FAA recommends that the adult “provide as uniform support as possible to the infant's head, neck, and body, and lean over the infant to minimize the possibility of injury due to flailing”[26] without specifying the orientation of the infant. Transport Canada’s recommended position (see Figure 2-3 below) is coherent with that of the FAA, adding that the adult should place one arm around the infant and the other against the forward seat-back and that the infant should be oriented upright and facing the adult. Australian passenger safety information guidelines[27] do not explicitly describe brace positions but instead reference those of the FAA[26].
2.4 Automotive child restraints in aircraft

Automotive child restraint systems have been found to be effective in reducing the risk of death to infants in severe road accidents[29]. In automobiles the use of CRS is widely mandated by law. Their use has been found to reduce the risk of death or injury for infants in automobile accidents by 70% compared to cases where the infant is unrestrained[30].

Research into the use of CRS in aircraft was first carried out in 1978[31] and the practice has been allowed by the FAA since 1982[4]. Studies have found that CRS are capable of improving the safety of an infant aircraft passenger under emergency landing conditions compared to an infant who is seated on the lap of an adult or directly in an aircraft seat[2, 3, 6-8]. Advisory material produced by CASA[8] states that a CRS is able to provide an infant with an equivalent level of safety to that of a separately-seated adult during a severe but survivable crash. However, several issues associated with the use of automotive CRS in aircraft have been identified:
• the CRS may be geometrically incompatible with the aircraft seat[4],
• the path of the aircraft seat lap belt through the CRS may result in poorly controlled CRS motion[3, 4],
• in Australia it is mandatory to use a connecting strap between the top of the CRS and the vehicle structure (a ‘top tether’) [2], and
• the dynamic performance of CRS installed by novel methods and the corresponding effect on the safety of a passenger seated directly aft are not well understood[2, 10].

This section presents an elaboration on the above points.

2.4.1 Geometrical incompatibility
Geometrical factors affecting the compatibility of CRS with aircraft seats are twofold. The geometry of some CRS models prevents them from physically fitting into the aircraft seat[3, 4], while some CRS geometries prevent the proper tensioning of the lap belt and operation of the lap belt buckle when the lap belt is passed through the CRS[3, 4]. This issue is presently being addressed in Australia and internationally by proposed amendments to CRS design standards.

2.4.2 Dynamic performance of CRS in aircraft
Through a dynamic test program involving mostly North American convertible (i.e. able to be installed forward- or rearward-facing) and aft-facing CRS, Gowdy and DeWeese[4] found that the use of a CRS, while preferable to restraint by a lap belt only, does not necessarily ensure the safety of the infant. The dynamic performance of aft-facing CRS was found to be satisfactory; however, that of forward-facing convertible CRS generally was not. The principal reason for this was excessive forward motion of the CRS during deceleration resulting from unfavourable belt anchor geometry on the aircraft seat.

The issue of excessive motion was also apparent in dynamic testing of eleven Australian forward- and aft-facing CRS by Gibson, Thai and Lumley[3]. It was found that all tested CRS “exhibited significant forward motion, rotation and rebound motion.” Reasons given for this were that the path of the lap belt was too close to vertical and that the top tether was either ineffective or absent. Despite this, however, the authors concluded that infant air passengers are “far safer” when restrained in a CRS than if they were lap-held or restrained only by the aircraft seat lap belt.
An extended investigation by Bathie[2, 6] included dynamic testing of forward- and aft-facing Australian CRS in aircraft seats. It was found that while the lack of an effective top tether arrangement led to excessive CRS motion, the CRS tested performed adequately and afforded the infant occupant a good level of protection[2, 6].

Modern transport category aircraft seat-backs are designed to ‘break over’ in the event of rapid forwards deceleration; i.e., the seat-back rotates forward in a controlled manner. The intent of this feature is to mitigate injury risk for a passenger who flails forward and contacts the seat-back. Prior studies have identified that interaction between the CRS and seat-back may lead to increased injury to both the CRS occupant[10] and the passenger seated directly aft[2].

**2.4.3 Top tether requirement**

Australian CRS design standards[32] require the use of a ‘top tether’, a webbing strap between the CRS and the vehicle structure intended to prevent fore-aft rotation of the CRS. CASA advisory material states that a CRS “must be secured to the aircraft seat in accordance with the child seat manufacturer’s instructions or an approved alternate method”. The combination of these two requirements effectively mandates the use of a top tether in the installation of Australian-designed CRS in aircraft.

Australian air carriers that permit the use of Australian-designed CRS satisfy the top tether requirement by making a limited number of ‘anchor straps’ available for top tether attachment. The webbing anchor strap is attached to the aircraft seat leg structure; the top tether is passed over the seat-back and attached to the anchor strap. This arrangement has been shown to be largely ineffective in controlling CRS rotation when installed on a seat with seat-back break-over capability[3, 6]. Additionally, this arrangement prevents the use of the tray table by the passenger seated directly aft of the CRS[6].

Gibson, Thai and Lumley recommended in their 2006 report[3] that the effectiveness of a top tether arrangement as described above be investigated for cases where seat-back break-over is prevented from occurring. Bathie[6] suggested that “limited benefit” may be provided by increased seat-back break-over stiffness in this arrangement.

**2.4.4 Novel CRS installation methods**

In their 2006 report, Gibson, Thai and Lumley[3] recommended an investigation into the use in aviation of CRS installed by two novel methods: ISOFIX and LATCH. Rather than making
use of the vehicle seatbelt, these CRS connect directly to hard points in the vehicle structure at the location where the seat back meets the seat base.

![Image](image1.png)

**Figure 2-4:** The rigid links of the ISOFIX system[2].

The European ISOFIX standard specifies that this connection is made using rigid links (see Figure 2-4 above), while the North American LATCH standard specifies the use of webbing straps (see Figure 2-5 below).

![Image](image2.png)

**Figure 2-5:** CRS installed in an aircraft seat using the LATCH method[2].

In a 2007 report[10], Olivares and Amesar detailed the results of dynamic testing of CRS installed in transport category business jet seats using ISOFIX and LATCH. That study found that both of these attachment methods provide the CRS occupant with an appropriate level of safety while overcoming the issue of excessive CRS motion associated with installation using the aircraft seat lap belt. The authors concluded that further research was required on several aspects regarding ISOFIX and LATCH CRS, including:

- the effect of seat-back break-over on CRS performance,
- interaction with the occupant of a seat directly aft of the CRS, and
- CRS anchor loads.
An extended investigation by Bathie at CASA included CRS installed in aircraft seats using the ISOFIX[2, 6] and LATCH[2] methods as well as the aircraft lap belt[2, 6]. All three CRS installation methods were found to afford the child CRS occupant a good level of protection and the ISOFIX and LATCH methods were found to mitigate the problem of CRS motion associated with installation using the aircraft seat lap belt[2]. The investigation resulted in the recommendation of further research into the following aspects of CRS use in aircraft[2]:

- the potential for neck injury to an adult seated behind a CRS,
- the potential for tibia injury to an adult seated behind a seat which has been modified to accept ISOFIX and LATCH CRS, and
- the effects of seat pitch, CRS installation stiffness, occupant size, and seat structural variations on CRS performance and injury to both the CRS occupant and an adult passenger seated directly aft of the CRS.

2.5 Air transport accidents involving infants

2.5.1 Risk and survivability

“There are two ways to prevent fatalities in air travel: by preventing accidents, and by protecting aircraft occupants in the accidents that do occur.”

- US National Transportation Safety Board[33]

The rate of fatal accidents in commercial air transport is low. In the period 2001 – 2009 inclusive, Australian-registered aircraft operating airline services made 3.6 million flights totalling 8.4 million hours with no fatal accidents[34] (where ‘fatal accident’ refers to events not caused by an illegal act and where there was at least one fatality). In the same period, airlines in the United States made 94 million flights totalling 161 million hours with 11 fatal accidents[35]. Airline travel carries a low level of risk for both adult and infant passengers in the sense that the risk of being involved in a fatal accident is low.

The US National Safety Transportation Board studied the survivability of accidents involving US air carriers in the period 1983 – 2000[33]. Nineteen accidents were classified as serious but survivable; these were accidents involving fire or at least one serious injury or fatality and where the aircraft was either destroyed or substantially damaged. Of the 1 988 occupants involved in these accidents, 1 523 (76.6 per cent) survived, 306 (15.4 per cent) died from impact injuries, 131 (6.6 per cent) died as a result of fire, and 28 (1.4 per cent) died from other
causes. The high proportion of survivors suggests that when severe but survivable accidents occur, occupants are offered some level of protection. This is likely due to safety considerations in the design of the aircraft structure, seating and restraint systems[33].

While the risk of being involved in a fatal aircraft accident may be equally low for an infant and an adult passenger, a disparity in injury risk arises when a fatal accident does occur. A 1981 study[36] of aircraft accidents in the period 1976 – 1979 sought to determine the risk of fatal injury for an unrestrained lap-held infant relative to that of a restrained adult. Five accidents were included in the study; three in the United States and one each in Finland and New Zealand. The criteria for an accident to be included were that there was at least one survivor, at least one fatality, at least one unrestrained lap-held infant and the major cause of injury was impact rather than post-crash fire. It was found that in these five accidents an unrestrained lap-held infant was, on average, 9.6 times more likely to be fatally injured than a restrained adult[36].

### 2.5.2 Accident synopses

The last three decades' worth of air transport accident investigation reports from Australia, the United States, Canada and the United Kingdom were searched for cases involving infant passengers in severe but survivable emergency landings. Cases were determined to be of interest if there was at least one fatality, at least one survivor and at least one infant passenger whose restraint condition is detailed. A synopsis of each of those cases is presented below including, where possible, details of injuries to the infant passengers.

#### 2.5.2.1 Continental Airlines flight 1713

Continental Airlines flight 1713 crashed in Denver, Colorado on November 15, 1987[37]. Twenty-five of the 77 passengers on board were killed. A six-month-old male infant seated on his mother’s lap in the front row of the main cabin suffered fatal blunt trauma injuries; his mother survived with serious injuries. Of the five non-infant passengers seated in this row, three were fatally injured. A six-week-old female infant seated on her father’s lap in the rearmost row of the main cabin was uninjured; her father suffered minor injuries. Each of the five non-infant passengers seated in this row received minor injuries.

#### 2.5.2.2 British Midland flight 92

British Midland flight 92 crashed at Kegworth, England on January 8, 1989[38]. Forty-seven of the 118 passengers aboard were killed. There was one lap-held infant aboard, restrained by
a supplementary loop belt on its mother’s lap. The infant survived with ‘major head and limb injuries’; the mother died later in hospital from injuries sustained in the crash.

2.5.2.3 United Airlines flight 232

United Airlines flight 232 crashed in Sioux City, Iowa on July 19, 1989[39]. One hundred and ten of the 285 passengers on board were killed. Five lap-held infants were on board. At the direction of flight attendants, each infant was placed on the floor prior to impact and held there by the accompanying adult passenger. Four infants survived; one infant was thrown into a separate part of the aircraft cabin and died from smoke inhalation.

2.5.2.4 Avianca flight 52

Avianca flight 52 crashed at Cove Neck, New York on January 25, 1990[22]. Sixty-five of the 149 passengers aboard were killed. Eleven passengers were aged between four and 27 months and were classified as infants for the purposes of seating. These infants were lap-held; some were unrestrained while others were “belted into the same seat with the [adult] passengers”[22]. This may be taken to mean that the aircraft seat lap belt was fastened around both the adult and the infant. One infant, a four-month-old, received fatal blunt trauma injuries; eight sustained serious injuries such as limb fractures and head injuries while the remaining two sustained minor injuries such as contusions and abrasions. The investigation into the accident found that “[s]urviving passengers who held infants reported that during the impact the infants were ejected from their grasp and that they were generally unable to locate them in the darkness after the impact”. Investigators were unable to ascertain where in the aircraft specific passengers were seated.

2.5.2.5 USAir flight 1016

USAir flight 1016 crashed in Charlotte, North Carolina on July 2, 1994[23]. Thirty-seven of the 52 passengers on board were killed. Two lap-held infants were aboard; at the time of impact, one infant was laid across unoccupied seats while the other was held on her mother’s lap. The latter, a nine-month-old girl, was fatally injured. Her mother stated that she was “unable to maintain a secure hold on her child during the impact sequence”[23].

2.5.2.6 American Airlines flight 1420

American Airlines flight 1420 crashed in Little Rock, Arkansas on June 1, 1999[40]. Ten of the 139 passengers on board were killed. The mother of a 25-month-old child who sustained only minor injuries while restrained in a CRS stated to accident investigators that it would have
been “impossible” for her to have maintained a hold on her child had she been seated on her lap.

2.5.3 Summary

The rate of accidents in air transport is low, precluding a strong empirical analysis of the safety of various restraint conditions for infant passengers. A 1981 study of five air accidents in the 1970’s found that an unrestrained lap-held infant is 9.6 times more likely to be fatally injured than a restrained adult[36].

The present study identified six accidents in the last three decades where a total of 22 infant passengers were involved in severe but survivable air transport accidents. Information regarding the nature and severity of injury to the infant passengers was generally sparse.

In only one case was an infant restrained in a CRS: the infant and its mother, who was presumably seated next to it, both survived. The 21 remaining infants were lap-held. Seventeen survived along with the accompanying adult: during impact these infants were located either on the lap of the adult (12), on the floor (4) or lying across an unoccupied seat (1). These cases, where adult and infant both survived, are not particularly instructive because the part of the aircraft where the occupants were seated may not have undergone severe acceleration or deformation.

Four infants were fatally injured in four separate incidents; three were located on the lap of an adult during impact, while one was located on the floor. In each of these four incidents the accompanying adult survived the crash. This is evidence that lap-held infants are at risk of fatal injury in situations that are survivable by an adult.

In one case a lap-held infant survived in a part of the aircraft known to have undergone potentially fatal acceleration or deformation. In that case, the Kegworth incident, the accompanying adult was fatally injured while the lap-held infant survived with “major head and limb injuries”[38]. Of the five other occupants in the immediate area, three were fatally injured. The accompanying adult, the infant’s mother, received major injuries that were attributable to “forcible flexion” over the infant during impact. This is the only case found that may be interpreted to suggest that a lap-held infant has equal or greater survivability than an adult.
From this limited historical data it is difficult to distinguish the degree to which the various restraint scenarios influence injury outcomes for the infant, highlighting the need for quantitative analysis.
3 Occupant injury potential: measurement and quantification

3.1 Introduction
This study analyses the relative injury potential of a variety of seating configurations and, in some instances, quantifies the associated risk of injury. Numerical models of anthropomorphic test devices (ATDs) were used to calculate kinetic and kinematic quantities known as injury parameters. Established biomechanical relationships were used to quantify the risk of the occurrence and severity of injury to the adult head and infant abdomen. This chapter describes the methods used to quantify and compare occupant injury potential.

3.2 Measuring injury potential
A variety of injury parameters were measured and the results were used in three possible ways:

1. In a comparative sense to study the relative effect of a particular seating configuration parameter. While this approach is widely applicable as it does not rely on an established relationship or standard, the comparison may be trivial: for example, all values being compared may relate to injury levels above or below the survivable limit.
2. To determine whether the risk of injury is acceptable based on an injury parameter reference value specified in occupant protection standards.
3. To predict the likelihood of a particular injury outcome. This approach was only possible where a direct relationship between the injury parameter and injury level was available in the literature.

3.2.1 Occupant modelling

3.2.1.1 Anthropomorphic test devices
Anthropomorphic test devices are mechanical human surrogates designed to closely mimic properties of the human body such as size, shape, mass, stiffness and joint articulation[41, 42]. As ATDs are designed for reuse, they do not undergo permanent deformation during experiments; therefore, a post-experiment inspection cannot be used to assess the occurrence of injury. Instead, transducers are installed in specific locations in the ATD so that physical quantities such as acceleration, force, moment and displacement may be measured.
3.2.1.2 ATD biofidelity

The ‘biofidelity’ of an ATD or any other human surrogate describes the degree to which the biomechanical response of the surrogate matches that of a live human[42]. The use of ATDs in assessing compliance with vehicle occupant safety standards dictates that the criteria of reusability and repeatability outweigh the requirement for biofidelity[42].

The issue of ATD biofidelity is widely relevant[42-46]; however, its relevance to the present study particularly concerns the spine of the Hybrid III ATD and the heads of the adult and infant ATDs. The rigid nature of the Hybrid III thoracic spine leads to an under-representation of forward flexion[42]; it is likely that this aspect of the ATD’s biomechanical response had a direct influence on the results of the lap-held infant study detailed in chapter six.

Loyd[47] conducted quasistatic and dynamic tests to compare the stiffness and dynamic response of PMHS heads with ATD heads. It was found that the stiffness and dynamic response of the Hybrid III 50th percentile male ATD head is similar to that of an adult human. Also tested were ATDs representing infant and child age groups; these did not match well with the small sample of PMHS heads tested. The series of infant and child ATD tested by Loyd (Q series) was not the one implemented in the CASA study and subsequently the present study (P series); however, lacking data to the contrary, the study treated the infant and child ATD head impact response as non-biofidelic.

3.2.2 Injury criteria

Injury potential was evaluated and compared using two ‘injury criteria’ specified in transport safety regulations. An injury criterion is “a mathematical relationship, based on empirical observation, which formally describes a relationship between some measurable physical parameter interacting with a test subject and the occurrence of injury that directly results from that interaction”[48]. Injury criteria are useful in quantifying injury potential in situations where the risk of injury is a function of multiple variables, e.g. acceleration and time[49].

3.2.2.1 The head injury criterion

The head injury criterion (HIC) is widely used to evaluate head injury [3, 6, 49-58]. Proposed by Versace[49] in 1971 and later modified by the US National Highway Traffic Safety Administration, HIC essentially provides a means of evaluating a complex acceleration-time signal using the Wayne State Tolerance Curve (WSTC). The WSTC provides a relationship between peak head acceleration, pulse duration and the presence of skull fracture and/or
concussion in adults[59]. In its final form the WSTC represents data gathered encompassing a variety of acceleration pulse shapes, cadavers, animal surrogates, human volunteers, clinical research and injury mechanisms[59].

The head injury criterion is defined as[49, 53]:

\[
HIC = \left( \frac{(t_2 - t_1) \left( \frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t) \, dt \right)^{2.5}}{t_2 - t_1} \right)_{\text{max}}
\]  

(3-1)

where \(t_1\) and \(t_2\) define an arbitrary time window measured in seconds and \(a(t)\) is the time history of ATD net linear head acceleration measured at the centre of gravity and expressed in multiples of \(g\). Some occupant protection standards place an upper limit on the time window over which HIC is measured; e.g. HIC_{15} is evaluated over a maximum window of 15 ms[56].

As HIC is calculated entirely from the net linear acceleration of the head, it does not account for the effects of any applied forces that do not result in such an effect. The principal criticism of HIC is its inability to capture the potentially injurious effects of rotational acceleration[60]; however, in adults the incidence of skull fracture and the incidence and severity of brain injury have been shown to be related to the linear acceleration of the head[54, 55, 60, 61]. A study by Loyd[47], which included a small number of infant cadavers, found that neither linear acceleration nor HIC is a good predictor of skull fracture in infants.

3.2.2.2 The neck injury criterion

The neck injury criterion, \(N_{ij}\), is widely used to evaluate neck injury potential[6, 50, 56, 59, 62-64]. This criterion considers the combined effects of fore-aft bending moment (flexion and extension) and axial force (tension and compression)[65] normalised against critical values which are uniquely specified for various ATDs[52] (see section 3.4.2).

The neck injury criterion is defined as:

\[
N_{ij} = \frac{F_z}{F_{zc}} + \frac{M_y}{M_{yc}}
\]

where \(F_z\) and \(F_{zc}\) are the measured and critical upper neck axial forces, respectively, and \(M_y\) and \(M_{yc}\) are the measured and critical upper neck fore-aft bending moments, respectively, at each instant in time during the experiment. The critical value of \(N_{ij}\) is 1.0[56].
3.2.3 Injury risk curves

The present study used logistic regression curves available in the literature which describe the relationship between injury parameters and the probability of the occurrence of a particular injury. These injury risk curves were developed through a program of physical tests and post-test inspections of post-mortem human subjects[54, 55, 61] or animal surrogates[66]. This method of analysing injury potential is applicable to studies such as this where ATDs are used to measure injury parameters[52, 64, 67].

Some of the biomechanical studies referred to below make use of the abbreviated injury scale (AIS)[68] as a means of classifying the extent of trauma. The AIS uses a six-point ordinal scale to classify specific injuries to individual parts of the body:

1. Minor
2. Moderate
3. Serious
4. Severe
5. Critical
6. Maximal (currently untreatable)

The AIS has been widely used to relate injury parameters measured using human surrogates to the risk of a particular injury outcome[54, 55, 60, 61, 66, 69]. In situations where the most-severe of a spectrum of potential injuries is of interest, the maximum AIS (MAIS) classification is used.

3.3 Biomechanical relationships

3.3.1 Infant injury potential

The literature is replete with investigations into the biomechanical tolerance of adults; however, the body of such information regarding infants is small[70]. One consequence of this is that the biomechanical tolerance data needed to form a relationship between an injury parameter observation and the risk of a particular injury generally does not exist for the infant[70, 71]. The evaluation of infant injury potential must instead rely on either critical injury parameter values scaled from adult values or anatomical finite element modelling – a method that remains at the forefront of research in infant injury biomechanics[71].
3.3.2 Head injury

The potential for injury to the infant head was paid particular attention as it has previously been identified that the head of the lap-held infant may be compressed between the adult head and the forward seat-back during the impact sequence[2].

The skull is responsible for protecting the brain from trauma[58, 72, 73]. Compared to the adult, the developing skull of the infant is compliant[21] due to reduced material stiffness[74] and structural immaturity[73]; the stiffness of the newborn’s skull is approximately 4% of the adult’s, increasing to 75% at six to eight years[75]. This results in the potential for substantial deformation under external load[72] and the consequence of brain deformation[74] and associated injury[76].

3.3.3 Abdominal injury

An effort was made to quantify abdominal injury risk for the lap-held infant in spite of a lack of biomechanical data specific to the ages of infants considered by this study. This necessitated the generalisation that a loading condition giving rise to a particular injury risk in a subject of a certain age will result in an equal or greater risk of the same injury in a younger subject. This generalisation was predicated on the facts that the growth and development of the human body is a continuous process from birth to old age[21] and that the tissues comprising the body are strongest at approximately twenty years of age[21].

The so-called ‘seat belt syndrome’ is an injury mechanism observed in motor vehicle accident victims restrained by a lap belt. It occurs when the lap belt is positioned over the abdomen and the restraining force is applied to the body through the abdomen rather than the bony pelvis, resulting in blunt trauma[48]. The belt may “intrude into the soft abdomen and rupture or lacerate internal organs”[77], or, in severe cases, cause a fracture of the lumbar vertebrae as the spine flexes about the belt[77]. This injury mechanism is particularly relevant in the case of the lap-held infant due to the positioning of the supplementary loop belt over the abdomen[7].

Kent et al.[66] studied the risk of abdominal injury in children due to loading by a lap belt. Using juvenile swine as a surrogate for the six-year-old child, the study found peak belt tension to be a strong discriminator for the level of abdominal injury sustained. Forty-six subjects were dynamically loaded by a 50 mm wide belt oriented laterally across the abdomen. Thorough internal post-mortem examinations were conducted and injuries were assigned AIS values. The maximum AIS (MAIS) score sustained by each of the forty-six subjects ranged
from zero (not injured) to four (severe). Injury risk curves were developed for a variety of mechanical loading parameters including peak belt tension ($F_b$).

### 3.4 Particular injury parameters and reference values

This section identifies and describes the injury parameters used and the values obtained from occupant protection standards and literature used to evaluate them. Occupant protection standards specify limits on injury parameter or injury criterion values as measured by a particular ATD type under a prescribed loading condition. The FAA specifies occupant protection standards for transport category aircraft under emergency landing dynamic conditions in Federal Aviation Regulation (FAR) 25.562[53]. The only occupant size considered by this standard is a 50th percentile adult male for which head acceleration (HIC) and femur compression limits are specified. Femur compression was not evaluated as this parameter did not reach a high level of agreement with physical test results during the model validation process (see chapter 5).

Occupant protection standards for motor vehicles fitted with airbags are specified in US Federal Motor Vehicle Safety Standard (FMVSS) 208[56]. Occupant protection standards for child restraint devices for use in motor vehicles and aircraft are specified in FMVSS 213[57]. These standards consider a range of occupant sizes.

The end use of model results for these varies according to the outcome of the model validation process described in section 5.8.

#### 3.4.1 Head injury

In FAR 25.562 a maximum allowable HIC value of 1000 is specified for a 50th percentile adult male ATD; this is the only occupant size for which HIC is evaluated under this standard. No upper limit is placed on the time window; however, HIC is only evaluated during contact between the head and cabin structure.

In FMVSS 213 a maximum allowable HIC$_{36}$ value of 1000 is specified for a variety of child ATDs representing ages from newborn to six years. The criterion is evaluated over the entire test period, not only during head contact.

In FMVSS 208 different HIC$_{15}$ values are specified for a range of occupant sizes from the twelve-month-old to the 50th percentile adult male. For the latter a HIC$_{36}$ limit of 1000 is also specified. Again, $a(t)$ is defined over the entire test period, not only during head contact.
The HIC$_{15}$ limits specified in FMVSS 208 for a range of occupant sizes are presented in Table 3-1 below. The values specified for the infant and child occupants were arrived at through the application of a variety of scaling techniques. These techniques took into account cranial bone stiffness, brain bulk modulus, head size and, in the case of the 12-month-old, cranial suture stiffness[52].

Table 3-1: Head injury criterion (HIC$_{15}$) limits defined by FMVSS 208.

<table>
<thead>
<tr>
<th>Occupant size</th>
<th>HIC Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>50$^{th}$ percentile adult male</td>
<td>700</td>
</tr>
<tr>
<td>5$^{th}$ percentile adult female</td>
<td>700</td>
</tr>
<tr>
<td>6-year-old child</td>
<td>700</td>
</tr>
<tr>
<td>3-year-old infant</td>
<td>570</td>
</tr>
<tr>
<td>12-month-old infant</td>
<td>390</td>
</tr>
</tbody>
</table>

Prasad and Mertz[78] studied adult cadaver data of forehead impacts to develop logistic regression curves relating HIC$_{15}$ score to the risk of skull fracture and brain injury of AIS classification ‘severe’ or greater[79]. This level of brain injury is described by Mertz and Irwin[54] as ‘life-threatening’. These relationships, presented in Figure 3-1 and Figure 3-2 below, were used to quantify head injury potential arising from forehead impact for the adult. This method of quantifying head injury potential is particularly instructive as it accounts for the nonlinear relationship between HIC score and injury potential.
Figure 3-1: Head injury risk curve relating HIC₁₅ score to the probability of life-threatening brain injury (AIS ≥ 4) in adults[79].

Figure 3-2: Head injury risk curve relating HIC₁₅ score to the probability of skull fracture in adults[79].
In the analysis phase, HIC_{15} values were calculated to enable comparison of results with FMVSS 208 limits and the evaluation of head injury risk according to the curves presented above.

### 3.4.2 Neck injury

The FMVSS 208 occupant safety standard specifies critical neck loads for a range of ATD sizes, presented in Table 3-2 below. Critical axial loads are defined as well as fore-aft bending moments for both the flexion (chin down) and extension (chin up) cases. Four combinations must therefore be considered in order to find the critical $N_{ij}$ case: tension-flexion, tension-extension, compression-flexion and compression-extension. The critical value of $N_{ij}$ is 1.0 by definition[56].

The critical loads specified in FMVSS 208 and presented below for the 50th percentile adult male and three-year-old child were arrived at through physical experiment[52]. This data was scaled based on neck circumference and tendon failure strength to produce values for the other occupant sizes[52].

**Table 3-2: Neck injury criteria critical values and peak neck axial loads specified in FMVSS 208[56].**

<table>
<thead>
<tr>
<th>Occupant size</th>
<th>Tension (N)</th>
<th>Compression (N)</th>
<th>Flexion (Nm)</th>
<th>Extension (Nm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>50th percentile adult male</td>
<td>6806</td>
<td>6160</td>
<td>310</td>
<td>135</td>
</tr>
<tr>
<td>5th percentile adult female</td>
<td>4287</td>
<td>3880</td>
<td>155</td>
<td>67</td>
</tr>
<tr>
<td>6-year-old child</td>
<td>2800</td>
<td>2800</td>
<td>93</td>
<td>37</td>
</tr>
<tr>
<td>3-year-old infant</td>
<td>2120</td>
<td>2120</td>
<td>68</td>
<td>27</td>
</tr>
<tr>
<td>12-month-old infant</td>
<td>1460</td>
<td>1460</td>
<td>43</td>
<td>17</td>
</tr>
</tbody>
</table>

While the neck injury criterion was evaluated by the present study, it was used in a comparative sense only; the validation process revealed an inconsistent level of agreement between the model prediction of relevant neck injury parameters and corresponding physical test results.

### 3.4.3 Thoracic injury

The potential for inertial and contact-based thoracic injury was considered in the lap-held infant using thoracic acceleration and contact force as proxies for injury potential. Thoracic acceleration has been implicated as a possible factor in potentially fatal injuries such as traumatic aortic rupture[80]. While thoracic acceleration limits are specified in vehicle and CRS occupant protection standards[56, 57], the present study considered values in a
comparative sense only as the validation process identified that numerical model prediction of thoracic acceleration is in poor agreement with physical test results.

### 3.4.4 Abdominal injury

The potential for a bdominal injury was considered in the lap-held infant. Two parameters were measured: the applied abdominal contact force and the tension in the supplementary loop belt, where present. The former was not included in the model validation process and was therefore evaluated in a comparative sense only. Supplementary loop belt tension exhibited a high level of agreement with physical test results; this parameter was therefore suitable for use in quantifying the risk of abdominal injury in restrained lap-held infants according to the injury risk curve developed by Kent et al.[66].

Figure 3-3 below describes the probability of MAIS 3 (serious) or greater injury as a function of peak belt tension. Fine lines represent the 95% confidence interval; circles represent experimental data points.

![Figure 3-3: Risk function for abdominal injury MAIS ≥ 3 using juvenile swine as a surrogate for a six-year-old child. [66]](image)

Abdominal injuries at the AIS 3 level observed during post-mortem analyses by Kent et al.[66] included perforation and laceration of the small bowel as well as large bowel injuries resulting in gross faecal contamination of the peritoneal cavity and significant blood loss into the same site. Rib fractures at the AIS 3 level were found in five of 46 subjects.
4 Aircraft cabin safety model: design and development

4.1 Introduction

A numerical model of a typical air transport seating arrangement was developed for use as an experimental platform. Numerical modelling has been widely used in biomechanical studies[71]. This method was chosen over physical experiments for its ability to predict important parameters such as contact force[15]. In addition, numerical analysis is well suited to sensitivity studies permitting relatively low cost per experiment[71]. The numerical model was validated using sensor signal data and high speed video footage from physical experiments carried out by CASA[2].

To ensure that the model would accurately represent contact and structural deformation, video footage was used to determine where contact occurs between the aircraft seats, ATDs and CRS and also where non-permanent deformation occurs in the aircraft seats and CRS. The post-experiment aircraft seats and CRS were examined to identify areas of significant permanent deformation.

A reverse engineering process was carried out on undeformed aircraft seats and CRS to collect joint characteristics and geometry and material data for input to the model. More complex components such as the seat base cushion and the seat-back frame were scanned using a three-dimensional stereo optical scanner. Specific materials and components which were identified as having a significant effect on the dynamic response of the aircraft seat and its occupants, such as the foam cushion material and the tray table, were subjected to mechanical testing.

Material and component behaviour were verified and validated through the replication of corresponding physical experiments using the numerical model. The geometry, material and component data were then combined in the model assembly process to arrive at a baseline aircraft cabin model.

4.2 Analysis of physical tests

All information on the physical experiments was provided by CASA in the form of raw sensor data, high speed video footage and technical reports[2, 6]. Aspects of the CASA physical experiments that are relevant to this study are described below.
4.2.1 Seat and occupant configuration

The seat configuration implemented by CASA was similar to that found in a typical airline economy class seating configuration. Two rows of seats at 30 inch pitch were installed on a deceleration sled in a two-abreast arrangement as in Figure 4-1 below.

![Figure 4-1: Example of physical experiment configuration. Two adult ATDs were seated directly aft of two CRS occupied by infant ATDs[2].](image)

Seats were empty, occupied by an ATD or occupied by both a CRS and an ATD. The majority of experiments employed one of two occupant arrangements:

1. A CRS with an infant ATD occupant was installed in seat directly aft of an empty seat.
2. A CRS with an infant ATD occupant was installed in a forward seat with an adult ATD occupying the seat directly aft.

Different arrangements were employed in a limited number of tests:

3. An infant ATD was seated on the lap of an adult ATD and restrained by a supplementary loop belt.
4. An infant was seated in its own seat and restrained by the aircraft seat lap belt only.
5. An adult occupant was seated directly aft of an unoccupied seat.

In accordance with FAR 25.562, adult occupants were placed in a normal upright sitting position. Numerical modelling was not a consideration at the time that the physical tests were being carried out; as a result, some important parameters such as ATD initial position, CRS initial position and initial lap belt tension were not measured.
4.2.2 Load case

FAR 25.562 specifies longitudinal and vertical load cases for the dynamic testing of aircraft seats and restraint systems. The longitudinal load case includes the requirement for a minimum peak floor deceleration of 16 g to be achieved no more than 90 ms after impact and a minimum forward velocity change of at least 44 ft∙s⁻¹ (13.4 m∙s⁻¹). The same load case is referred to by FAA Technical Standard Order (TSO) C100b, which defines minimum performance standards for CRS approved for use on aircraft in the United States.

Limitations of the deceleration sled system used in the CASA test program prevented the application of the ideal pulse specified in FAR 25.562 (see Figure 4-2 below). The load case applied in the CASA tests typically resulted in a forward velocity change of approximately 14.5 m∙s⁻¹ with a 21 g peak floor deceleration 50 ms after impact.

![Figure 4-2: Comparison of typical sled deceleration pulse from physical testing and minimum allowable pulse under FAR 25.562.](image)

4.2.3 Aircraft seat design

The seats used in physical testing were typical economy class airline seats which had been removed from service, model BA3.4-2-41. They were labelled as meeting the standards of FAA Technical Standard Order C39b[81], Type I, which does not include dynamic performance or occupant protection criteria. The seat base cushions were a non-flotation type comprising of a laminate of three different polyurethane foams. The seat design incorporated a break-over feature (see section 2.4.2) comprising an energy-absorbing device (see Figure 4-9) located on the hinge of one side of each seat position; the other hinge was unconstrained.
Lower anchorage points compatible with the ISOFIX and LATCH CRS installation methods were added to the aircraft seats for the purpose of the CASA study. The anchorages were small loops of 6 mm round steel bar welded to a length of 20 mm diameter, 2 mm thick circular steel tube. This tube spanned the width of both seat places and was fastened to the aft edge of each of the three spreaders at a height of 395 mm above the floor.

![Lower anchorage points added to the aircraft seats as part of the CASA study](image)

**Figure 4-3: Lower anchorage points added to the aircraft seats as part of the CASA study[2].**

### 4.2.4 Child restraint systems

The CASA study considered forward- and aft-facing CRS models as well as three different CRS installation methods: the lap belt, ISOFIX and LATCH (see Figure 4-4 below). However, this study considered only the forward-facing orientation as high-speed footage of the physical tests revealed that forward-facing restraints affect the dynamic behaviour of the aircraft seat to a greater extent than aft-facing models.

Of the forward-facing models tested, one CRS was able to be attached to the aircraft seat by each of the three installation methods of interest. The Britax Duo Plus is fitted with ISOFIX rigid linkages and also has a traditional belt path that provides for installation using the aircraft seat lap belt or a separate LATCH webbing strap. For this reason the Britax Duo Plus was chosen as the only CRS model to be considered by this study. This CRS has a mass of approximately 8.5 kg and an allowable occupant mass range of nine to 18 kg.
4.2.5 Anthropomorphic test devices

The adult ATDs used by the CASA study were Hybrid III models representing the 5th percentile female, 50th percentile male and 95th percentile male. The infant ATDs used were the TNO P3 (three-year-old), P1.5 (eighteen-month-old), and P¾ (nine-month-old). The TNO P¾ ATD was not instrumented. The ATD joints were calibrated by technicians at the test facility according to manufacturer specifications prior to use.

4.3 Model development

The numerical model development process was carried out in the following stages:

1. Geometry measurement and meshing.
2. Material and component mechanical testing.
3. Component connectivity and contact interaction definition.
4. Verification of model behaviour and validation of model output.

Two modelling methods are applicable to the type of model required by this project: the multibody method and the finite element method. The principal difference between these methods is in their representation of geometry, with different treatments of mass distribution, contact interaction, component connectivity and deformation as a consequence.

In general, a multibody model is less computationally-intensive than a finite element model. In a multibody model the geometry of a structure is represented by discrete ellipsoidal, cylindrical and planar surfaces[42]. These surfaces are attached to rigid bodies to which the mass and inertial properties of the part are assigned. The degrees of freedom of the rigid bodies relative to each other and to the environment (the ‘reference space’) are defined analytically, thus creating joints between parts. Multibody surfaces are non-deformable; contact interaction behaviour between multibody parts is modelled using a known force- or stress-based
characteristic. For these reasons multibody models are not well suited to representing complex geometries and are unsuitable for use in situations involving large-scale deformation.

The finite element method is able to represent infinite variations in the geometry of a structure, defined at finite points called nodes[42]. Nodes are connected to one another to form elements; multiple connected elements form a part. By assigning a material property to an element, the nodes are given mass and the stiffness and damping relationship between the nodes is defined. Joints may be modelled by defining relative motion between nodes with an equation or by defining contact interaction characteristics between elements.

4.3.1 Solver selection
The MADYMO[82] software package was selected for its emphasis on vehicle occupant safety analysis and its integrated library of numerical ATDs. The MADYMO solver is a combined finite element and multibody solver, enabling the user to maximise computational efficiency by using the multibody method to represent parts of the model where appropriate. This solver has been employed previously by similar studies[51, 62, 63, 83-90] and is approved by the FAA for aircraft seat dynamic certification[91].

4.3.2 Method
The model was designed to be as simple and modular as possible while maintaining fidelity in critical areas to enable reconfiguration of the model and maximise computational efficiency. Aircraft seat and CRS geometry and material properties were measured in a process of ‘reverse engineering’. Measurements of complex geometries were made using an articulated-arm coordinate measurement machine and three-dimensional optical scanning system (see Figure 4-5 below). Material properties were either gathered from literature or measured by physical testing.
4.3.2.1 Mesh convergence

During testing of an initial model assembly, the aircraft seat base cushion and tray table were found to undergo significant deformation and also greatly influence the head acceleration signal of an adult ATD. These two components were subjected to a convergence test to avoid a mesh that was unnecessarily fine, and therefore computationally intensive, while ensuring that the mesh size was not influencing model output. For both components a simple physical test was conducted and replicated in MADYMO; this replica simulation was run multiple times with increasingly finer mesh while checking the output parameter for convergence. The cushion was tested in quasi-static compression and the force-displacement curve was analysed (see section 4.3.2.2 below for details). The tray table was subjected to a drop test (see Figure 4-6 below) with the acceleration of the impactor as the output parameter.
A suitable balance between accuracy of output and computational requirements was found with element lengths of 14 mm and 15 mm in the tray table and seat base cushion, respectively.

### 4.3.2.2 Foam materials

The material characteristics of the aircraft seat cushion foams were determined by mechanical testing. Specimens provided by the cushion manufacturer were tested at three quasi-static loading rates using a method derived from ASTM D3574\[92\]. Square foam samples of side length 380 mm and varying thickness were compressed on a solid surface by a solid round anvil of 200 mm diameter.

Testing revealed that all three foam materials used in the seat design exhibited rate-dependent behaviour; however, the maximum tested loading rate of 500 mm·min\(^{-1}\) was estimated to be
significantly below that experienced by the material in sled testing. An assessment of similar materials in quasi-static and dynamic loading by Bhonge[93] provided the basis for a rate-dependent stress scaling characteristic which was then applied in the model (see Appendix A).

![Figure 4-8: Comparison of foam material model behaviour in loading, physical material test and MADYMO simulation. LRGR45 foam specimen, 100 mm thickness, 500 mm·min⁻¹ loading rate.](image)

The physical material test was replicated in MADYMO for verification and validation (see Figure 4-8 above). Properties were defined using the MADYMO MATERIAL.FOAM characteristic. In the foam materials a 15 mm element length resulted in a favourable combination of both computation time required and degree of match with physical test data. Properties of the three foam materials are presented Appendix A.

### 4.3.2.3 Joints

Joints were modelled by rigidly attaching the nodes of each side of a joint to a rigid body; a numerical joint was then defined between the corresponding rigid bodies. This method was chosen to avoid computationally-intensive contact evaluations while enabling precise control over joint behaviour. A particular example of this approach is the modelling of the energy-absorbing device fitted to one seat-back hinge of each seat position. This device comprises two steel plates which plastically deform as the seat-back rotates forward. A basic test rig
(Figure 4-9) was manufactured to enable the resistance torque caused by the buckling plates to be measured as a function of seat-back angular displacement.

![Figure 4-9: Seat-back energy absorber test apparatus, post-test.](image)

The resulting characteristic curve (Figure 4-10) was used to define a restraint on the motion of the left seat-back hinge. Hysteresis was applied to account for energy dissipation due to plastic deformation.

![Figure 4-10: Seat-back energy absorber loading characteristic.](image)
4.3.2.4 Contact

Contact between several surfaces was defined in MADYMO. The method used depended on the contacting surfaces were defined by finite elements or multibody objects. For contact between two multibody surfaces (e.g. ATDs), CONTACT.MB_MB was used. Where one of the surfaces was a part of the Hybrid III 50th percentile male ATD, that surface was used as the master. For contact between a multibody surface and an FE surface, CONTACT.MB_FE was used. Here, the multibody surface was always defined as the master. Contact between two FE surfaces was modelled using CONTACT.FE_FE and the adaptive node-to-surface method.

Contact involving a multibody surface is modelled using the ‘penalty’ method, whereby surfaces may penetrate one another to a degree commensurate with the contact force being generated. This effect takes into account, to some extent, the non-deformability of multibody objects. It can be seen in Figure 4-14 below where the CRS harness passes slightly below the infant ATD’s upper torso and shoulder surfaces.

4.3.2.5 Belts

Webbing belts may be modelled in MADYMO using the multibody or finite element methods. Using the multibody method one-dimensional belt segments with a defined force-strain characteristic were used to connect the belt anchor points to the ATD or CRS. This avoided the requirement for computationally-intensive contact calculations associated with finite element belts. The aircraft seat lap belt was modelled using this method as the belt path through the CRS and across the pelvis of the ATD is simple and remained constant during physical tests.

The finite element method was used to model the CRS five-point harness, supplementary loop belt and top tether. In these components belt geometry and interaction with the ATD are more complex than in the case of the lap belt.

4.3.2.6 Aircraft seat

To reduce model complexity and maximise computational efficiency, only the left-hand seat position of the two-abreast configuration was modelled. A symmetry condition was imposed at the centre spreader to maintain correct structural behaviour. The aircraft seat model is fully deformable with the exception of the arm rests. This component was found not to deform significantly during physical testing so were modelled as rigid parts to maximise computational efficiency.
4.3.2.7 Child restraint system

A review of the CASA footage and a post-test inspection of the Britax Duo Plus CRS (see section 4.2.4) revealed that it does not undergo significant deformation during the impact sequence. With the exception of its five-point harness, the CRS was modelled as a rigid part in order to minimise computation time and model complexity (see Figure 4-12 below).

The geometry of the CRS was defined by two surfaces comprising of three- and four-node shell elements. The thin foam padding that covers the CRS seating surface was assigned a force-based contact characteristic using the LRGR45 foam relationship presented in Appendix A.
4.3.2.8 Anthropomorphic test devices

The numerical ATDs used were multibody representations of the ATDs used in physical testing. In addition to geometrical features, numerical ATDs have pre-assigned contact characteristics and joint stiffness characteristics closely matching the physical ATD they represent. The numerical ATD models implemented by the present study were included with the MADYMO software package. Numerical ATD validation is described in section 5.9.

4.3.3 Model assembly

The model was assembled beginning with the sled, a multibody plane. As each new component was added a short simulation was run to verify joint behaviour. The resulting numerical model of a single aircraft seat position consisted of 37 discrete parts defined by 15873 elements (see Table 4-1 below). The two armrests were defined by a total of six ellipsoid surfaces.

<table>
<thead>
<tr>
<th>Element Type</th>
<th>Description</th>
<th>No. Elements</th>
</tr>
</thead>
<tbody>
<tr>
<td>BEAM2</td>
<td>Two-node beam</td>
<td>64</td>
</tr>
<tr>
<td>HEXA8</td>
<td>Eight-node hexahedral</td>
<td>4706</td>
</tr>
<tr>
<td>QUAD4</td>
<td>Four-node quadrilateral</td>
<td>11008</td>
</tr>
<tr>
<td>TRIA3</td>
<td>Three-node triangular</td>
<td>95</td>
</tr>
</tbody>
</table>

Where there were multiple options for a given element type, for example three- and six-node triangular elements, preference was given to the least computationally-intensive. Solid elements expected to be involved in contact were covered in a layer of thin shell elements to ensure correct contact behaviour; this was necessary in the cases of the seat cushions and the tray table. Parts were typically meshed with a characteristic element length of between five and ten millimetres.

The solution time step of $5 \times 10^{-7}$ seconds determined by MADYMO resulted in an acceptable solution time of approximately five hours; the mass scaling technique was deemed unnecessary.

Once the general arrangement of the model components was complete, a pre-simulation was used to position the ATDs and CRS in the aircraft seats. This step was necessary to ensure an equilibrium condition in elements, joints and contacts.
With an initial position slightly above the seating surface, the ATDs and CRS were released with an applied acceleration field of 1 g downwards. The simultaneous application of 0.1 g acceleration aft was found to result in a seating position similar to that observed in the CASA high-speed footage.
4.3.3.1 Aft passenger initial position

Video footage of the physical tests indicated that there was generally little variation in adult ATD initial position between tests. However, a small variation in knee angle was noted in one case (see Figure 4-15). The effect of this was assessed using the model and was found to be significant: as the ATD slides forward on the seat, the tibias make contact with the forward seat structure. The tibias are thereby loaded in three-point bending between the aircraft floor, the forward seat and the knees which are moving forwards at the time of contact. As this loading effectively jammed the tibias between the forward seat and the floor, the upper leg and upper body then pivoted about the knee. A small variation in initial knee angle was found to change this pivoting behaviour and consequently the location of head impact.

![Figure 4-15: Variation in aft passenger initial knee angle.](image)

The aft passenger knee angle used in the model validation process resembled the corresponding physical test as closely as possible. For all analyses, however, the more-common knee angle depicted in the upper image of Figure 4-15 was used.
4.3.3.2 CRS initial position

Video footage of the physical tests revealed that for each of the three CRS installation methods tested the initial compression of the aircraft seat base cushion was slightly different (see Figure 4-16 below). While seemingly insignificant, these differences were found to noticeably affect the model's prediction of CRS behaviour. These CRS initial positions were treated as a property of the associated installation method and so were maintained in the model.

![Figure 4-16: CRS initial positions relative to seat base cushion.](image-url)
5 Aircraft cabin safety model: verification, validation and implementation

5.1 Introduction

This chapter describes the process of verification and validation applied to the model through the replication of eight of the experiments carried out by the CASA study. Verification is the process of ensuring that the input variables (such as geometry and material properties) and model configuration parameters (such as joint and contact definitions) are being implemented correctly[94]. A continual cycle of verification was implemented during model development at the material, component and assembly levels. The MADYMO solver is an established numerical code that has already been subjected to a process of verification and validation by its developer.

Validation is the process of comparing model output with physical test data in order to determine the degree to which the output from a numerical model represents reality[95]. The purpose of the validation process was to ensure that the end use of a particular model output parameter was appropriate to the level of agreement observed in that parameter.

Model behaviour was verified through visual analysis of kinematics, energy balance plots and floor reaction loads (for further details see Appendix B). Some further checks were performed on the numerical model prior to the formal validation process. Hourglass energy was analysed and found to reach a steady state of 1.4% of total internal energy after an initial peak of 8%. This is satisfactory as it is within the 10% rule-of-thumb generally applied to such models[93].

The mass of the aircraft seat finite element model for a two-abreast configuration was found to be approximately 700 grams lighter than the original seat (56.99 lb, 25.85 kg). This difference of approximately 3% is likely due to the steel hardware such as fasteners and hinges being represented as numerical joints in the finite element model. No further data, such as centre of gravity or moment of inertia, were available from the aircraft seat manufacturer when requested.

5.2 Model execution

Analyses were run on the RMIT University high-performance computing cluster. A typical analysis required approximately five hours to solve on a single node consisting of two quad-core AMD Opteron 2.3 GHz CPUs and 32 Gb RAM.
5.3 Load case

The deceleration pulse applied in all model validation and analysis was representative of a typical pulse applied in physical testing, with a peak of 21.5 g occurring at 43 ms and net velocity change of 14.5 m s$^{-1}$ (see Figure 5-1 below).

The pulse was applied as a prescribed acceleration of the joint connecting the sled to the reference space with roll, pitch and yaw values of zero. The sled floor was undeformed.

![Figure 5-1: Sled longitudinal deceleration pulse applied in all model experiments (forwards positive).](image)

Though this acceleration pulse differs from the ideal pulse set out in aircraft design standards[53] (see Figure 4-2), it was considered throughout the present study to be representative of a severe but survivable emergency landing. For further detail see section 4.2.2 above.

5.4 Signal processing

It was necessary to process the inherently noisy raw data produced by the CASA study before it could be used in any analysis. Sensor signals were filtered according to the Society of Automotive Engineers (SAE) standard J211-1[96]. Sensor signals not identified in SAE J211-1 were processed using the same type of filter with parameters chosen to minimise noise without significantly modifying the signal.

Where peak signal values were used, the value taken was the ‘three millisecond clip’ value. This is the highest value that the signal reaches for a cumulative period of at least three
milliseconds. Vehicle occupant protection standards[56, 57] make use of this method of determining a signal peak.

5.5 Validation metrics

In many instances the validation process required the comparison of two transient time signals. Signal properties such as the time and magnitude of the peak value are useful but only consider a single point in time. While there is no single best method for approaching this complex task, the Sprague and Geers error metric[97] has been identified as being appropriate to this type of application[98] and has been found to be in good agreement with subjective assessments by subject matter experts[99].

Four graphical examples of the validation metrics outlined below are presented in Appendix C.

5.5.1 Whole-signal comparison

The Sprague and Geers metric[97] individually assesses the phase and magnitude error between two signals over a given period; these two error values are then combined into a single comprehensive error factor. The metric is based on the time integrals of a benchmark (physical test) signal \( b(t) \) and a candidate (numerical model) signal \( c(t) \) over a time period \( t_i < t < t_f \).

The dimensionless magnitude error factor \( M_{SG} \) and phase error factor \( P \) are defined as:

\[
M_{SG} = \sqrt{\frac{\vartheta_{cc}}{\vartheta_{bb}}} - 1 \quad (5-1)
\]

\[
P = 1 - \frac{\vartheta_{cb}}{\sqrt{\vartheta_{cc} \vartheta_{bb}}} \quad (5-2)
\]

where

\[
\vartheta_{bb} = (t_2 - t_1)^{-1} \int_{t_1}^{t_2} b^2(t)dt \quad (5-3)
\]
\[
\theta_{cc} = (t_2 - t_1)^{-1} \int_{t_1}^{t_2} c^2(t)dt \\
\theta_{cb} = (t_2 - t_1)^{-1} \int_{t_1}^{t_2} c(t) \cdot b(t)dt
\]

These magnitude and phase error factors are combined to give the comprehensive error factor, \( C \):

\[
C = \sqrt{M_{SG}^2 + P^2}
\]

The comprehensive error factor was used to quantify the level of agreement between the output of the numerical model and physical test results. It is presented in the form of a percentage for convenience. A value of zero for \( C \) indicates a perfect match between signals; greater values of \( C \) indicate a poorer level of agreement between the benchmark and candidate signals. With reference to the equations presented above it can be seen that \( C \) is unbounded; i.e. an extremely poor match between signals can result in a value greater than 100%.

### 5.5.2 Peak values

The comprehensive error factor \( C \) does not directly take into account the difference between peak signal values. Peak values, both positive and negative, are particularly important properties of force and moment signals. Physical test and numerical model signal peak values were compared by expressing their difference as a percentage of the physical test value. Where a positive (+) peak exists in the physical test signal, the magnitude error \( (M) \) between the positive peak values is given by:

\[
M_{pos} = \left(1 - \frac{\text{Peak}_{model(+)}}{\text{Peak}_{test(+)}}\right) \times 100
\]

Similarly, where a negative (-) peak exists in the physical test signal, the magnitude error between the negative peak values in is given by:
Where signals exhibit both positive and negative peaks, weighting factors \( W \) were used to determine the relative importance of each peak according to the physical test. These factors take into account the magnitude of the positive and negative peaks as a proportion of the total envelope of the physical test signal and are given by:

\[
W_{pos} = \frac{\text{Peak}_{test}(+) - \text{Peak}_{test}(-)}{\text{Peak}_{test}(+) - \text{Peak}_{test}(-)} \times 100 \quad (5-9)
\]

\[
W_{neg} = \frac{\text{Peak}_{test}(-) - \text{Peak}_{test}(+)}{\text{Peak}_{test}(+) - \text{Peak}_{test}(-)} \times 100 \quad (5-10)
\]

A low value of \( W \) (i.e. close to zero) indicates that a peak is relatively insignificant, a high value of \( W \) (i.e. close to 1) indicates that a peak is significant, while a value of \( W \) close to 0.5 indicates that the positive and negative peaks are of equal significance. This weighting does not take into account whether a positive or negative peak is more significant in terms of injury potential.

### 5.6 Validation parameters

At the time physical testing was undertaken by CASA, later numerical modelling was not taken into consideration. Data describing some potentially useful validation parameters, such as floor anchor load, were not available. Of the parameters measured in physical testing, the following were deemed relevant to the end use of the model and were used in model validation:

1. Head acceleration (aft passenger and infant)
2. Upper neck axial force (aft passenger)
3. Upper neck moment in the longitudinal vertical plane (aft passenger)
4. Thoracic acceleration (infant)
5. Supplementary loop belt tension

Signals were validated in terms of comprehensive error factor and peak value(s).
Strong emphasis was placed on the head acceleration signal. This parameter was the closely matched of all ATD signals in numerical ATD validation (see section 5.9) and it is sensitive to a range of model parameters:

- Whole-body ATD kinematics
- Lap belt properties
- Tray table structural and material properties
- Seat base and back cushion material properties
- Seat-back break-over behaviour

For model configurations designed to measure the baseline performance of the CRS (i.e. with no passenger seated aft), the infant ATD head acceleration signals are equally useful in assessing the global response of the model.

The time period used for all signal validation was the full 150 ms numerical model test period, giving a good indication of the overall match between physical test and model output data.

### 5.7 Validation configurations

Eight of the configurations used in CASA physical testing were replicated. These were:

- Three CRS baseline configurations, where an occupied CRS was installed aft of an empty seat using the lap belt, ISOFIX and LATCH methods.
- Three aft passenger configurations, where a Hybrid III 50th percentile male ATD was seated behind an occupied CRS installed using the lap belt, ISOFIX and LATCH methods.
- An aft passenger baseline configuration, where a Hybrid III 50th percentile male ATD was seated directly aft of an empty seat.
- A configuration in which an eighteen-month-old infant ATD was seated on the lap of a Hybrid III 50th percentile male ATD and restrained using a supplementary loop belt.

The seat pitch in each validation configuration was thirty inches (0.762 m). The CRS, where present, was the forward-facing model described in section 4.3.2.7. In each instance it was occupied by a TNO P3 three-year-old ATD. The acceleration pulse applied in each case was the sled acceleration signal from the associated CASA physical test. A typical pulse is described in section 4.2.2.
5.8 Validation results

The results of numerical model validation, describing the level of agreement between model output and the results of corresponding physical experiments, are presented below. Tables 5-1 through 5-8 present the values of the abovementioned five validation metrics for the signals relevant to each corresponding configuration.

A reasonable level of experimental error was estimated as ±20%; parameters where the comprehensive error factor and magnitude error were both less than 20% were deemed to be good approximations of reality and therefore suitable for use in an absolute sense. The remaining parameters were deemed useful in a comparative sense only.

Four graphical examples are presented in Appendix C to assist interpretation of metric values, including head acceleration curves for the adult and infant.

5.8.1 CRS baseline configuration – Lap belt CRS

The CRS was installed using the aircraft seat lap belt. A good level of agreement was observed in infant head acceleration signals. The thorax acceleration signal matched physical test data to a reasonable degree; however, a spike in the model output resulted in the peak acceleration being over-predicted by 181%.

<table>
<thead>
<tr>
<th>Signal</th>
<th>C %</th>
<th>Mpos %</th>
<th>Wpos %</th>
<th>Mneg %</th>
<th>Wneg %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Infant head acceleration</td>
<td>10.8</td>
<td>-6.9</td>
<td>100.0</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>Infant thorax acceleration</td>
<td>28.4</td>
<td>181.1</td>
<td>100.0</td>
<td>N/A</td>
<td>N/A</td>
</tr>
</tbody>
</table>

5.8.2 CRS baseline configuration – ISOFIX CRS

The CRS was installed using the ISOFIX method. The infant head and thorax acceleration signals matched physical test data well.

<table>
<thead>
<tr>
<th>Signal</th>
<th>C %</th>
<th>Mpos %</th>
<th>Wpos %</th>
<th>Mneg %</th>
<th>Wneg %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Infant head acceleration</td>
<td>4.9</td>
<td>-16.7</td>
<td>100.0</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>Infant thorax acceleration</td>
<td>15.7</td>
<td>13.6</td>
<td>100.0</td>
<td>N/A</td>
<td>N/A</td>
</tr>
</tbody>
</table>
### 5.8.3 CRS baseline configuration – LATCH CRS

The CRS was installed using the LATCH method. The infant head and thorax acceleration signal matched well with test data.

#### Table 5-3: CRS baseline configuration – LATCH CRS signal validation results

<table>
<thead>
<tr>
<th>Signal</th>
<th>C %</th>
<th>$M_{pos}$ %</th>
<th>$W_{pos}$ %</th>
<th>$M_{neg}$ %</th>
<th>$W_{neg}$ %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Infant head acceleration</td>
<td>14.2</td>
<td>-7.0</td>
<td>100.0</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>Infant thorax acceleration</td>
<td>23.3</td>
<td>0.2</td>
<td>100.0</td>
<td>N/A</td>
<td>N/A</td>
</tr>
</tbody>
</table>

### 5.8.4 Aft passenger configuration – Lap belt CRS

A Hybrid III 50\textsuperscript{th} percentile male ATD was seated behind an occupied CRS which was installed using the lap belt. The infant head acceleration signal did not match physical test data as closely as in the corresponding CRS baseline configuration. The aft passenger head acceleration signal exhibited a close match with physical test data. Upper neck signals gave mixed results; the fore-aft moment signal matched well with test data while axial force did not.

#### Table 5-4: Aft passenger configuration - Lap belt CRS signal validation results

<table>
<thead>
<tr>
<th>Signal</th>
<th>C %</th>
<th>$M_{pos}$ %</th>
<th>$W_{pos}$ %</th>
<th>$M_{neg}$ %</th>
<th>$W_{neg}$ %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Infant head acceleration</td>
<td>19.2</td>
<td>-1.7</td>
<td>100.0</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>Aft passenger head acceleration</td>
<td>3.3</td>
<td>-4.6</td>
<td>100.0</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>Aft passenger upper neck axial force</td>
<td>43.8</td>
<td>-26.3</td>
<td>53.2</td>
<td>-45.6</td>
<td>46.8</td>
</tr>
<tr>
<td>Aft passenger upper neck fore-aft moment</td>
<td>22.5</td>
<td>100.6</td>
<td>5.8</td>
<td>-9.9</td>
<td>94.2</td>
</tr>
</tbody>
</table>

### 5.8.5 Aft passenger configuration – ISOFIX CRS

A Hybrid III 50\textsuperscript{th} percentile male ATD was seated behind an occupied CRS which was installed using the ISOFIX method. The infant head acceleration signal matched physical test data well. The aft passenger head acceleration signal matched very closely with test data. Upper neck force and moment signals matched reasonably well with test data.

#### Table 5-5: Aft passenger configuration - ISOFIX CRS signal validation results

<table>
<thead>
<tr>
<th>Signal</th>
<th>C %</th>
<th>$M_{pos}$ %</th>
<th>$W_{pos}$ %</th>
<th>$M_{neg}$ %</th>
<th>$W_{neg}$ %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Infant head acceleration</td>
<td>12.9</td>
<td>4.4</td>
<td>100.0</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>Aft passenger head acceleration</td>
<td>5.3</td>
<td>2.4</td>
<td>100.0</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>Aft passenger upper neck axial force</td>
<td>28.1</td>
<td>-23.9</td>
<td>62.3</td>
<td>-0.6</td>
<td>37.7</td>
</tr>
<tr>
<td>Aft passenger upper neck fore-aft moment</td>
<td>15.5</td>
<td>3.7</td>
<td>7.8</td>
<td>-18.4</td>
<td>92.2</td>
</tr>
</tbody>
</table>
5.8.6 Aft passenger configuration – LATCH CRS

A Hybrid III 50th percentile male ATD was seated behind an occupied CRS that was installed using the LATCH method.

The infant head acceleration signal matched physical test data well. The aft passenger head acceleration signal score exhibited a lower level of agreement with test data than in the lap belt and ISOFIX CRS aft passenger configurations. In physical test results, the aft passenger head acceleration signal for this configuration exhibits a sharp peak (indicated by the arrow in Figure 5-2 below) approximately 6 ms after the initial peak resulting from contact with the tray table.

![Figure 5-2: Aft passenger head acceleration signals for LATCH and ISOFIX CRS configurations](image)

A review of the footage of this test suggested that this secondary peak, which is present in the model but to a smaller degree, is likely to be due to the tray table penetrating the seat back and making contact with the CRS. This idea is supported by Figure 5-3 below in which the position of the aft passenger head is overlaid at the time of this secondary peak.
With the exception of the peak extension moment, upper neck force and moment signals generally did not match well with physical test data.

Table 5-6: Aft passenger configuration - LATCH CRS signal validation results

<table>
<thead>
<tr>
<th>Signal</th>
<th>C %</th>
<th>M_{pos} %</th>
<th>W_{pos} %</th>
<th>M_{neg} %</th>
<th>W_{neg} %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Infant head acceleration</td>
<td>12.9</td>
<td>17.1</td>
<td>100.0</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>Aft passenger head acceleration</td>
<td>9.4</td>
<td>4.7</td>
<td>100.0</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>Aft passenger upper neck axial force</td>
<td>56.4</td>
<td>-27.0</td>
<td>50.5</td>
<td>24.2</td>
<td>49.5</td>
</tr>
<tr>
<td>Aft passenger upper neck fore-aft moment</td>
<td>47.7</td>
<td>144.8</td>
<td>25.5</td>
<td>0.4</td>
<td>74.5</td>
</tr>
</tbody>
</table>

5.8.7 Aft passenger baseline configuration

A Hybrid III 50th percentile male ATD was seated behind an empty seat. The aft passenger head acceleration and upper neck force signals matched well with test data. The upper neck moment peak value matched well with test data though the shape was not a good match.
Table 5-7: Aft passenger baseline configuration signal validation results

<table>
<thead>
<tr>
<th>Signal</th>
<th>C %</th>
<th>$M_{pos}$ %</th>
<th>$W_{pos}$ %</th>
<th>$M_{neg}$ %</th>
<th>$W_{neg}$ %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Aft passenger head acceleration</td>
<td>5.7</td>
<td>-18.3</td>
<td>100.0</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>Aft passenger upper neck axial force</td>
<td>11.3</td>
<td>161.1</td>
<td>3.9</td>
<td>-14.3</td>
<td>96.1</td>
</tr>
<tr>
<td>Aft passenger upper neck fore-aft moment</td>
<td>41.0</td>
<td>542.3</td>
<td>5.2</td>
<td>3.8</td>
<td>94.8</td>
</tr>
</tbody>
</table>

5.8.8 Lap-held eighteen-month-old infant

A TNO P1.5 eighteen-month-old ATD was seated on the lap of a Hybrid III 50th percentile male ATD and restrained using a supplementary loop belt. Model prediction of head acceleration was in good agreement with physical test results for the adult and infant. The supplementary loop belt tension signal exhibited a high level of agreement.

Table 5-8: Lap-held eighteen-month-old infant configuration signal validation results

<table>
<thead>
<tr>
<th>Signal</th>
<th>C %</th>
<th>$M_{pos}$ %</th>
<th>$W_{pos}$ %</th>
<th>$M_{neg}$ %</th>
<th>$W_{neg}$ %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Infant head acceleration</td>
<td>7.6</td>
<td>-3.1</td>
<td>100</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>Aft passenger head acceleration</td>
<td>7.5</td>
<td>-17.4</td>
<td>100</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>Supplementary loop belt tension</td>
<td>6.5</td>
<td>5.2</td>
<td>100</td>
<td>N/A</td>
<td>N/A</td>
</tr>
</tbody>
</table>

5.9 Numerical ATD Validation

The numerical ATD models used were developed by the producers of MADYMO, TASS BV. Limited validation data was provided by TASS for the Hybrid III 50th percentile and TNO P3 ATDs; no data is available for the nine- and 18-month-old ATDs. The validated ATD models were subjected to a series of impacts in simple configurations. The validation results provided by TASS BV[100] are subject to a confidentiality agreement and cannot be reproduced here. Instead, the level of agreement between physical and numerical ATD output in selected signals is described in broad terms in Table 5-9 below.
Table 5-9: Validation of numerical ATD models based on confidential data provided by TASS BV.

<table>
<thead>
<tr>
<th>ATD Signal</th>
<th>Level of agreement</th>
</tr>
</thead>
<tbody>
<tr>
<td>TNO P3 head acceleration</td>
<td>High ($M_{pos} ≈ 5%$)</td>
</tr>
<tr>
<td>TNO P3 thorax acceleration</td>
<td>Moderate ($M_{pos} ≈ 10%$)</td>
</tr>
<tr>
<td>Hybrid III 50th percentile head acceleration</td>
<td>High ($M_{pos} ≈ 5%$)</td>
</tr>
<tr>
<td>Hybrid III 50th percentile upper neck axial force</td>
<td>Moderate ($10% \leq M_{pos/neg} \leq 35%$)</td>
</tr>
<tr>
<td>Hybrid III 50th percentile upper neck fore-aft moment</td>
<td>Moderate ($10% \leq M_{pos/neg} \leq 35%$)</td>
</tr>
</tbody>
</table>

The properties of the numerical ATDs (e.g. geometry, joint stiffness) implemented in the model were not modified in any way.

### 5.10 Conclusions

As each seating configuration considered by the CASA study was tested only once there are no data available describing the experimental variability of output parameters. Such variability would result from:

1. Test setup inconsistencies, e.g. variations in belt pretension, ATD initial position, and CRS initial position.
2. Structural failure. Minor structural failure (e.g. cracked lateral tubes) was witnessed in a small number of physical tests. This is attributed to some aircraft seats having been used in multiple tests. The numerical model does not account for structural failure. Tests where structural failure occurred were not excluded from the validation process because the effect would likely have been minor.
3. Measurement errors. In any physical test there is a degree of random error associated with measurement. This is especially true of the types of sensors used in sled testing; signals from accelerometers and load cells are noisy in nature and are prone to ‘drift’.

These sources of error are further compounded by the variability in the level of agreement between physical and numerical ATD results described in section 5.9 above. The level of agreement between model output and the results of physical experiments varied according to the parameter in question. The ATD head acceleration signals for both the infant and aft passenger exhibited the highest level of agreement, with comprehensive error factors and differences in peak magnitudes of less than 20% in all cases.
Levels of agreement observed in the aft passenger upper neck force and moment signals were quite varied, with comprehensive error factors of between 16% and 48%. The level of agreement observed in infant thoracic acceleration signals was also quite varied, with variations as high as 28% and 181% in the comprehensive error factor and magnitude error, respectively.

The supplementary loop belt tension signal, measured in only one case, exhibited a high level of agreement with physical test results. For this parameter the comprehensive error factor and magnitude error were both less than 10%.

The outcome of this validation process was that the occupant head acceleration and supplementary loop belt parameters were deemed suitable for use in quantifying injury and in direct comparison with occupant safety standards. All other parameters, including those that were not considered in the validation process, were used only to compare seating configurations with one another. Thus, the model is henceforth treated as ‘partially validated’.

All parameters considered, along with their method of evaluation, relevant occupants and application are detailed in Table 5-10 below. ‘Application’ refers to whether that injury parameter was evaluated by means of an occupant protection standard or known injury response; where the level of agreement observed in the validation phase did not permit such a use (see section 5.8), the parameter was used in comparative analysis only.

<table>
<thead>
<tr>
<th>Part</th>
<th>Injury parameter</th>
<th>Metric</th>
<th>Occupant</th>
<th>Application</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head</td>
<td>Acceleration</td>
<td>HIC</td>
<td>Infant, adult</td>
<td>Adult injury prediction: Mertz et al.[79] Standard: FMVSS 208</td>
</tr>
<tr>
<td></td>
<td>Contact force</td>
<td>3 ms</td>
<td>Lap-held infant</td>
<td>Comparison only</td>
</tr>
<tr>
<td>Neck</td>
<td>Fore-aft bending</td>
<td>Nij</td>
<td>Infant, adult</td>
<td>Comparison only</td>
</tr>
<tr>
<td></td>
<td>Axial force</td>
<td>Nij</td>
<td>Infant, adult</td>
<td>Comparison only</td>
</tr>
<tr>
<td>Thorax</td>
<td>Acceleration</td>
<td>3 ms</td>
<td>Lap-held infant</td>
<td>Comparison only</td>
</tr>
<tr>
<td></td>
<td>Contact force</td>
<td>3 ms</td>
<td>Lap-held infant</td>
<td>Comparison only</td>
</tr>
<tr>
<td>Abdomen</td>
<td>Contact force</td>
<td>3 ms</td>
<td>Lap-held infant</td>
<td>Comparison only</td>
</tr>
<tr>
<td></td>
<td>Belt tension</td>
<td>3 ms</td>
<td>Lap-held infant</td>
<td>Injury prediction: Kent et al.[66]</td>
</tr>
</tbody>
</table>

Table 5-10: Occupant injury parameters considered
6 Safety of the lap-held infant: restraint condition

6.1 Introduction
According to different international aviation safety regulations, lap-held infants may either be carried unrestrained or restrained by a supplementary loop belt. No prior study has compared the safety of the lap-held infant in these opposing restraint conditions; however, individual studies have made the following conclusions:

- Lap-held infants are vulnerable to potentially dangerous contact with the forward seat and the adult passenger on whose lap they are seated regardless of restraint condition[2-5, 7].
- An unrestrained infant is at risk of being projected into the aircraft cabin if the adult is unable to restrain them[5], an outcome that has been realised in real-world air accidents[20, 22, 23].
- An infant restrained by the supplementary loop belt is at increased risk of abdominal injury[7], a conclusion arrived at by visual observation rather than by measurement of injury parameters.

6.2 Objectives
The work described in this chapter sought to quantify and compare the safety of the unrestrained and restrained lap-held infant with a view to determining whether either restraint condition represents a best case. An effort was made to quantify the effect of the supplementary loop belt on infant abdominal injury potential.

6.3 Method
The partially validated numerical model described in chapter 4 was used to conduct four experiments comparing the safety of nine- and eighteen-month-old infants in two restraint conditions: unrestrained and restrained by a supplementary loop belt. The following injury parameters were measured; the method of evaluation is given in parentheses:

- Infant head acceleration ($\text{HIC}_{15}$)
- Infant head contact force (3 ms clip)
- Infant neck fore-aft moment ($N_{6}$)
• Infant neck axial force (N<sub>j</sub>)
• Infant thoracic acceleration (3 ms clip)
• Infant thoracic contact force (3 ms clip)
• Infant abdominal contact force (3 ms clip)
• Infant thorax velocity
• Supplementary loop belt tension (3 ms clip, restrained condition only)

Output parameter values from the different experiments were compared with one another to determine the relative safety of the restrained and unrestrained conditions for the lap-held infant. Infant head injury parameter values were also compared with occupant protection standards. The risk of infant abdominal injury due to the supplementary loop belt was evaluated using the injury risk curve of Kent et al.[66] (see section 3.4.4). The same approach was not used to predict infant neck and thoracic injury potential as the associated parameters did not exhibit a good level of agreement with physical test results during model validation (see chapter 5).

The aircraft seat model does not include any representation of the cabin ceiling or other structure; therefore, in the case of the unrestrained lap-held infant, the model is only able to measure the loads acting on the infant while it is within the space bounded by the adult ATD and the forward seat. The potential effects of any secondary impacts were not able to be considered; instead, the velocity of the infant’s thorax as it left this space was evaluated as a proxy for injury risk. The analysis was run for 200 ms in order to measure the velocity of the unrestrained infant after the initial impact.

It is possible that this analysis under-predicted the loading experienced by the lap-held infant as a result of the non-biofidelic stiffness of the Hybrid III thoracic spine (see section 3.2.1).

6.3.1 Seating configuration

The aircraft seat model described in chapter 4 was employed with a seat pitch of 30 inches. The aft row was occupied by ATDs while the forward row was left empty. Four experiments were carried out in total; in each instance, the adult passenger was represented by a 50<sup>th</sup> percentile male Hybrid III ATD. As there is no available data describing the size and mass of adults who typically carry children on the lap in aircraft, this ATD was chosen because it most closely represents a mid-sized adult. Nine- and eighteen-month-old lap-held infants were represented by TNO P series ATDs. An example configuration is presented in Figure 6-1 below.
6.3.2 ATD and restraint positioning
Belts and ATDs were positioned with the aid of footage from CASA physical experiments[2]. In each case the adult ATD was restrained by a lap belt and seated in an upright position with the hands in front of the infant’s abdomen. The infant ATD was seated in an upright position on the adult’s lap.

6.3.3 Interaction between ATDs
Contacts were defined between each part of the infant ATD and each part of the adult ATD. Contacts between surfaces of the same ATD were also defined where relevant, such as between the infant’s legs and upper body.

The effect of the adult’s grip on the infant was considered to be negligible based on physical experiments by Hardy[5] involving an unrestrained lap-held infant. Hardy[5] cites a study by Mohan and Schneider[101] which evaluates the clasping strength of adults attempting to restrain lap-held infants in frontal crashes.
6.4 Results and discussion

6.4.1 Occupant motion

Infant restraint condition had only a minor effect on infant kinematics prior to impact with the forward seat. In the unrestrained condition the infant moved entirely forward of the adult’s body before being projected upwards by the rebound of the forward seat-back. The infant’s head, torso and legs impacted the forward seat. In the restrained condition the infant remained in the space bounded by the forward seat and the adult’s legs and upper body. The infant’s head and legs impacted the forward seat.

Figure 6-2 and Figure 6-3 below illustrate the relative positions of the infant and adult occupants at the time of the first significant contact between the adult’s head and the infant’s body. In the case of the unrestrained nine-month-old, the adult’s head contacted the infant’s head and thorax, while in the unrestrained eighteen-month-old the adult’s head contacted only the infant’s thorax.

Figure 6-2: Nine-month-old infant lap-held by 50th percentile adult male at time 100 ms. Restrained by supplementary loop belt (left) and unrestrained (right).
At an elapsed time of 200 ms, the forward and aft seat-backs were still undergoing rebound motion. At this time the restrained infants remained contained in the space between the forward seat and the adult’s legs and upper body, while the nine- and eighteen-month-old unrestrained infants were translating freely upwards at $3 \text{ m/s}^{-1}$ and $5 \text{ m/s}^{-1}$ respectively as measured at the thorax. These velocities are equivalent to those reached during falls of approximately 0.5 m and 1.25 m, respectively.

### 6.4.2 Head injury

In each configuration the infant’s head contacted the forward seat; the different impact locations (see Figure 6-2 and Figure 6-3 above) resulted in minor variation in HIC$_{15}$ score. In each case infant HIC$_{15}$ scores, presented below in Figure 6-4, far exceeded the FMVSS 208 HIC$_{15}$ limits for both a twelve-month-old (390) and an adult (700). Infant HIC$_{15}$ scores for the nine-month-old indicated a lower degree of head injury in the unrestrained condition than the restrained condition. This result was reversed in the case of the eighteen-month-old, though the difference in HIC$_{15}$ scores was smaller.
In every case except the unrestrained eighteen-month-old, the adult’s head contacted the infant’s head while the latter was in contact with the forward seat. This resulted in two different loading scenarios for the infant’s head:

- where the adult’s head did not make contact with the infant’s, the load acting on the infant’s head was due to its own inertia, and
- in cases where the adult’s head made contact with the infant’s, the load acting on the infant’s head was due to the inertia of both the infant’s head the adult’s.

This distinction is important as it means that HIC is unsuitable as a tool for assessing head injury potential in the lap-held infant. As HIC is calculated entirely from the net head acceleration of the relevant occupant it is unable to account for effects of applied forces which do not result in acceleration of the head. Such a case exists here, where the infant’s head is loaded in compression between the forward seat and the adult’s head; for this reason, head contact force was chosen as an injury parameter.

Peak head contact force, presented below in Figure 6-5, was substantially higher for the nine-month-old than for the eighteen-month-old. In the case of the nine-month-old, the restrained condition resulted in a 4% lower peak force than the unrestrained condition. The opposite was found to be true in the case of the eighteen-month-old and to a much greater degree: the restrained condition resulted in a peak force nearly twice that measured in the unrestrained condition.
Figure 6-5: Head contact force 3 ms clip values for restrained and unrestrained nine- and eighteen-month-old lap-held infants.

The head contact force observed in the eighteen-month-old was significantly lower than in the nine-month-old as a result of its greater stature. For the eighteen-month-old the smaller initial distance between the two occupants’ heads resulted in a correspondingly lower relative velocity at impact than for the nine-month-old. In the case of the unrestrained eighteen-month-old the head contact force was further reduced by the infant head having translated far enough forward that the adult’s head did not make contact.

### 6.4.3 Neck injury

Occupant protection standards do not specify critical neck injury parameter values for nine- or eighteen-month-old infants. Instead, the values specified in FMVSS 208 for a twelve-month-old were used to calculate $N_{ij}$ for use in a comparative sense. It is therefore likely that the results presented here under-estimate $N_{ij}$ for the nine-month-old and over-estimate $N_{ij}$ for the eighteen-month-old. For this reason, $N_{ij}$ here is denoted $N_{ij}^*$. Neck injury parameter values are interpreted here in a comparative sense only in accordance with the level of agreement with physical test data evident in the validation process.
Neck injury results are presented in Figure 6-6 above. In each instance, the maximum $N_{ij}^*$ value occurred while the neck was loaded in extension (head tilted back) and the head was in contact with the forward seat.

For both infant sizes, the restrained condition produced a much higher $N_{ij}^*$ value than the unrestrained condition. The reason for this was that in the restrained condition the infant’s head and upper body rotated forwards in an arc about the supplementary loop belt. Once the infant’s head had contacted the forward seat its upper body continued to rotate, causing the head to tilt backwards. A similar effect was observed in the unrestrained infants, though to a much lesser degree, as a result of contact with the head and upper body of the adult as it arced about its own lap belt. The bending stiffness of the nine-month-old’s neck is approximately 1.7 times that of the 18-month-old.

### 6.4.4 Thoracic injury

Results indicated a similar level of thoracic acceleration for all cases. The lowest maximum value was observed in the restrained eighteen-month-old and the highest in the unrestrained nine-month-old. Peak thoracic acceleration values, presented in Figure 6-7 below, were a result of contact with the adult ATD.

The restrained condition was associated with a reduction in maximum thoracic acceleration for both the nine-month-old (8%) and eighteen-month-old (13%).
Thoracic contact force was significantly higher in the unrestrained condition than in the restrained condition, by a factor of 2.3 for the nine-month-old and a factor of 2.1 for the eighteen-month-old (see Figure 6-8 below). The increased force in the unrestrained condition was a result of the infant’s body translating forward sufficiently to place the thorax in the path of the adult’s head.
6.4.5 Abdominal injury

Supplementary loop belt use resulted in significantly greater abdominal contact force than the unrestrained condition (see Figure 6-9). For the nine-month-old the restrained condition resulted in an abdominal force 2.5 times that of the unrestrained condition; in the eighteen-month-old the abdominal force associated with the restrained condition was 6.6 times that in the unrestrained condition.

Figure 6-9: Abdominal contact force 3 ms clip values for restrained and unrestrained nine- and eighteen-month-old lap-held infants.

In the restrained condition the abdominal contact force was due to the supplementary loop belt while in the unrestrained condition the force was a result of contact with the adult or the infant's own legs. Tension in the supplementary loop belt (see Figure 6-10) was similar in both the nine- and eighteen-month-old cases. The maximum tension values measured were 3250 N and 3000 N, respectively.
The size and positioning of the supplementary loop belt matched the experimental method employed by Kent et al. [66] in the development of abdominal injury risk curves for the six-year-old child (see section 3.4.4). The maximum belt tension values observed in the ATDs representing the nine- and eighteen-month old are associated with an approximately 95% probability of serious (AIS 3) or greater abdominal injury in a six-year-old child (95% confidence interval 80% – 100%).

**6.5 Conclusion**

The work detailed in this chapter is the first in the open literature to make a quantitative comparison and evaluation of the safety of the two mandated restraint conditions for the lap-held infant: unrestrained and restrained by a supplementary loop belt. A numerical model was used to assess the safety of infants aged nine months and 18 months held on the lap of a mid-sized adult male. The safety parameters considered were adult and infant ATD kinematics, the contact loading acting on the infant ATD and, where present, the peak tension in the supplementary loop belt. There is insufficient data on real-world emergency landing events to evaluate the results of this study in terms of the historical occurrence of such injuries (see section 2.5.3); however, results were evaluated in the context of the conclusions relevant prior studies.

Neither of the restraint conditions for the lap-held infant mandated by aviation safety authorities presented a clear best-case in terms of infant safety. This analysis confirmed that
an unrestrained lap-held infant is at risk of being projected through the aircraft cabin during the impact sequence, an outcome previously observed in experimental investigations[5] and air accidents[20, 22]. At the end of the impact sequence, the ATDs representing the nine- and eighteen-month-old infants were projected towards the ceiling of the aircraft cabin at speeds of approximately 3 m·s\(^{-1}\) and 5 m·s\(^{-1}\) respectively. These speeds are equivalent to those reached during falls of approximately 0.5 m and 1.25 m, respectively.

It has been reported, based on a qualitative evaluation, that the supplementary loop belt may cause “extremely serious” internal abdominal injuries in infants[7]. In the present study an abdominal injury risk curve developed for the six-year-old child by Kent et al.[66] was used to evaluate injury to the restrained infant caused by the supplementary loop belt. Under the presumption that the susceptibility of the nine- or eighteen-month-old to blunt abdominal trauma is equal to or greater than that of the six-year-old child, the loads measured by the present study resulted in a greater than 80% probability of abdominal injuries at the abbreviated injury scale classification of ‘severe’ or greater. Such injuries include perforation and laceration of the small bowel as well as large bowel injuries resulting in gross faecal contamination of and significant blood loss into the peritoneal cavity[66].

The restrained and unrestrained infant ATDs, particularly the head, was loaded in compression between the forward seat and the adult’s body. This effect has been observed in prior studies[2, 3, 7]. One implication of this is that the head injury criterion, a widely used head injury metric, is insufficient in quantifying head injury potential. The contact loading applied by the accompanying adult did not result in acceleration of the infant head and was therefore not accounted for by HIC; this resulted in an under-estimation of infant head injury potential by this method. Nonetheless, HIC\(_{15}\) was evaluated for the lap-held infant in the restrained and unrestrained conditions. Its value greatly exceeded the limit for both infants and adults set out in automotive occupant protection standards[56]. The nine-month-old in the unrestrained condition recorded the minimum HIC\(_{15}\) score, while the nine-month-old in the restrained condition recorded the maximum HIC\(_{15}\) score.

No prior study has evaluated head contact load for the lap-held infant. In this study, the peak contact load experienced by the infant’s head was between 1.5 kN and 4.5 kN across all experiments. Interaction between two variables - the restraint condition and the infant’s stature - appeared to influence the magnitude of the peak head contact load. The smaller body of the nine-month-old was contained within the path of the adult’s body as it flailed forward, resulting in head-to-head contact regardless of restraint condition. A lower load was observed
in the eighteen-month-old as a result of its taller stature: in the restrained condition the peak load was lesser than for the nine-month-old due to the lower relative velocity between infant and adult head at impact. In the unrestrained condition the infant translated far enough forward that no head-to-head contact occurred.

Restraint condition and infant stature were found to have effects on the peak thoracic contact force opposite to those observed in the head contact force results. Restraint condition was a significant factor: the unrestrained condition resulted in a doubling of peak thoracic contact force for both infant sizes. The taller stature of the eighteen-month-old placed its thorax more within the path of the adult’s head while the unrestrained condition allowed the infant to translate far enough forward that its thorax was in the path of the adult’s head regardless of stature. This resulted in the unrestrained eighteen-month old, having received the lowest head contact force, sustaining the highest observed contact force to the thorax. The restrained condition was associated with reduced thoracic acceleration.

The restrained condition resulted in significantly greater potential for neck injury than the unrestrained condition. The mechanism for this was a difference in the nature of the contact with the forward seat as the infant flailed about its lap belt.

It is possible that in both restraint conditions the loading experienced by the infant was above the survivable limit. This is especially true in the case of head contact force where large forces were applied to the soft and pliable infant skull. The following chapter presents a characterisation and further analysis of these forces.
7 Safety of the lap-held infant: head loading characteristics

7.1 Introduction

Head contact force is an injury mechanism strongly associated with skull fracture[102]. Physical experiments[2, 7] and the analysis detailed in the previous chapter have shown that the lap-held infant's body, particularly the head, is loaded in compression between the forward seat-back and the adult's body during deceleration (see Figure 7-1 below). The loading experienced by the lap-held infant under emergency landing conditions has been described as “extremely serious to fatal”[7] and likely to cause “excessive head and neck trauma”[2]. These qualitative findings are based on observation, HIC score or a combination of both.

![Figure 7-1: The restrained lap-held nine-month-old pre-impact (left) and during contact with the forward seat-back (right)[2].](image)

The nature and magnitude of potential head injury to the lap-held infant have not previously been determined for two principal reasons:

1. The contact force acting on the head has not previously been characterised as it cannot be measured by physical testing. The analysis detailed in the previous chapter confirmed that the head acceleration parameter in isolation cannot account for the compressive loading to the infant head.

2. There is presently no simple injury criterion able to predict head injury under the compressive loading condition experienced by the head of the lap-held infant. Instead, it is likely that an accurate prediction of head injury risk will only become possible through finite element modelling of the infant head.
A significant amount of work has been done towards a finite element model for the study of paediatric head impact[72, 103-108]. The results of the work detailed in this chapter will support the use of such a model in predicting head injury mechanisms and severity for the lap-held infant.

As a first step towards quantifying head injury potential for the lap-held infant the contact force applied to the infant forehead during impact was isolated and used to estimate the equivalent acceleration that would have occurred were the head not loaded in compression. This allowed the calculation of a more instructive HIC value for the infant than was arrived at in the previous chapter.

7.2 Objectives
The work detailed in this chapter sought to characterise the contact forces acting on the head of the lap-held infant to inform a potential future study aimed at predicting injury mechanisms and severity. A further objective was to use this information to refine the assessment of infant head injury potential made in the preceding chapter.

7.3 Method
The results of the work detailed in the previous chapter were analysed to determine the force-time characteristic and therefore impulse delivered to the nine-month-old infant’s head by the forward seat and adult head in both the restrained and unrestrained conditions. The nine-month-old infant was chosen because the work detailed in the previous chapter identified it as the worst case of the two available infant ages.

The contact force applied to the infant forehead during the impact sequence was isolated. Through application of Newton’s second law the contact force was translated into an estimate of the head acceleration that would result from such a force if the head were not loaded in compression; i.e. the force-time curve was divided by the ATD head mass of 2.0 kg to arrive at an acceleration-time curve. This data was then used to calculate a force-based HIC_{15} for the infant, denoted HIC_{f}.
7.4 Results and discussion

The force-time characteristics of the contact loading applied to the infant head by the forward seat and adult head are presented in Figures 7-2 and 7-3 below. The forward seat applied force to the face and frontal bone, while the adult head applied force to the parieto-occipital region.

![Figure 7-2: Force-time characteristic of contact to the restrained nine-month-old infant head by the forward seat and adult head.](image)

Figure 7-2 above illustrates the approximately equal application of force by the forward seat and adult head in the restrained condition. The infant’s head was loaded in compression between the tray table and adult head for approximately thirty milliseconds. In Figure 7-3 below it is evident that the unrestrained lap-held infant’s head sustained contact loading over a much greater period of time. It was loaded in compression for approximately twenty milliseconds; however, a notable difference is that the infant’s head was in contact with the forward seat-back as the seat-back rebounded. This rebound effect was responsible for the prolonged duration of contact.
Figure 7-3: Force-time characteristic of contact to the unrestrained nine-month-old infant head by the forward seat and adult head.

The impulse values derived from the force-time characteristics in Figure 7-2 and Figure 7-3 are presented below in Table 7-1.

**Table 7-1: Impulse delivered to nine-month-old infant head through contact with forward seat and adult head**

<table>
<thead>
<tr>
<th>Impulse (N·s)</th>
<th>Fwd seat</th>
<th>Adult head</th>
</tr>
</thead>
<tbody>
<tr>
<td>Restrained</td>
<td>83</td>
<td>71</td>
</tr>
<tr>
<td>Unrestrained</td>
<td>231</td>
<td>29</td>
</tr>
</tbody>
</table>

The force-time characteristics of the load applied to the infant forehead (labelled ‘Forward seat’ in Figure 7-2 and Figure 7-3) were translated into estimated equivalent acceleration curves by dividing by the ATD head mass of 2.0 kg. The HIC\(_F\) scores derived from these curves were 4993 in the restrained condition and 7997 in the unrestrained condition.

7.5 Conclusion

Through qualitative analysis, prior studies have described the loading experienced by the lap-held infant under emergency landing conditions as “extremely serious to fatal”[7] and likely to
cause “excessive head and neck trauma”[2]. It is likely that an accurate prediction of injury outcomes will only be possible through the application of a finite element model of the infant head; such a technique remains at the forefront of research in the field of paediatric injury biomechanics. To facilitate the use of such a model in the future, the impact loading applied to the nine-month-old lap-held infant’s head in both the restrained and unrestrained conditions has been characterised by the present study.

Building on the initial assessment of head injury potential for the lap-held infant described in the preceding chapter, a refined prediction was made of the head injury criterion values for the nine-month-old in this seating configuration. The HIC scores derived in this analysis were 4993 in the restrained condition and 7997 in the unrestrained condition. These values are an order of magnitude greater than the limits set out in motor vehicle occupant safety standards (see section 3.4.1).
8 Safety of the lap-held infant: emergency brace positions

8.1 Introduction
Prior studies have considered emergency brace positions for adult passengers through physical experiments[25] and the analysis of a real-world air accident has proven these to be effective in reducing injury to adult passengers[14]. Brace positions involving lap-held infants have been identified[25] and are recommended for use by some aviation safety authorities[26, 28]; however, no study has yet quantified or compared their efficacy.

8.1.1 Objective
The work described in this chapter sought to determine the degree to which emergency brace positions provide protection to the lap-held infant. Also considered was whether the risk of head injury to the infant may be mitigated by a lateral offset between the occupants’ heads.

8.1.2 Traditional brace positions
Two distinct emergency brace positions were identified for analysis; one suitable for both restrained and unrestrained infants and one suitable for unrestrained infants only. The position recommended by Transport Canada[28] (see section 2.3.3) places the infant in a rearward-facing orientation. Its implementation in the numerical model is depicted in Figure 8-1 below.
Figure 8-1: Rearward-facing unrestrained nine-month-old ATD on the lap of a braced 50th percentile male ATD.

The rearward-facing orientation is suitable in cases where the infant is unrestrained; however, it would be unsuitable for restrained infants as the inertial loading of the infant’s body during impact would result in the undesirable condition of the infant’s spine being loaded in extension.

A brace position suitable for the restrained condition was chosen from the passenger safety card of a major Australian air carrier. In this forward-facing position, depicted in Figure 8-2 below, the adult leans forward over the infant as recommended by the FAA[109]; however, no support is provided to the infant’s body.
8.1.3 Modified brace position

The results of the previous two chapters highlighted the risk of head-to-head contact between the lap-held infant and the accompanying adult. As the descriptions of the traditional brace positions identified above do not make mention of the alignment of the two occupants’ heads, the same outcome is possible in the braced condition. To assess the potential benefit of a deliberate misalignment of the occupants’ heads a modified brace position is proposed in which the occupants’ heads are laterally offset from one another and the infant is forward-facing to ensure applicability to either restraint condition (see Figure 8-3 below).
8.2 Method

Six initial configurations, described below in Table 8-1, were tested to assess the potential safety benefit of the two traditional brace positions for nine- and eighteen-month-old infants. For the forward-facing position, both restrained and unrestrained conditions were tested; for the rearward-facing position, only the unrestrained condition was tested (see section 8.1.2).

<table>
<thead>
<tr>
<th>Infant age</th>
<th>Restraint condition</th>
<th>Infant orientation</th>
</tr>
</thead>
<tbody>
<tr>
<td>9 months</td>
<td>Supplementary loop belt</td>
<td>Forward (FF)</td>
</tr>
<tr>
<td>9 months</td>
<td>Unrestrained</td>
<td>Forward (FF)</td>
</tr>
<tr>
<td>9 months</td>
<td>Unrestrained</td>
<td>Rearward (RF)</td>
</tr>
<tr>
<td>18 months</td>
<td>Supplementary loop belt</td>
<td>Forward (FF)</td>
</tr>
<tr>
<td>18 months</td>
<td>Unrestrained</td>
<td>Forward (FF)</td>
</tr>
<tr>
<td>18 months</td>
<td>Unrestrained</td>
<td>Rearward (RF)</td>
</tr>
</tbody>
</table>

A seventh experiment was conducted to test the potential benefit to infant head injury of a modified forward-facing brace position in which the occupants’ heads are laterally offset. This analysis considered the nine-month-old in the forward-facing restrained and unrestrained conditions. The peak (3 ms clip) head contact force was measured to enable comparison with the results of the six initial experiments.
All experiments were carried out using the numerical model described in chapter 4 with a seat pitch of 0.762 m. In each experiment, ATDs occupied the aft row and the forward row was unoccupied. The adult passenger was represented by a 50th percentile male Hybrid III ATD. Contacts were defined as in the previous lap-held analysis and the effect of the adult’s grip on the infant was considered to be negligible (see section 6.3.3).

Results for the traditional braced configurations described here were compared with those of the unbraced lap-held infant analysis described in chapter 6. Infant head contact force was the only injury parameter evaluated as the work of the preceding two chapters identified that an acceleration-based evaluation of HIC is unsuitable.

The forehead contact force measured in the restrained and unrestrained nine-month-old in the forward-facing traditional and modified positions was used to evaluate $HIC_F$ by the method described in the previous chapter. This enabled the comparison of results with those of the previous chapter and hence an evaluation of the efficacy of the traditional and modified brace positions.

As in the lap-held analyses described in the preceding chapters, it is possible that this analysis under-predicted the loading experienced by the lap-held infant due to the non-biofidelic stiffness of the Hybrid III thoracic spine[42].

### 8.3 Results and discussion

Sections 8.3.1 to 8.3.5 compare the results of the experiments employing the traditional brace positions with those of the unbraced lap-held analysis described in Chapter 6. The effects of the modified brace position on infant head injury potential are evaluated in section 8.3.6.

#### 8.3.1 Occupant motion

The infant’s head and upper body contacted the forward seat-back in each of the six experiments. The closer initial proximity to the forward seat-back meant that this contact occurred at a lower relative velocity than in the unbraced cases. In each experiment, the infant’s head was compressed between the adult’s head and the forward tray table. In the restrained condition the infant’s body remained within the space bordered by the adult’s upper body and upper legs for the duration of the experiment. The unrestrained infants moved forward onto the forward seat-back, though not to the degree observed in the unbraced case (see section 6.4). The unrestrained condition did not result in any significant vertical velocity component at the end of the 200 ms experiment as was the case in the
unbraced analysis (see section 6.4.1). The legs of the forward-facing infants contacted the forward seat-back early in the experiment; the legs of infants in the rearward-facing configuration did not contact the forward seat-back.

### 8.3.2 Head injury

In each of the experiments involving the nine-month-old the traditional brace positions resulted in significantly reduced maximum head contact force. The mechanism for this was the smaller initial distance between the infant’s head and the forward seat. This smaller distance resulted in a lower relative velocity between the infant’s head and the forward seat-back at the time of impact and correspondingly lower contact force. The rearward-facing brace position resulted in better mitigation of head contact force than the forward-facing brace position. This effect was most significant in the case of the nine-month-old where the peak force was 27% less than in the unrestrained forward-facing position.

![Figure 8-4: Comparison of maximum head contact force results for braced and unbraced nine-month-old lap-held infant. Infant forward-facing (FF) and rearward-facing (RF).](image)

The case of the restrained eighteen-month-old followed this pattern also; however, the head contact forces measured in the unrestrained forward- and rearward-facing braced cases for this infant were greater than for the unrestrained and unbraced case. The reason for this was that the unrestrained and unbraced eighteen-month-old translated far enough forward that its head was not within the trajectory of the adult’s head, resulting in a particularly low head contact force in the infant.
To assess the effect of the traditional forward-facing brace position on infant head injury potential, HIC$_F$ scores were calculated for the nine-month-old using the method described in the preceding chapter.

In the nine-month-old the adoption of the forward-facing brace position resulted in a significant reduction in HIC$_F$ score regardless of restraint condition.
8.3.3 Neck injury

As in the unbraced analysis, infant neck injury was evaluated using the critical values specified in FMVSS 208 for a twelve-month-old. It is likely that this method resulted in an under-prediction of $N_{ij}$ for the nine-month-old and an over-prediction for the eighteen-month-old. For this reason, $N_{ij}$ here is denoted $N_{ij}^*$; results are presented in Figure 8-7 and Figure 8-8 below. Neck injury parameter values are interpreted here in a comparative sense only in accordance with the level of agreement with physical test data evident in the validation process.

![Figure 8-7: Comparison of $N_{ij}^*$ values for the nine-month-old lap-held infant in the braced unbraced conditions, with infant forward-facing (FF) and rearward-facing (RF).](image)

The $N_{ij}^*$ results revealed that the forward-oriented braced condition reduces infant neck injury potential. The improvement ranged from a small 5% reduction in the unrestrained eighteen-month-old to a significant 80% reduction in the restrained infant of the same size. The rearward-facing brace position was detrimental to infant neck injury potential; $N_{ij}^*$ results were significantly increased compared with the forward-facing brace position.
8.3.4 Thoracic injury

Thoracic acceleration results are presented in Figure 8-9 and Figure 8-10 below. In both the nine- and eighteen-month-old infants the braced condition led to lower maximum thoracic acceleration values than the unbraced condition. The mechanism for this reduction was a lower relative velocity between the infant’s thorax and the forward seat-back at the time of impact.

Figure 8-9: Comparison of maximum thoracic acceleration results for braced and unbraced nine-month-old lap-held infant, with infant forward-facing (FF) and rearward-facing (RF).
The forward-facing brace position resulted in a small benefit for the unrestrained eighteen-month-old as a result of the adult’s head making a secondary impact with the infant’s thorax. In the case of the unrestrained rearward-facing eighteen-month-old the lower position of the infant’s thorax prevented this secondary impact from occurring.

Figure 8-10: Comparison of maximum thoracic acceleration results for braced and unbraced eighteen-month-old lap-held infant, with infant forward-facing (FF) and rearward-facing (RF).

Peak thoracic contact force results are presented in Figure 8-11 and Figure 8-12 below. The traditional brace positions resulted in significantly lower peak thoracic contact forces than the corresponding unbraced configurations except for the unrestrained forward-facing eighteen-month-old, a result of the adult’s head making a secondary impact on the infant’s thorax as discussed above. In each of the other cases, the reduced thoracic contact force was a result of smaller relative velocity between the adult and infant occupants and the forward seat-back at the time of impact.
Figure 8-11: Comparison of maximum thoracic contact force results for braced and unbraced nine-month-old lap-held infant, with infant forward-facing (FF) and rearward-facing (RF).

In the nine-month-old the rearward-facing brace position was associated with significantly higher peaks in thoracic acceleration and contact force than the forward-facing brace position. In this infant the forward-facing position resulted in head-first contact with the tray table which mitigated subsequent thoracic contact with the same surface; in the rearward-facing position, head and thoracic contact with the tray table were concurrent. An inverse effect was observed in the eighteen-month-old but was the result of a different mechanism: in the rearward-facing position, thoracic contact was with the soft fabric below the tray table; in the forward-facing position, thoracic contact was with the tray table itself.
8.3.5 Abdominal injury

Figure 8-13 and Figure 8-14 below present a comparison of peak abdominal contact force in the braced and unbraced configurations. Peak values were lower in the traditional brace positions than in the corresponding unbraced condition. In the unrestrained condition the peak force was a result of contact with the abdomen or thorax of the adult ATD, while in the restrained condition the peak was due to the restraint force applied by the supplementary loop belt. The braced condition most greatly benefitted the unrestrained forward-facing nine-month-old and the restrained forward-facing eighteen-month-old.
Figure 8-13: Comparison of maximum abdominal contact force results for braced and unbraced nine-month-old lap-held infant, with infant forward-facing (FF) and rearward-facing (RF).

Figure 8-14: Comparison of maximum abdominal contact force results for braced and unbraced eighteen-month-old lap-held infant, with infant forward-facing (FF) and rearward-facing (RF).

Peak supplementary loop belt tension results observed in the restrained configuration are presented below in Figure 8-15). The braced condition resulted in a significant reduction of supplementary loop belt tension in both infant sizes. The infant’s closer proximity to the forward seat-back in the braced condition led to the seat-back itself providing ‘restraint’ sooner in the impact sequence and therefore lower maximum tension in the supplementary loop belt.

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As in the unbraced analysis, an abdominal injury risk curve developed for a six-year-old child [66] was used to predict the probability of abdominal injury of AIS classification ‘serious’ or greater. For the nine-month old the braced condition was associated with a probability of 35% (95% confidence interval 10 – 60%), while for the eighteen-month-old the braced condition was associated with a probability of 60% (95% confidence interval 35 – 80%).

### 8.3.6 Modified brace position

The modified brace position was more effective in reducing the peak head contact force than the traditional brace position (see Figure 8-16 below). The improvement was most pronounced in the unrestrained infant with a reduction in peak force of 18%. In the restrained case the modified position resulted in a smaller improvement of 9%.
Figure 8-16: Comparison of peak head contact force for the nine-month-old lap-held-infant in forward-facing traditional and modified brace positions.

A reduction in impact severity was also observed in forehead contact force results, manifesting as the reduced HIC<sub>F</sub> values presented below in Figure 8-17.

Figure 8-17: Comparison of HIC<sub>F</sub> values for nine-month-old lap-held infant in forward-facing traditional and modified brace positions.

8.4 Conclusion

Though brace positions involving lap-held infants have previously been identified[25] and are recommended for use by some aviation safety authorities[26, 28], no prior study has quantified or compared their efficacy. The present study analysed the safety of lap-held nine-
and eighteen-month-old infants in situations where the accompanying passenger, a 50th percentile adult male, had adopted an emergency brace position. The results were compared with those of the unbraced analysis described in Chapters 6 and 7 to determine the efficacy of brace positions in reducing loading on the infant.

The analysis primarily considered two traditional brace positions, one with the infant forward-facing and the other with the infant rearward-facing. Because the recommended brace positions made no mention of head position, the occupants’ heads were aligned laterally to represent a worst-case scenario for the infant. A modified brace position in which the infant was forward-facing and the heads of the infant and accompanying adult were laterally offset was evaluated for the nine-month-old in terms of head injury parameters. The restrained and unrestrained conditions were considered in the forward-facing positions; in the rearward-facing position the restrained condition was not considered as the restraint force would result in the undesirable condition of the infant’s spine being loaded in extension.

In general, the braced condition resulted in reduced loading on the infant compared to the unbraced condition in a corresponding restraint condition. With the exception of the unrestrained eighteen-month-old, the traditional brace positions resulted in lower inertial and contact loading to the head, thorax and abdomen regardless of infant age or restraint condition. This reduced loading was due to the closer initial proximity of the adult and infant bodies and the corresponding lower relative velocity at the time of impact.

The traditional brace positions resulted in a substantial reduction of peak head contact force in the nine-month-old, with values approximately halved in both restraint conditions. A 21% reduction in peak force was observed in the restrained eighteen-month-old; however, a significant increase occurred in the unrestrained condition for this infant. This was due to its body being contained more within the trajectory of the adult’s head and upper body in the braced condition than in the unbraced condition.

The traditional forward-facing brace position was associated with benefits to infant neck injury potential ranging from minor (5% reduction) to significant (80% reduction) across all configurations. Compared with the forward-facing brace position, the rearward-facing position was found to significantly promote infant neck injury potential.

The traditional brace positions resulted in reduced thoracic acceleration in all cases. Thoracic contact force was reduced in all cases except for the forward-facing unrestrained eighteen-month-old; in the nine-month-old cases a substantial reduction of greater than 50% was
observed. The mechanism responsible for the increased force observed in the unrestrained eighteen-month-old was the same as for the increased head contact force in this configuration. The rearward-facing brace position better protected the eighteen-month-old from thoracic injury than the forward-facing position; however, the opposite was true in the nine-month-old.

Abdominal loading was effectively reduced in the traditional brace positions. Contact loading on the abdomen was reduced by at least 27% in all cases. Supplementary loop belt tension was substantially reduced in the restrained configurations. Based on a model developed for the six-year-old child, the 95% risk of serious or greater abdominal injury associated with the unbraced condition was reduced to 35% and 60% for the nine- and 18-month-old, respectively. However, as these predictions are based on a model developed for an older and more structurally-developed child (see section 3.3.3), it is probable that they represent an under-estimation of the true risk of injury to the developing infant abdomen.

A modified brace position in which the infant was forward-facing and the heads of the infant and accompanying adult were laterally offset was evaluated for the nine-month-old in terms of head injury parameters. Compared to the traditional forward-facing position, in which occupant head alignment was considered as a worst case, this modified position resulted in further reductions in peak infant head contact load of 9% and 18% in the restrained and unrestrained conditions, respectively.

While the adoption of an emergency brace position was generally effective in reducing injury potential for the lap-held infant, injury parameter values observed in the braced condition remained at high levels according to automotive safety standards. In cases where the infant was unrestrained, the rearward-facing brace position did not have any clear benefits compared with the forward-facing brace position; it produced small-to-moderate reductions in head injury potential but strongly promoted neck injury potential.
9 Automotive child restraint systems: implementation in air transport

9.1 Introduction
Several studies have found that CRS, particularly those installed by novel methods, are capable of providing infant aircraft passengers with a sufficient level of protection under emergency landing conditions[2, 3, 6-8]. However, questions remain as to how CRS might be most safely implemented in air transport[2-4, 6, 10]. This chapter details the work carried out to answer the research questions presented in section 1.2 related to the use of CRS in air transport.

Three CRS installation methods were considered: the traditional aircraft seat lap belt and the novel ISOFIX and LATCH methods (see section 2.4.4). A top tether strap was not used except in the analysis detailed in section 9.7.

The experiments described in this chapter were carried out using the numerical model detailed in chapter four. Unless otherwise specified, the following seating configuration was used:

- The row-to-row seat pitch was 0.762 m (30 in.).
- The CRS was the forward-facing model described in section 4.3.2.7.
- The CRS occupant, where present, was a TNO P3 three-year-old ATD.
- The adult passenger, where present, was a Hybrid III 50th percentile male ATD.
- The CRS were installed in the aircraft seat in their corresponding reference positions (see section 4.2.4).

The three-year-old ATD was selected as the worst-case CRS occupant in terms of promoting CRS motion and increasing loading on the aircraft seat. This infant size was chosen to enable the comparison of some of the results of this analysis with those of the CASA study[2, 6].

9.2 Effect of CRS installation method on CRS and seat-back motion

9.2.1 Objective
The objective of this work was to determine the relative effects of CRS installation method on the translational and rotational behaviour of an occupied CRS throughout the impact sequence.
9.2.2 Method

Four experiments were carried out involving occupied CRS; three to measure CRS behaviour and effects on aircraft seat behaviour related to the use of the lap belt, ISOFIX and LATCH CRS installation methods and a fourth test to measure seat-back rotation behaviour for the case of an empty seat.

Two reference points on the CRS body (see Figure 9-1 below) were used to describe its motion. The lower point was chosen as it gives a good indication of whether the CRS base is at risk of sliding off the seat base cushion. The upper point was chosen because physical test footage revealed that it is generally the first (and sometimes only) point of contact between the CRS and the seat-back during impact.

![Figure 9-1: CRS reference points](image)

The parameters measured for use in this analysis were:

- Position of the CRS upper and lower reference points
- Angular displacement of the CRS in the vertical longitudinal plane
- Seat-back angular displacement, measured at free hinge

Seat-back rotation was measured as a precursor to the study of aft occupant injury described in later sections. Infant head and neck injury parameters were also measured for later analysis.
9.2.3 Results and discussion

9.2.3.1 CRS translation

Each of the three installation methods resulted in distinct CRS translational behaviour. Figure 9-2 below presents a comparison of the maximum forward translation of the CRS lower reference point for the lap belt, ISOFIX and LATCH cases.

![Figure 9-2: Effect of CRS installation method on CRS forward translation, measured at lower reference point.](image)

The greatest forward translation was observed in the lap belt case. By contrast, the LATCH case resulted in forward translation of approximately half of the value measured in the lap belt case. This difference is attributable to the more-favourable belt angle associated with the LATCH installation, approximately 50° above horizontal compared with the 72° belt angle associated with the lap belt installation. The effect of belt angle is also evident in Figure 9-3 below, which illustrates the path prescribed by the CRS lower reference point for each of the three configurations.

The small forward translation observed in the ISOFIX configuration was a result of the deformation of the lower anchorage bar and rotation of the CRS about the ISOFIX loops.
Figure 9-3: Comparison of CRS translational motion associated with lap belt, ISOFIX and LATCH installation methods, measured at lower reference point.

9.2.3.2 CRS rotation

Rotation behaviour varied in each of the three CRS installation methods (see Figure 9-4 below). The ISOFIX method effectively constrained the CRS to rotation about the ISOFIX loops, with very little vertical motion of the pivot point, resulting in the high level of forward rotation observed.
The translation allowed by the lap belt left less of the CRS base in contact with the base cushion; the CRS was therefore less-supported in the vertical direction and more able to rotate forwards. This effect occurred to a much lesser extent in the LATCH case, with a corresponding lower level of forward rotation.

### 9.2.3.3 CRS general motion

Each of the three installation methods restrained the CRS sufficiently to keep it in place on the aircraft seat base cushion. The motion of the CRS upper reference point was used to assess the combined effect of CRS translation and rotation on seat-back motion. The displacement of this point in the forward direction is plotted in Figure 9-5 below.
Figure 9-5 illustrates three key points:

1. The LATCH installation method was associated with a substantially smaller forward displacement of the upper aft edge of the CRS than the other two methods.
2. The ISOFIX installation method did not result in significantly smaller displacement of the upper aft edge of the CRS than the lap belt method.
3. The lap belt and LATCH methods both exhibited rebound motion to a greater extent than the ISOFIX method.

The more-severe rebound effects associated with the lap belt and LATCH methods apparent in Figure 9-5 above is consistent with the findings of physical tests[2]. Restraining forces were applied to the CRS during deceleration through direct contact between the CRS and the aircraft seat base cushion and a force transmitted through the CRS installation hardware (i.e., the lap belt, the lower anchorage, and the LATCH belt-ISOFIX bar combination). The application of this restraining force led to:

1. Compression of the seat base cushion. At the high compressive strain rates associated with this scenario, the foam comprising the cushion exhibits hysteretic behaviour. This
means that a significant amount of the CRS kinetic energy transferred to the seat base cushion is dissipated.

2. *Elongation of the webbing belt in the lap belt and LATCH cases.* Webbing belts generally exhibit a degree of elasticity at low strain, meaning that CRS kinetic energy transferred to the belt will reappear as kinetic energy (and therefore CRS motion) once the peak forwards inertial force on the CRS has passed.

3. *Permanent deformation of the lower anchorage bar in the ISOFIX and LATCH cases.* The permanent deformation seen in the lower anchorage bar after physical tests and numerical model simulations indicates that this component dissipates energy.

4. *Other small-scale deformation.* Small-scale deformation occurs in the aircraft seat structure and sled deceleration system but is not significant to the overall motion of the CRS.

In light of the above points, the better rebound behaviour observed in the ISOFIX case may be attributed to the degree of kinetic energy dissipation inherent to that installation method as a result of:

1. The rigid connection to the lower anchorage bar, as opposed to the use of a webbing belt.
2. The plastic deformation of the lower anchorage bar.
3. The motion of the CRS inherent to this method which leads to greater compression of the seat base cushion.

### 9.2.3.4 Seat-back motion

Seat-back rotation was limited to within half of the range observed in the ‘empty seat’ case regardless of CRS installation method. High-speed footage of the physical tests and animation generated by the numerical model both indicated that the location of the CRS upper reference point was generally the first, and sometimes only, point of contact between the CRS and the seat-back. A comparison of time history plots of the forward displacement of this point (see Figure 9-5) and seat-back angular displacement measured at the free hinge (i.e. without energy absorber) (Figure 9-6) for each of the three installation methods indicates that these two parameters are closely linked: greater displacement of this point allows greater seat-back motion.
In the LATCH case the seat-back initially rotated forward to within approximately 5° of the maximum value for the lap belt and ISOFIX cases. However, this was largely due to elastic deformation in the seat-back frame. The frame returned to its normal shape after initial contact with the CRS, resulting in a forward rotation of approximately 17°, 8° less than that observed in the lap belt and ISOFIX cases.

Figure 9-6: Forward angular displacement of seat-back, measured at free hinge.
9.3  Effect of CRS installation method on infant injury

9.3.1  Objective
The objective of this analysis was to compare the relative effects of different CRS installation methods (lap belt, ISOFIX and LATCH) and the seat-back break-over action on infant kinematics and head and neck injury potential.

9.3.2  Method
The experiments described in the previous section were repeated using a ‘baseline’ configuration in which the numerical model of the aircraft seat was modified to prevent the break-over action of the seat-back. The infant head and neck injury parameters from both series of experiments were then analysed and compared in order to determine the relative effects of both the CRS installation method and seat-back interaction on CRS occupant injury parameters.

9.3.3  Results and discussion

9.3.3.1  Infant kinematics
In every case tested, the combined CRS rotation and translation was sufficient to cause the lower part of the infant’s legs to make contact with the forward seat-back. Though generally minor in all cases, this was most pronounced in the lap belt CRS case (see Figure 9-7 below).

Figure 9-7: Infant lower leg contact with forward seat, lap belt CRS
9.3.3.2 Infant head injury

The LATCH method resulted in the minimum observed \( \text{HIC}_{15} \) scores for cases with and without seat-back break-over (see Figure 9-8). The maximum score was 494 observed in the lap belt case with no break-over. All values were below the allowable limit of 570 for a three-year-old set out in FMVSS 208.

![Figure 9-8: Infant \( \text{HIC}_{15} \) results for lap belt, ISOFIX and LATCH CRS installation methods.](image_url)

9.3.3.3 Infant neck injury

Figure 9-9 below presents infant neck injury parameter scores which were similar across all configurations. The critical neck injury mechanism in all cases was the tension-flexion combination. Neck injury parameter values are interpreted here in a comparative sense only in accordance with the level of agreement with physical test data evident in the validation process.
Figure 9-9: Infant neck injury results for lap belt, ISOFIX and LATCH CRS installation methods
9.4 Effect of CRS and its installation method on aft passenger injury

9.4.1 Objective

The objective of this analysis was to compare the relative effects of the presence of an occupied CRS installed by the lap belt, ISOFIX and LATCH on the injury potential of an adult seated directly aft.

9.4.2 Method

Four experiments were carried out in total; in three, a 50th percentile Hybrid III numerical ATD was seated directly aft of a seat in which an occupied CRS was installed using either lap belt, ISOFIX or LATCH methods. In the fourth experiment, the forward seat was left empty. The CRS, where present, was occupied by a three-year-old ATD.

The injury parameters measured in the adult occupant were head linear acceleration at the centre of gravity and neck axial force and fore-aft moment. Infant injury parameters were also measured for use in an analysis of the effect of the aft passenger on infant injury (see section 9.8).

9.4.3 Results and discussion

9.4.3.1 Aft passenger head injury

Each CRS installation method, as well as the configuration with an empty forward seat, resulted in aft passenger HIC₁₅ scores significantly above the limit of 700 set out in FMVSS 208 (see Figure 9-10 below).
Figure 9-10: The effect of the presence of a CRS and its installation method on aft passenger HIC\textsubscript{15} score.

The LATCH method resulted in the highest HIC\textsubscript{15} score with a value of 1779. Graphical model output revealed that in the LATCH case the CRS was positioned further aft at the time of aft passenger head impact than in the lap belt and ISOFIX cases. For this reason the tray table made contact with the CRS through the seat-back fabric during aft passenger head impact, resulting in contact force being transferred to the aft passenger head and a consequently higher HIC\textsubscript{15} score for the LATCH case.

The HIC\textsubscript{15} scores were used in conjunction with the head injury risk curves discussed in section 3.4.1 to determine the effect of CRS installation method on the risk of severe-or-greater brain injury or skull fracture for the aft passenger. The empty seat configuration was associated with an 18% risk of severe-or-greater brain injury (see Figure 9-11 below). The ISOFIX method resulted in the lowest head injury potential of the three CRS installation methods: for this case the HIC\textsubscript{15} score of 1173 is associated with an approximately 7% increase in risk of severe or greater brain injury over the empty seat configuration; the LATCH method resulted in the greatest increase of approximately 62%.
The predictions of skull fracture risk closely mirrored those for brain injury; the only notable difference was in the LATCH case where the skull fracture risk of approximately 75% was slightly below the value of 80% predicted for brain injury.

9.4.3.2 Aft passenger neck injury

Neck injury criterion scores are presented in Figure 9-12 below. The critical neck injury mechanism in each case was the tension-extension combination. The peak $N_{\text{ij}}$ score invariably occurred just prior to head impact; upper neck moment generally became insignificant after impact, while axial force turned from tensile to compressive. Interestingly, the presence of a CRS installed by any of the three methods tested served to reduce neck injury potential; this reduction was greatest in the LATCH case and least in the lap belt case. This follows a trend identified in the CASA study[2]. Neck injury parameter values are interpreted here in a comparative sense only in accordance with the level of agreement with physical test data evident in the validation process.

Figure 9-11: The effect of the presence of a CRS and its installation method on the risk of severe or greater brain injury (AIS ≥ 4) to aft passenger.
Figure 9-12: The effect of the presence of a CRS and its installation method on aft passenger neck injury potential.

A step was observed in the neck moment signal at 80 ms in numerical model results, an effect also present in physical test results. Model graphical output and physical test footage revealed that this effect was caused by the impact of the ATD’s elbows with the forward armrests, as shown below in Figure 9-13.

Figure 9-13: Aft passenger elbow impact with forward armrest, t = 80 ms, CASA footage.
This elbow-to-armrest contact occurred for all CRS installation methods in both the numerical model and physical tests. This contact caused the ATD upper torso to be partially constrained in forward rotation prior to head impact. The effect of this on upper neck moment is evident in Figure 9-14 below which compares the cases of an empty forward seat, an empty forward seat armrests removed and the standard configuration with a CRS installed using ISOFIX.

![Figure 9-14: Effect of elbow-to-armrest contact on neck upper moment](image)

A comparison of these signals suggests that the peak in upper neck moment at approximately 90 ms was due solely to the elbow-to-armrest contact; the presence of a CRS serves to reduce the magnitude of this peak, resulting in the observed decrease in neck injury potential.
9.5 Effect of empty CRS on aft passenger injury

9.5.1 Objective
The objective of this work was to analyse a scenario where an empty CRS may cause reduced seat back motion and therefore increased aft passenger injury. The mass of the infant ATD used in the foregoing analysis of aft passenger injury was close to the CRS maximum allowable occupant mass of 18 kg. During deceleration the CRS occupant exerts a force on the CRS in the forward direction, acting to ‘pull’ the CRS away from the rotating seat-back.

9.5.2 Method
Three experiments were conducted in which a 50th percentile male ATD was seated directly aft of a seat containing an unoccupied CRS installed using the lap belt, ISOFIX and LATCH methods. Adult head and neck injury parameters were measured and compared with the results of the analysis detailed in the previous section.

9.5.3 Results and discussion

9.5.3.1 Aft passenger head injury
The absence of the CRS occupant produced a significant effect in all cases, leading to either increased or decreased aft passenger head injury depending on CRS installation method (see Figure 9-16 below). In each case this was a result of reduced seat-back rotation caused by subdued CRS motion. The lap belt case is presented as an example in Figure 9-15 below, where the seat-back angular displacement measured in this analysis is plotted alongside data from the corresponding experiment in section 9.2 with the CRS occupied by a three-year-old. In this case seat-back rotational displacement was approximately 5° less at the time of aft passenger impact on account of the CRS being empty.
In the lap belt case the reduced seat-back rotation caused the aft passenger’s head to impact the rigid structure at the top of the tray table, resulting in an extremely high HIC\textsubscript{15} score of 3548. This effect occurred to a lesser extent in the LATCH case where a moderate increase in HIC\textsubscript{15} was observed. The ISOFIX case, however, resulted in a modest reduction in HIC\textsubscript{15}. The reduced motion of the CRS and seat-back in this case was beneficial; the seat-back was still rotating forward at the time of impact and the angular displacement at this time was such that the aft passenger’s head impacted the relatively compliant centre of the tray table.

Figure 9-15: Effect of the absence of the CRS occupant on seat-back rotation, lap belt method.
The presence of an empty CRS installed using the lap belt and LATCH methods increased the aft passenger’s risk of severe-or-greater brain injury from 25% and 80% respectively to greater than 99%. In the ISOFIX method this risk was reduced from 25% to approximately 15%.

9.5.3.2 Aft passenger neck injury

Figure 9-17 below illustrated the minor influence of the absence of the CRS occupant on aft passenger HIC15 score. The greatest effect as observed in the LATCH case where the neck injury criterion value was reduced by approximately 10%. Neck injury parameter values are interpreted here in a comparative sense only in accordance with the level of agreement with physical test data evident in the validation process.
Figure 9-17: Effect of the absence of the CRS occupant on aft passenger neck injury
9.6 Effect of seat pitch on aft passenger injury

9.6.1 Objective
The objective of this analysis was to investigate the effect of seat pitch on aft passenger head and neck injury potential for configurations involving occupied CRS installed using the aircraft lap belt, ISOFIX, and LATCH.

9.6.2 Method
A 50th percentile male ATD was seated directly aft of a seat in which a CRS was installed by the lap belt, ISOFIX or LATCH methods at varying seat pitch. Twenty-four experiments were conducted, considering each of the three CRS installation methods at eight seat pitches in the range of 28 to 34 inches (0.762 – 0.864 m) in one inch (0.0254 m) increments. The CRS was occupied by a three-year-old ATD. Aft passenger head and neck injury parameters were measured.

9.6.3 Results and discussion

9.6.3.1 Aft passenger head injury
Figure 9-18 below illustrates the variation in head injury potential with seat pitch. In each configuration the aft passenger’s head contacted the forward seat-back, resulting in a HIC_{15} value greater than the critical value of 700 defined in FMVSS 208. The minimum HIC_{15} observed across all configurations was 1173 in the ISOFIX case at a pitch of 30 inches. The lowest value observed in the lap belt case was 1386 at the same pitch. The LATCH case recorded a minimum of 1228 at a pitch of 34 inches; however, this result is due to an anomaly: during head impact, the tray table deformed to the extent that it was no longer supported by the seat-back frame on one side. The next lowest score in the LATCH case was 1597 at a pitch of 29 inches.
The animations produced by the numerical model were useful in determining the reason for the minimum HIC$_{15}$ score occurring at 29 or 30 inches in each of the three installation methods: at smaller seat pitches, the adult’s head impacted the forward seat at the top of the tray table, which is supported behind by the stiff carbon composite seat-back frame. At greater seat pitches, the head impacts at the more-compliant centre of the tray table; however, as the pitch increases, so does the relative velocity between the head and tray table at the time of impact.

9.6.3.2 Aft passenger neck injury
Aft passenger neck injury values, presented in Figure 9-19 below, were less than the ‘no CRS’ value measured in section 9.4 (i.e. no CRS present) for all seat pitches and all CRS installation methods except the 28” ISOFIX case where this value was slightly exceeded. For all installation methods neck injury criterion values generally decreased with increased seat pitch.
As the seat pitch increased, the head contacted the forward seat-back later in the impact sequence when the upper body had rotated further forward. This resulted in a progressive decrease in the upper body inertia acting to load the neck in flexion and therefore reduced $N_{ij}$ values.

Figure 9-19: Effect of seat pitch on aft passenger neck injury
9.7 Effect of top tether on CRS motion and infant injury

9.7.1 Objective
The objective of this analysis was to determine whether the use of a top tether is effective in reducing both CRS motion and infant injury potential.

9.7.2 Method
Two series of three experiments were conducted; both series considered the lap belt, ISOFIX and LATCH CRS installation methods. In the first series no modification was made to the aircraft seat in which the CRS was installed; i.e., seat-back break-over behaviour was controlled by the energy absorber described in section 4.3.2.3. In the second series the aft seat-back was prevented from breaking over to determine whether such a modification is able to increase top tether efficacy as suggested by Bathie[6]. The top tether routing employed was representative of a typical installation used in practice by airlines (see Figure 9-20 below).

![Figure 9-20: Model representation of a typical top tether installation.](image)

The top tether was modelled using finite element belts 30 mm wide with the same material properties as the aircraft seat lap belt. A pre-simulation was used to load the top tether to a tension of 10 N.
In each of the experiments the CRS was occupied by a three-year-old ATD. The results of this analysis were compared with those of sections 9.2 and 9.3. The parameters measured were:

- Position of the CRS lower reference point
- Angular displacement of the CRS
- Infant head acceleration
- Infant neck axial force and fore-aft moment

9.7.3 Results and discussion

9.7.3.1 CRS motion

In all experiments the lower part of the infant’s legs made minor contact with the forward seat-back; this was also the case in experiments where no top tether was used.

The implementation of a top tether was found to very slightly reduce the forward translation of the CRS lower reference point (see Figure 9-21 below). A small further decrease in translation resulted from preventing the seat-back from breaking over.

![Figure 9-21: Effect of top tether on maximum forward translation of CRS lower reference point](image)

The effect of preventing the seat-back from breaking over was partially negated by the flexibility of the seat-back frame; some elastic deformation of this component occurred due to tension in the top tether.
The implementation of a top tether resulted in a small decrease in CRS forward rotation in the lap belt and ISOFIX cases (see Figure 9-22 below). In the LATCH case, however, the path of the top tether over the seat-back head cushion led to increased CRS rotation. This behaviour was an effect of the lower position of the CRS in the LATCH case; the routing of the top tether over the seat-back head cushion meant that tension in the top tether acted to pull the CRS forward.

![Figure 9-22: Effect of top tether on maximum forward rotation of CRS](image)

The effect of preventing seat-back break-over on CRS rotation was minor, with reductions of two to three degrees observed in the lap belt and ISOFIX cases. In the LATCH case, CRS rotation was reduced but still slightly greater than the result where no top tether was used.

### 9.7.3.2 Infant injury

Infant head injury results are presented in Figure 9-23 below. For the lap belt and ISOFIX cases, where the top tether served to slightly reduce CRS motion, infant head injury was increased. The increase was slight for cases where the seat-back was able to break over and the tether was largely ineffective. However, for the cases where seat-back break-over was prevented, HIC$_{15}$ scores for the lap belt and ISOFIX cases increased by 30% and 17% respectively. For the LATCH case the top tether served to slightly increase CRS rotation, leading to a corresponding decrease in head injury potential.
The effect of top tether use on infant neck injury criterion was minimal except in the lap belt case with no seat-back break-over where it resulted in a 16% increase (see Figure 9-24 below). This is attributed to the decreased CRS forward rotation observed in this case. Neck injury parameter values are interpreted here in a comparative sense only in accordance with the level of agreement with physical test data evident in the validation process.

Figure 9-23: Effect of top tether on infant head injury

Figure 9-24: Effect of top tether on infant neck injury
9.8 Effect of aft passenger on infant injury

9.8.1 Objective
The objective of this analysis was to determine the degree to which the presence of an adult seated directly aft of a CRS might affect the safety of the CRS occupant.

9.8.2 Method
The results of the analyses described in sections 9.3 and 9.4 were compared to determine the effect of a passenger seated aft of a CRS on the head and neck injury potential of the CRS occupant.

9.8.3 Results and discussion
For all three CRS installation methods, the presence of an aft passenger had the effect of decreasing infant head injury (see Figure 9-25 below). This effect was most apparent in the lap belt case with a reduction in HIC$_{15}$ of approximately 16%.

![Figure 9-25: Effect of aft passenger on infant head injury](image)

Similarly, infant neck injury was decreased in each case by the presence of an aft passenger (see Figure 9-26 below). The greatest decrease was observed in the LATCH case where the neck injury criterion value was reduced by 17%. Neck injury parameter values are interpreted
here in a comparative sense only in accordance with the level of agreement with physical test data evident in the validation process.

**Figure 9-26: Effect of aft passenger on infant neck injury**

Contact between the aft passenger and the forward seat-back caused the seat-back to exert a force on the CRS, in turn causing greater forward rotation of the CRS than would otherwise have occurred. The result was a more gradual deceleration of the infant’s head, particularly during the aft passenger head impact occurring at approximately 90 – 100 ms. This effect is evident in Figure 9-27 below.
Figure 9-27: Effect of aft passenger on infant head acceleration, lap belt CRS installation method.
9.9 Conclusion

Several series of experiments were conducted to examine the safety-related effects of a variety of seating configuration parameters involving CRS. Three CRS installation methods were considered: the traditional aircraft seat lap belt and the novel ISOFIX and LATCH methods.

Infant head injury potential remained within the limit prescribed by automotive occupant protection standards[56] regardless of CRS installation method when the CRS was installed in an aircraft seat with normal break-over function. Infant neck injury potential was not significantly affected by CRS installation method. The impact of an adult seated aft of a CRS was found to slightly reduce head and neck injury potential for the CRS occupant.

The novel installation methods were found to significantly reduce CRS forward translation compared with the lap belt; the LATCH method reduced CRS translation by approximately 50% and the ISOFIX method reduced translation by over 90%. Despite this, the combination of CRS motion and infant flailing resulted in contact between the forward seat and the infant’s legs in every case tested.

While the ISOFIX method was highly effective in reducing CRS translation, it was found to promote CRS rotation. Compared with the lap belt method, the ISOFIX method resulted in an approximately 50% increase in rotation; LATCH, meanwhile, resulted in a 50% decrease. The increased rotation observed in the ISOFIX case, however, was a significant factor in that method’s enhanced rebound performance. Rather than energy being stored in a webbing strap as in the other two methods, the ISOFIX method results in energy being dissipated by the rate-sensitive foam of the aircraft seat base cushion.

As has been found previously[3, 6], the use of a top tether in the installation of a CRS in an unmodified aircraft seat does not significantly affect CRS motion or infant head or neck injury potential. A recommendation from prior studies[3, 6] that the efficacy of the top tether may be increased by preventing the seat-back from rotating forward was tested. This modification was found by the present study to promote infant head and neck injury due to slightly diminished CRS motion.

Prior studies have raised the concern that CRS installed by novel methods may negatively interact with the break-over action of the aircraft seat-back, potentially increasing injury to the infant CRS occupant[10] and an adult seated directly aft[2]. This study found that CRS installation method and CRS occupant size both influenced seat-back break-over behaviour. When occupied by a three-year-old, CRS installed using the aircraft seat lap belt and ISOFIX
constrained the forward rotation of the seat-back to approximately half of its maximum potential value. The LATCH CRS further constrained seat-back rotation, restricting angular displacement to approximately 35% of its maximum potential value. While this interaction between the CRS and the seat-back was found to have an insignificant effect on infant head and neck injury potential, it was implicated in the increased head injury potential observed in an adult seated directly aft.

With the forward seat entirely unoccupied the risk of life-threatening brain injury to the aft passenger was found to be approximately 18%. This risk varied according to CRS installation method and occupant size. With the CRS occupied by a three-year-old ATD, the smallest risk was observed in the ISOFIX case at 25%, next greatest was the lap belt case at 40%; the LATCH method resulted in the most significant risk at 80%.

Compared to configurations where the CRS was occupied by a three-year-old ATD, the absence of an occupant in CRS installed using the lap belt and LATCH methods was found to significantly increase head injury potential for an adult seated aft. The risk of life-threatening brain injury in these configurations rose to greater than 99%. Conversely, an unoccupied CRS installed using ISOFIX resulted in a moderate reduction in risk from 25% to 15%.

Bathie[2] identified a phenomenon whereby the presence of a CRS in the forward seat serves to reduce neck injury potential for the aft-seated adult. The present study found that the abovementioned cause of increased head injury potential the adult acted to mitigate neck injury potential in that occupant regardless of CRS installation method or the presence of a CRS occupant. The bending moment in the aft passenger’s neck was found to rapidly increase prior to head impact due to the passenger’s elbows making contact with the forward armrests. The earlier occurrence of head contact resulting from the presence of a CRS in the forward seat effectively prevented the neck bending moment from reaching its potential maximum.

Seat pitch was found to influence the head and neck injury potential of a passenger seated aft of a CRS. For all three CRS installation methods, head injury potential was generally minimised at seat pitches of 29 to 30 inches; neck injury potential generally decreased slightly with increasing seat pitch.
10 Conclusions

This study represents the first detailed evaluation and comparison of injury potential across a variety of air transport seating configurations involving infants. A numerical model of a typical air transport seating arrangement was developed and validated against data from physical experiments. The validation process revealed that important output parameters such as head acceleration were generally in good agreement with physical test data while others, such as neck axial force, were not. Hence, the model was treated as ‘partially validated’. This model was used to develop answers to a series of research questions that arose from a review of Australian and international transport safety regulations, air transport accident reports and past studies relevant to the safety of infant air transport passengers. The conclusions of each of the individual analyses carried out by this study are summarised below as responses to those research questions.

10.1 The safety of the lap-held infant

The first phase of this research project considered the broad question, ‘what are the safety risks for the lap-held infant?’ This was answered in three parts:

i. How is the risk of injury for the lap-held infant affected by the use or non-use of a supplementary loop belt?

The lap-held seating configuration does not represent a means of protecting an infant from injury in an emergency landing scenario regardless of restraint condition. This confirms the findings of several prior studies[2-5, 7]. An unrestrained lap-held infant is likely to be projected through the aircraft cabin during the impact sequence, an outcome that has been observed in past experimental investigations[5] and actual aircraft accidents[20, 22]. An infant restrained by a supplementary loop belt is at significant risk of serious abdominal injuries such as perforation and laceration of the small bowel and large bowel injuries resulting in gross faecal contamination of the peritoneal cavity and significant blood loss into the same site.

ii. What are the characteristics of the forces acting on the head of the lap-held infant?

During the impact sequence the infant’s body, particularly the head, becomes compressed between the forward seat and the adult’s body irrespective of restraint condition. No prior study has analysed the forces acting on the infant head in this scenario. Due to the
compressive force the head injury criterion, when evaluated in the traditional way, is likely to under-predict injury potential for the lap-held infant.

The compressive force acting on the head of the lap-held infant has been characterised to enable a future detailed prediction of injury risk using a paediatric head finite element model. A force-based evaluation of the head injury criterion resulted in values an order of magnitude greater than those specified in automotive safety standards.

iii. Are the recommended emergency brace positions involving lap-held infants effective in preventing injury to the infant?

This study presents the first evaluation of the efficacy of emergency brace positions involving lap-held infants. While emergency brace positions are generally effective in reducing the magnitude of the load applied to the lap-held infant’s body, the infant remains at risk of injury. A modified brace position in which the infant is forward-facing and the occupants’ heads are laterally offset results a better outcome in terms of infant head injury potential than other recommended lap-held brace positions. In cases where the infant is unrestrained, a rearward-facing brace position does not have any clear benefits compared with a forward-facing brace position. The rearward-facing position produces a small-to-moderate reduction in head injury potential but strongly promotes neck injury potential.

10.2 The safety of automotive child restraints in air transport

The second phase of this study considered the question, “What are the safety risks associated with the use of automotive child restraints in air transport, particularly those installed by novel methods?” This work represents the first parametric analysis of the performance of automotive CRS in aircraft. The work was undertaken in three parts:

i. How do novel CRS installation methods affect the dynamic performance of the CRS and what is the corresponding effect on injury potential for the infant?

Three different CRS installation methods were studied: the traditional method utilising the aircraft seat lap belt as well the novel ISOFIX and LATCH methods. Compared with the lap belt method, the novel ISOFIX and LATCH methods do not significantly affect head or neck injury potential for the infant. The forward translation of the CRS is significantly improved by novel installation methods; the LATCH method reduces forward translation by approximately 50% on account of the more-favourable belt path associated with this method, while the ISOFIX method reduces forward translation by approximately 90% on due to the rigid
connection between the CRS and seat structure. The rebound motion of a CRS installed using ISOFIX is significantly better-controlled than that of a CRS installed by either the lap belt or LATCH.

ii. How do novel CRS installation methods alter the behaviour of the aircraft seat and what is the corresponding effect on the safety of an adult seated directly aft?

When occupied by a large infant such as the three-year-old considered by this study, the presence of a CRS significantly affected the break-over behaviour of the aircraft seat regardless of installation method. When installed using the lap belt and ISOFIX, CRS constrain the forward rotation of the seat-back to approximately half of its maximum potential value. The LATCH method further constrains seat-back rotation, restricting angular displacement to approximately 35% of its maximum potential value.

In general, the presence of a CRS in the seat directly forward of an adult passenger results in increased head injury potential for the adult regardless of installation method. The ISOFIX installation method presents the least risk of the three methods considered. Compared with a vacant forward seat, the presence of a CRS installed using ISOFIX and occupied by a large (three-year-old) infant results in a 7% increase in the risk of life-threatening brain injury; where the CRS is unoccupied, the risk is reduced by 3%.

The lap belt and LATCH CRS installation methods most greatly promote head injury potential for an adult passenger seated aft. Compared with a vacant forward seat, CRS occupied by a large infant and installed by the lap belt or LATCH methods increase the risk of life-threatening brain injury to the adult by 22% and 62%, respectively. For these methods the risk of injury increases significantly when the CRS is unoccupied, exceeding 99% in both cases.

Bathie[2] identified a trend indicating that the presence of a CRS in the forward seat has a positive effect on the neck injury potential of an adult seated aft. This study produced further evidence of this effect and found that it occurs regardless of whether the CRS is occupied.

Seat pitch influences the head and neck injury potential of a passenger seated aft of a CRS. For all three CRS installation methods, head injury potential is generally minimised at seat pitches of 29 to 30 inches; neck injury potential generally decreases slightly with increasing seat pitch.
iii. Are current practices of top tether use effective in controlling CRS motion and what is the corresponding effect on infant injury potential?

This study supports the results of previous work[3, 6] showing that the top tether in CRS installation does not provide any benefit in terms of CRS motion or injury risk for the CRS occupant. The prevention of seat-back break-over does not improve the efficacy of the top tether; in fact, this serves to promote infant head and neck injury.

10.3 Limitations and future studies

While this study adds useful information to the body of knowledge on the safety of infant restraint systems in air transport, it has been developed and conducted within the bounds of several limitations.

The principal limitation was that the numerical model used to carry out experimental work was validated against a limited set of data from physical experiments which were not carried out with future modelling in mind. The physical experiments used as a basis for the model encompassed multiple occupant configurations; however, each configuration was tested only once and for one load case. The variability of the results of the physical experiments was therefore unable to be evaluated. This meant that during the validation phase, model output was compared with a single curve or data point rather than with an envelope or spectrum of possible results. Judgement was therefore required to determine whether or not a parameter matched closely enough with reality for use in injury prediction. Future studies should evaluate the variability of tests of this nature to enable this effect to be accounted for in the interpretation of results and thereby allow stronger conclusions to be drawn.

This study considered multiple occupant configurations and restraint methods; however, only one CRS model was tested and in only one orientation. This limitation had minimal effect on the conclusions of the present study because the CRS model tested is typical of its category and the forward-facing orientation tested was identified as a worst-case. However, future studies should identify the relationship between CRS geometry, centre of gravity location and belt path or anchor location in terms of CRS kinematics in an air transport environment. Such information is potentially useful in setting design approval parameters on CRS for use in this environment.

The numerical models of anthropomorphic test devices (ATDs) used to assess occupant kinematics are essentially twice-removed from the human beings they represent. The
requirement that ATDs accurately represent the biomechanical response of the live human is outweighed by the requirements of repeatability and reusability [42]. While there is generally a paucity of biomechanical tolerance data for infants [70], work has been done towards defining a relationship between the impact response of the human infant head and that of ATDs [47, 110]. Future numerical studies should implement infant-sized numerical ATD models with head contact characteristics which have been modified for increased biofidelity. This approach would potentially allow head injury risk to be evaluated for the infant as it was in this study for the adult.

10.4 Summary and recommendations

In a severe but survivable accident an infant seated on the lap of an adult, whether unrestrained or restrained by a supplementary loop belt, may receive an impact to the head resulting in a head injury criterion value an order of magnitude greater than that allowable under automotive safety standards. Based on this finding the carriage of infants on the lap should be prohibited. Until this is the case, advice on emergency brace positions involving lap-held infants should be revised to include a lateral offset between the head of the infant and that of the adult.

The use of an automotive child restraint reduces infant head injury potential to within the limit prescribed by US motor vehicle occupant protection standards and prevents the infant from substantial contact with the forward seat. Automotive child restraints therefore represent a means of restraining infant air transport passengers that is substantially safer than the lap-held configuration. Based on this finding the use of automotive child restraints to restrain infant passengers should be mandated. Furthermore, the use of a top tether in CRS installation is ineffective in mitigating infant injury and any requirement for its use in aircraft CRS installations should be removed.

The three CRS installation methods considered by the present study were each found to significantly affect the break-over function of the aircraft seat, a feature which is intended to mitigate head injury. Two of the methods, the lap belt and LATCH, were found to significantly promote head injury to an adult seated directly aft. The third method, ISOFIX, slightly promoted head injury to the aft occupant when the CRS was occupied and slightly mitigated injury when it was unoccupied. Based on this finding, CRS should be installed in outboard seats directly forward of a bulkhead, unoccupied space or another CRS. A typical airline seat configuration in a narrow-body aircraft with approximately fifteen rows of
economy class seating aft of the over-wing exits could thus accommodate up to thirty infants in CRS without an adult being seated behind a CRS or a seat being left vacant.
References


27. Civil Aviation Safety Authority, *Passenger safety information: Guidelines on content and standard of safety information to be provided to passengers by aircraft operators.* 2004, Civil Aviation Advisory Publication: Canberra.


47. Loyd, A.M., *Studies of the Human Head from Neonate to Adult: An Inertial, Geometrical and Structural Analysis with Comparisons to the ATD Head*, in Department of Biomedical Engineering. 2011, Duke University.


100. TASS BV, MADYMO Quality Report Release Update. 2010: Rijswijk.


Appendix A: Foam material characteristics

Three foam materials were present in the model, designated AF60, LD24, and LRGR45. Presented below are the MADYMO input cards used to define these materials and their loading characteristic curves. In each material model, a rate-depended stress scaling[93] was applied according to the function “func_foam_rate”. The unloading function was identical to the loading function and was scaled by a factor of 0.001 using FUNC_USAGE.2D.

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Appendix B: Model verification

A representative model verification example is presented below. Figures B-1 through B-3 below present comparisons of numerical model and physical test kinematics at elapsed times of 50, 100 and 150 ms, respectively. The seating configuration presented here is the aft passenger lap belt configuration discussed in section 5.8.4.

Figure B-1: Verification of numerical model kinematics against physical test; CRS installed by lap-belt occupied by three-year-old ATD, Hybrid III 50th percentile male ATD seated aft.

Time 50 ms.
Figure B-2: Verification of numerical model kinematics against physical test; CRS installed by lap-belt occupied by three-year-old ATD, Hybrid III 50th percentile male ATD seated aft. Time 100 ms.
Figure B-3: Verification of numerical model kinematics against physical test; CRS installed by lap-belt occupied by three-year-old ATD, Hybrid III 50th percentile male ATD seated aft. Time 150 ms.

Figure B-4 below presents an energy balance plot for the abovementioned configuration. An energy balance plot was used to ensure that no unexpected changes in energy occurred in the numerical model throughout the simulation.
Figure B-4: Energy balance plot; CRS installed by lap-belt occupied by three-year-old ATD, Hybrid III 50th percentile male ATD seated aft.

Figure B-5 below presents the seat/sled reaction loads for the rear seat in the abovementioned configuration. This load was not measured in the physical tests; however, it was assessed during the verification phase to ensure that there were no unexpected effects throughout the simulation.

Figure B-5: Sled reaction loads for aft seat; CRS installed by lap-belt occupied by three-year-old ATD, Hybrid III 50th percentile male ATD seated aft. Positive directions: Fx - aft, Fy - right, Fz - up.
Appendix C: Validation examples

Three representative model validation examples are presented below with a graphical comparison of signals and also the values of the validation metrics.

Example one

Model prediction of adult head acceleration (see Figure C-1) was generally in excellent agreement with physical test results.

![Graph showing simulation vs. physical test acceleration over time](image)

**Figure C-1: Aft passenger head resultant acceleration, lap belt CRS case.**

<table>
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<tr>
<th>Metric</th>
<th>Description</th>
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<tbody>
<tr>
<td>C = 3.3%</td>
<td>A comprehensive error factor of 3.3% indicates a good match between signals.</td>
</tr>
<tr>
<td>$M_{\text{pos}} = -4.6%$</td>
<td>The simulation positive peak is smaller in magnitude than the physical test positive peak by a value of 4.6% of the physical test positive peak.</td>
</tr>
<tr>
<td>$W_{\text{pos}} = 100.0%$</td>
<td>The magnitude of the physical test positive peak is 100% of the amplitude of the physical test signal.</td>
</tr>
<tr>
<td>$M_{\text{neg}} = N/A$</td>
<td>No part of the physical test signal is negative.</td>
</tr>
<tr>
<td>$W_{\text{neg}} = N/A$</td>
<td>No part of the physical test signal is negative.</td>
</tr>
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Example two

Model prediction of adult neck injury parameters (see Figure C-2) was generally in reasonable agreement with physical test results; however, the degree of match was not sufficient to allow injury prediction.

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<td>$C = 28.1%$</td>
<td>A comprehensive error factor of 28.1% indicates a reasonable match between signals.</td>
</tr>
<tr>
<td>$M_{pos} = -23.9%$</td>
<td>The simulation positive peak is smaller in magnitude than the physical test positive peak by a value of 23.9% of the physical test positive peak.</td>
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<tr>
<td>$W_{pos} = 62.3%$</td>
<td>The magnitude of the physical test positive peak is 62.3% of the amplitude of the physical test signal.</td>
</tr>
<tr>
<td>$M_{neg} = -0.6%$</td>
<td>The simulation negative peak is smaller in magnitude than the physical test positive peak by a value of 0.6% of the physical test negative peak.</td>
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<tr>
<td>$W_{neg} = 37.7%$</td>
<td>The magnitude of the physical test negative peak is 37.7% of the amplitude of the physical test signal.</td>
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</table>

Figure C-2: Aft passenger upper neck axial force, ISOFIX CRS case.
Example three

Model prediction of adult leg injury parameters (see Figure C-3) was generally poor. The model was not used to assess or compare leg injury.

![Figure C-3: Aft passenger right femur axial force, baseline case.](image)

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<td>A comprehensive error factor of 95.4% indicates a poor match between signals.</td>
</tr>
<tr>
<td>$M_{\text{pos}} = 131.8%$</td>
<td>The simulation positive peak is larger in magnitude than the physical test positive peak by a value of 131.8% of the physical test positive peak.</td>
</tr>
<tr>
<td>$W_{\text{pos}} = 29.1%$</td>
<td>The magnitude of the physical test positive peak is 29.1% of the amplitude of the physical test signal. The positive peak is significant, but not as significant as the negative peak.</td>
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<td>$M_{\text{neg}} = -18.9%$</td>
<td>The simulation negative peak is smaller in magnitude than the physical test positive peak by a value of 18.9% of the physical test negative peak.</td>
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<tr>
<td>$W_{\text{neg}} = 70.9%$</td>
<td>The magnitude of the physical test negative peak is 70.9% of the amplitude of the physical test signal.</td>
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Example four

As with adult occupants, model prediction of infant head acceleration (see Figure C-4) was generally in excellent agreement with physical test results.

![Figure C-4: 18-month-old infant head resultant acceleration, restrained lap-held configuration.](image)

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<td>The simulation positive peak is smaller in magnitude than the physical test positive peak by a value of 3.1% of the physical test positive peak.</td>
</tr>
<tr>
<td>$W_{\text{pos}} = 100%$</td>
<td>The magnitude of the physical test positive peak is 100% of the amplitude of the physical test signal.</td>
</tr>
<tr>
<td>$M_{\text{neg}} = \text{N/A}$</td>
<td>No part of the physical test signal is negative.</td>
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<tr>
<td>$W_{\text{neg}} = \text{N/A}$</td>
<td>No part of the physical test signal is negative.</td>
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