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**Mechanisms of head injuries in road  
traffic accidents; a potential solution for  
data collection**

MPhil/PhD STM895 Postgraduate research skills in science,  
maths & computing

**by J A Carroll**

**A697847X**

# **Mechanisms of head injuries in road traffic accidents; a potential solution for data collection**

PhD Thesis

**by J A Carroll**

**Prepared for: Project Record: A697847X**

**Organisation: The Open University,**

**TRL - Affiliated Research Centre**

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This Thesis has been prepared for The Open University for the sole purpose of supporting the application of Jolyon Carroll for a PhD.

The views expressed are those of the author and not necessarily those of The Open University or TRL.

<b>Name</b>	
<b>Director of studies</b>	Dr. David Hynd
<b>Supervisor</b>	Prof. Andrew Parkes



# Thesis abstract

Despite advances in road safety, head injuries still account for many of the most serious and fatal injuries in road traffic accidents. This PhD thesis provides a summary of knowledge regarding the current position in head injury research with regard to:

- Previous assertions as to head injury mechanisms
- Existing head injury criteria
- The availability of data to explore potential confounding factors in predicting head injury risk and to propose or validate a new injury criterion or criteria.

On the basis of the existing information the question was posed, whether it is possible to validate advanced head injury criteria and head models using additional (new) head injury case data so as to make their application more robust in efforts to mitigate future injuries. In order to answer this question, priority was given to the pursuit of new data, offering six degree of freedom time-series data with detailed information on the exact injuries sustained.

A working in-ear sensor system was deemed to offer a potential solution in obtaining elusive data regarding the kind of impact events that could cause head injuries for road users. An in-ear accelerometer system used by the FIA Institute was evaluated through experimentation. Then a low-cost solution was developed with the aim to give similar sensor performance for a wider market of potential wearers.

The prototype low-cost sensor system was evaluated in a small series of drop tests and also in a very small real-world data collection trial. This evaluation identified a series of issues that need to be resolved before the system can be used to generate valuable data. A viable system is not ready immediately, but could be following modifications to the prototype system evaluated. Taking this revised system, the next step would be to initiate a larger trial to start the collection of high fidelity data and impact event details; in order to address the need for such information and confirm that even the low-cost system would be fit for that purpose.

# Contents

<b>List of figures</b>	<b>v</b>
<b>List of tables</b>	<b>xi</b>
<b>Thesis publications</b>	<b>xii</b>
<b>1 Introduction</b>	<b>1</b>
<b>2 Background</b>	<b>3</b>
2.1 Kinematic theories of head injury	3
2.2 Description of structures within the head	5
2.3 Head injury	12
2.4 Motion described through acceleration	31
2.5 Summary	32
<b>3 Head injury mechanisms</b>	<b>33</b>
3.1 Car occupants	38
3.2 Vulnerable road users	40
3.3 Existing criteria	43
3.4 Confounding factors	71
<b>4 Comments on existing criterion effectiveness</b>	<b>83</b>
4.1 APROSYS Project findings	83
4.2 New analysis of published data	84
4.3 Summary of current position	95
<b>5 Advances needed</b>	<b>99</b>
5.1 Detailed case data	99
5.2 Research questions	100

<b>6</b>	<b>Head acceleration measurement possibilities</b>	<b>107</b>
6.1	Summary of existing solutions	107
6.2	Forthcoming applications	116
6.3	Summary regarding existing sensor systems	119
<b>7</b>	<b>System design specification</b>	<b>121</b>
7.1	Design specification	121
7.2	Accelerometers	129
7.3	Ears	133
<b>8</b>	<b>System performance</b>	<b>134</b>
8.1	First prototype evaluation – vibration tests	134
8.2	First prototype evaluation – drop tests	140
8.3	Updated sensor evaluation	165
8.4	Revised sensor evaluation	170
8.5	Second phase of shaker table trials	179
<b>9</b>	<b>Development of a low-cost alternative system</b>	<b>193</b>
9.1	Production of a system	193
9.2	Integration into ear-pieces	212
9.3	Validation of positioning	219
9.4	Results from field trials	230
<b>10</b>	<b>Discussion – Application in transport safety</b>	<b>239</b>
10.1	Usability	239
10.2	Injury prediction	241
10.3	Risk validation	242

<b>11 Summary</b>	<b>245</b>
11.1 Introduction	245
11.2 Addressing the research question	245
11.3 Contributions of the study	246
11.4 Limitations of the study	247
11.5 Recommendations for application in transport safety	248
11.6 Recommendations for future work	248
<b>12 Conclusions</b>	<b>251</b>
<b>Acknowledgements</b>	<b>253</b>
<b>References</b>	<b>255</b>
<b>Glossary of terms and abbreviations</b>	<b>275</b>
<b>Appendix A Field trial participant information</b>	<b>281</b>

## List of figures

Figure 2-1: Anterior view of skull.....	6
Figure 2-2: Lateral view of skull .....	6
Figure 2-3: Image of human brain model (as if sliced along the longitudinal fissure, sagittal mid-line).....	7
Figure 3-1: Tolerance of the human head expressed in terms of average acceleration and duration of the event.....	48
Figure 3-2: Concussion threshold relating maximum angular acceleration, $\ddot{\theta}$ , with impact duration (Sano et al., 1972).....	57
Figure 4-1: Percentage of human head impact cases where the outcome was predicted correctly by the linear or angular accelerations, or HIC .....	86
Figure 4-2: Cox regression plot for probability of avoiding a skull fracture with HIC, for human head impact cases .....	88
Figure 4-3: Head injury outcome prediction efficacy for AIS $\geq 2$ injuries .....	89
Figure 4-4: Head injury outcome prediction efficacy for AIS $\geq 3$ injuries .....	89
Figure 8-1: Image of small metal cube produced to house the accelerometer chip and align the sensitive axes to the outer planes of the cube .....	136
Figure 8-2: Accelerometer RMS values as a fraction of the reference for different input frequencies of sine wave oscillations.....	138
Figure 8-3: Moulded epoxy ear to surround ear-piece.....	142
Figure 8-4: Peak resultant acceleration values.....	148
Figure 8-5: Bland-Altman plot of peak resultant acceleration values from first prototype accelerometer ear-pieces (please note that right and left ear means lie close to one another).....	150
Figure 8-6: HIC <sub>15</sub> values .....	152
Figure 8-7: Acceleration 3 ms exceedance values .....	152

Figure 8-8: Bland-Altman plot of HIC <sub>15</sub> values from first prototype accelerometer ear-pieces .....	153
Figure 8-9: Bland-Altman plot of 3 ms exceedance values from first prototype accelerometer ear-pieces .....	154
Figure 8-10: Head acceleration results from top of head rigid impact surface test...	155
Figure 8-11: Component head accelerations from top of head impact surface test ..	156
Figure 8-12: Head acceleration results from top of head padded impact surface test .....	157
Figure 8-13: Head acceleration results from centre of gravity padded impact surface test.....	158
Figure 8-14: Head acceleration results from forehead padded impact surface test ..	158
Figure 8-15: Head acceleration results from oblique padded impact surface test ....	159
Figure 8-16: Impact point on the dummy head for the oblique tests .....	161
Figure 8-17: Head resultant acceleration results from oblique padded impact surface test.....	163
Figure 8-18: Head acceleration results from forehead padded impact surface test ..	167
Figure 8-19: Head acceleration results from forehead padded impact surface test – without time-shifting .....	167
Figure 8-20: Head acceleration results from top of head rigid concave impact surface test – 200 mm drop height – 20 kHz and 10 kHz sampling.....	168
Figure 8-21: Head acceleration results from top of head rigid concave impact surface test – 200 mm drop height– 20 kHz and 5 kHz sampling .....	168
Figure 8-22: Head acceleration results from top of head rigid concave impact surface test – 200 mm drop height– 20 kHz and 2 kHz sampling .....	169
Figure 8-23: Head acceleration results from top of head rigid concave impact surface test – 200 mm drop height– 20 kHz and 1 kHz sampling .....	169

Figure 8-24: Impact to the top of the dummy head, dropped from 2.5 m onto flat surface padded with 25 mm depth of blue Confor, Test 1 .....	173
Figure 8-25: Impact to the top of the dummy head, dropped from 2.5 m onto flat surface padded with 25 mm depth of blue Confor, Test 2 .....	173
Figure 8-26: Nominal applied acceleration for frequency response tests .....	180
Figure 8-27: Ear-pieces glued to block for x-axis frequency tests.....	181
Figure 8-28: Ear-pieces glued to block for y-axis frequency tests.....	181
Figure 8-29: Right ear-piece glued to block for z-axis frequency tests .....	181
Figure 8-30: Accelerometer mounted within rigid block (8 mm sides) but surrounded by approximately 1 mm of rubber .....	182
Figure 8-31: Frequency response for x-axis of accelerometer when mounted rigidly to the shaker table.....	183
Figure 8-32: Frequency response for y-axis of accelerometer when mounted rigidly to the shaker table.....	184
Figure 8-33: Frequency response for z-axis of accelerometer when mounted rigidly to the shaker table.....	184
Figure 8-34: Frequency response for x-axis of accelerometer with ear-pieces mounted to the shaker table .....	186
Figure 8-35: Frequency response for y-axis of accelerometer with ear-pieces mounted to the shaker table .....	187
Figure 8-36: Frequency response for z-axis of accelerometer with right ear-piece mounted to the shaker table.....	188
Figure 8-37: Frequency response for y-axis of accelerometer when mounted in silicone rubber .....	189
Figure 8-38: Frequency response for z-axis of accelerometer when mounted in silicone rubber .....	189

Figure 9-1: Shaker table test result for the x-axis; mini accelerometer output plotted with the root-mean-square (RMS) of the reference acceleration.....	196
Figure 9-2: Shaker table test result for the y-axis .....	196
Figure 9-3: Shaker table test result for the z-axis.....	197
Figure 9-4: Setup for drop tests with mini-accelerometer used in parallel with reference accelerometer. ....	198
Figure 9-5: Back-to-back outputs from the mini and reference accelerometer when dropped 1.5 m onto 50 mm of CONFOR® CF-45 foam.....	199
Figure 9-6: Peak resultant acceleration values .....	202
Figure 9-7: Bland-Altman plot of peak resultant acceleration values .....	202
Figure 9-8: Back-to-back outputs from the mini and reference accelerometer when dropped 1.5 m onto 50 mm of CONFOR® CF-45.....	203
Figure 9-9: Back-to-back outputs from the mini and reference accelerometer when dropped 1 m onto STYROFOAM.....	204
Figure 9-10: HIC <sub>15</sub> values.....	206
Figure 9-11: 3 ms exceedance values .....	206
Figure 9-12: Bland-Altman plot of HIC <sub>15</sub> values from drop tests with miniature and reference accelerometer systems.....	207
Figure 9-13: Bland-Altman plot of 3ms exceedance values from drop tests with miniature and reference accelerometer systems .....	207
Figure 9-14: Data logger, with unnecessary components removed and battery placed on top .....	209
Figure 9-15: Box housing the data logger, or acquisition unit.....	209
Figure 9-16: Image showing a miniature printed circuit board containing an accelerometer chip pressed back-to-back with a circuit board containing an angular rate sensitive chip .....	213

Figure 9-17: Cheap earplug containing sensor circuit boards fitted into the modified Hybrid III dummy 'ear' .....	213
Figure 9-18: Earplug material as provided in two containers .....	214
Figure 9-19: Earplug material ready to be removed from containers and blended together.....	214
Figure 9-20: Earplug material pressed into the ear and setting.....	215
Figure 9-21: Image of the silicone earplug during the fitting process into the right ear of the modified Hybrid III dummy including sensor units.....	215
Figure 9-22: Moulded silicone ear-piece for the modified Hybrid III dummy ear containing a miniature accelerometer and angular rate sensor, which can be seen exposed through the earplug material.....	216
Figure 9-23: A red Proguard UK mould-your-own 'Customised Personal Earplug' in blue OOMOO® cast .....	217
Figure 9-24: OOMOO® ear plug cast once the plug has been demoulded.....	217
Figure 9-25: A pair of instrumented ear-pieces ready to be used by a trial participant .....	218
Figure 9-26: The instrumented earplug component of the ear-pieces .....	218
Figure 9-27: FIA Institute in-ear accelerometer ear-piece fitted to the modified Hybrid III dummy head.....	219
Figure 9-28: Prototype sensor ear-piece fitted to the modified Hybrid III dummy head .....	220
Figure 9-29: Prototype sensor system fitted to the dummy head via a fabric headband .....	220
Figure 9-30: Prototype sensor system fitted to the dummy head via a fabric plaster .....	220
Figure 9-31: Measured accelerations from a 0.5 m drop test onto 50 mm Confor™ foam .....	223

Figure 9-32: Peak resultant acceleration values from ear-piece mounted accelerometers .....	226
Figure 9-33: Bland-Altman plot of peak resultant acceleration values from ear-piece mounted accelerometers .....	226
Figure 9-34: Measured accelerations from a 1.5 m drop test onto 50 mm Confor™ foam .....	227
Figure 9-35: Box housing the data acquisition unit as used in the trials.....	234
Figure 9-36: Connector on the cable linking the ear-pieces with the data acquisition unit.....	234

## List of tables

Table 3-1: Mild concussion risk, with and without loss of consciousness (AIS $\geq$ 1, AIS $\geq$ 2, respectively) (Somers et al., 2011) .....	46
Table 6-1: Assessment of coefficient of variation scores (Rhule et al., 2005) .....	112
Table 6-2: Rating scale to assess repeatability (Bortenschlager et al., 2007) .....	113
Table 7-1: Small accelerometer products advertised at time of writing .....	130
Table 8-1: Tri-axial accelerometer device characteristics .....	135
Table 8-2: Reference and miniature accelerometer RMS values for different input frequencies of sine wave oscillations.....	137
Table 8-3: Test matrix.....	145
Table 8-4: Peak resultant acceleration values.....	147
Table 8-5: Coefficients of variation from peak resultant acceleration values in repeated tests.....	151
Table 8-6: Top of head – padded (25 mm Confor) flat anvil tests; peak acceleration, HIC and 3 ms exceedance values .....	172
Table 9-1: Tri-axial accelerometer device characteristics .....	194
Table 9-2: Test matrix and results from drop test comparisons .....	201
Table 9-3: Coefficients of variation from peak resultant acceleration values in repeated tests.....	205
Table 9-4: Tri-axial angular rate sensitive device characteristics.....	213
Table 9-5: Peak resultant acceleration values.....	224

## Thesis publications

Part of the initial literature review and consideration of the data available from the literature was published as a TRL Published Project Report:

Carroll, J. A. (2010). Enhanced statistical analysis of head injury data, TRL Published Project Report (PPR467). Crowthorne, UK: Transport Research Laboratory (TRL).

An abbreviated version of this report was presented at the Universities' Transport Studies Group Conference in 2011:

Carroll, J. A. (2011). Enhanced statistical analysis of head injury data. Universities' Transport Studies Group Conference, January 2011. Open University, Milton Keynes, UK.

Knowledge from an exploration of the data on head injuries sustained by pedestrians in England is cited within the thesis. This work was published as:

Richards, D. & Carroll, J. (2012). Relationship between types of head injury and age of pedestrian. *Accident Analysis and Prevention*, 47 (2012), 16-23.

# 1 Introduction

*"Road traffic injuries claim more than 1.2 million lives each year and have a huge impact on health and development. They are the leading cause of death between 15 and 29 years, and cost governments approximately 3 % of GDP."* (World Health Organization, 2015)

Despite advances in road safety, head injuries still account for many of the most serious and fatal injuries in road traffic accidents and in trauma in general (Tagliaferri et al., 2006; Page et al., 2012). As an example, injuries to the head and neck are the main cause of death, severe injury and disability among motorcyclists (World Health Organization, 2015). Within Europe, head injuries constitute a high proportion of all road injuries; in particular, they represent a large part of the more severe injuries which lead to hospital admissions (EuroSafe, 2013). Traumatic brain injury is an important global public health problem (Hyder et al., 2007) and a large proportion of the severe accidental head injuries sustained by children are as a result of motor vehicle accidents (Billmire and Myers, 1985). As well as physical consequences (Hawley et al., 1999), traumatic brain injury is a major cause of disability and survivors can often suffer language, communication, social integration (Galski et al., 1998), cognitive, mood, and behavioural disorders; the societal cost of which can be substantial due to loss of years of productive life and a need for long-term or lifelong services (Critchley and Memon, 2009). In a series of patients admitted to hospital with a head injury, Lewin et al. (1979) determined that even ten years after the injury, four percent were totally disabled and 14 percent severely disabled to a degree precluding normal occupation or social life. Dikmen et al. (1995) also noted neuropsychological deficits when assessed one year post-injury. Beyond this, Harrison-Felix et al. (2004) identified that in a cohort of traumatic brain injury survivors with continued disability after rehabilitation, life expectancy was reduced by, on average, seven years compared with the general population. In their review of secondary safety priorities, Welsh et al. (2006) identified head injuries as being the most costly for vehicle occupants, when minor injuries were excluded.

In 1932, within his paper regarding cerebral involvement in head injury, Russell commented that,

*"From the neurological point of view, however, the clinical study of head injury has received relatively little attention, and offers a wide field for investigation."* (Russell, 1932)

Since then, a wealth of research has been carried out in an effort to determine the principal mechanisms responsible for head injuries and the principal parameters that should be considered in the development of appropriate head injury criteria. At a superficial level it would appear that the efforts to understand the mechanisms of head injuries have been extensive. However, despite the long term investment in head injury research there are still conflicting notions, even at a fundamental level, regarding the mechanisms that govern or influence head injuries and head injury risk.

A possible explanation for this incoherence is the involvement of confounding factors that appear to have been explored insufficiently in prior research work. Therefore, as a start to this study, it was decided to investigate the previous head injury research in greater detail. The objective was to identify the confounding factors from previous research that contribute to conflicting notions regarding head injury mechanisms and inhibit the development of advanced head injury criteria. To begin this process the literature was reviewed to establish the current position in head injury research with regard to:

- Previous assertions as to head injury mechanisms
- Methods of describing head motion using accelerations
- Existing head injury criteria

Beyond this it was considered to be important to understand the data that are available to identify and explore potential confounding factors and that could also be used to propose or validate a new injury criterion or criteria.

## **2 Background**

The following four subsections provide some background to the common ways in which the head, head injuries and head motions leading to injury are described. Whilst the face is a part of the head, it is typical for the 'head' to exclude the face. Therefore this assumption has been followed throughout the thesis unless otherwise stated.

### **2.1 Kinematic theories of head injury**

Newman (1998) provides a concise description of the principles involved when describing the motion behind head injury.

*"The most popular theories of the biomechanics of head injury ascribe movement of the head or some part of it, as the means by which injury occurs. A scalp laceration is the result of a mechanical action (cutting or tearing) that separates formerly contiguous pieces of scalp. A skull fracture will occur when the skull bone bends more than it is capable of doing without breaking. A brain contusion for example, is a collection of blood caused by the rupture of blood vessels that have been stretched too much. Separating, bending, and stretching are merely descriptors of somewhat different kinds of movement. Brain injury can occur if any part of it is distorted, stretched, compressed, or torn away from the interior of the skull. An impact to the head can cause the skull to deform and, even if it does not fracture, the underlying brain tissue can be injured as it distorts under the influence of the deforming skull. Even if the skull does not bend, if it is caused to move violently, distortion within the brain will occur. That is, injury is caused by movement; movement that can be described by the laws of motion. The notion that brain injury is associated with acceleration/deceleration is conceptually rather attractive and not incorrect. It must be remembered though that acceleration is a kinematic response to some forcing function (either applied or generated by impact) and injury is a physiological response.*

*Both are results of something. One does not directly cause the other. Whether or not these two responses or outputs can correlate to each other has been a major underlying focus of brain injury biomechanics research.”*

Head impact is a mechanism through which brain injury can occur; however, it has been demonstrated that even without a direct impact, inertial loading of the head can bring about brain injury (Ommaya et al., 1967). Despite this finding, it has never been confirmed that it is possible to concuss a primate subject by applying acceleration to the body and not directly to the head.

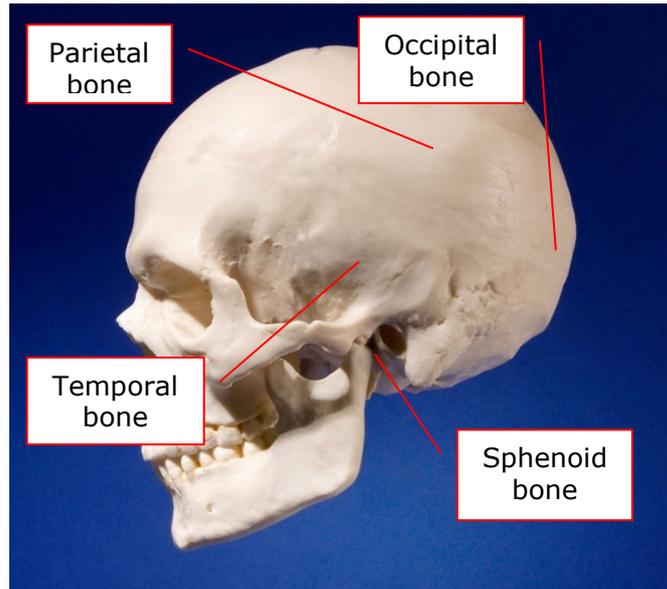
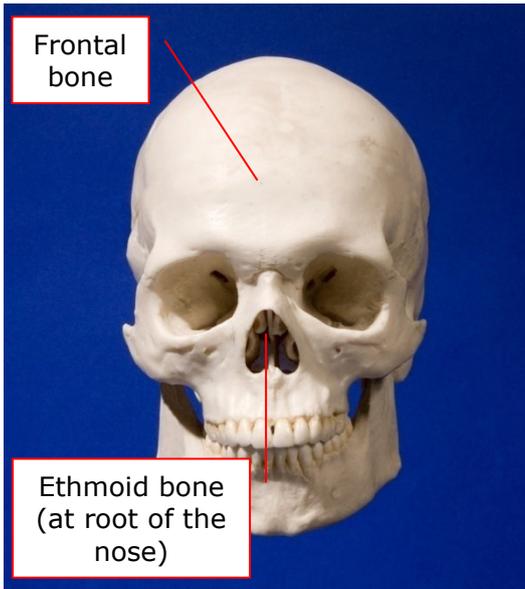
Whilst some form of head contact may occur, the absence of significant skull deformation is a typical expectation for an occupant in a modern vehicle during an accident. In such cases, the forces that act to bring the head to rest are induced through the neck of the occupant and the restraint system (e.g. airbag). Without significant skull deformation, the skull can be considered as a container for the brain. It is the movement of the skull with respect to the brain that leads to disruption of the brain tissue and associated injury.

We may consider that before a road traffic accident, the persons involved are travelling at a constant velocity and therefore the acceleration acting on the head is zero (excluding the continuous action due to gravity). After the collision event, the acceleration will also be zero. However, during the event itself the acceleration of the person, including the head, will go from zero to some maximum acceleration (or maxima) then back to zero. For impacts of a similar type, it follows that a greater change in velocity, from before to after the event, or a shorter event period will generate a higher acceleration. The question then arises as to the causal link between the forces acting on the head, the acceleration and head motion and the injurious potential of that event. Whatever the causal link, linear and rotational kinematics allow us to describe head motion during an event.

## 2.2 Description of structures within the head

A detailed description of the head and its constituent structures is given by Carroll (2010). A brief description of the fundamental elements within the head is reproduced here:

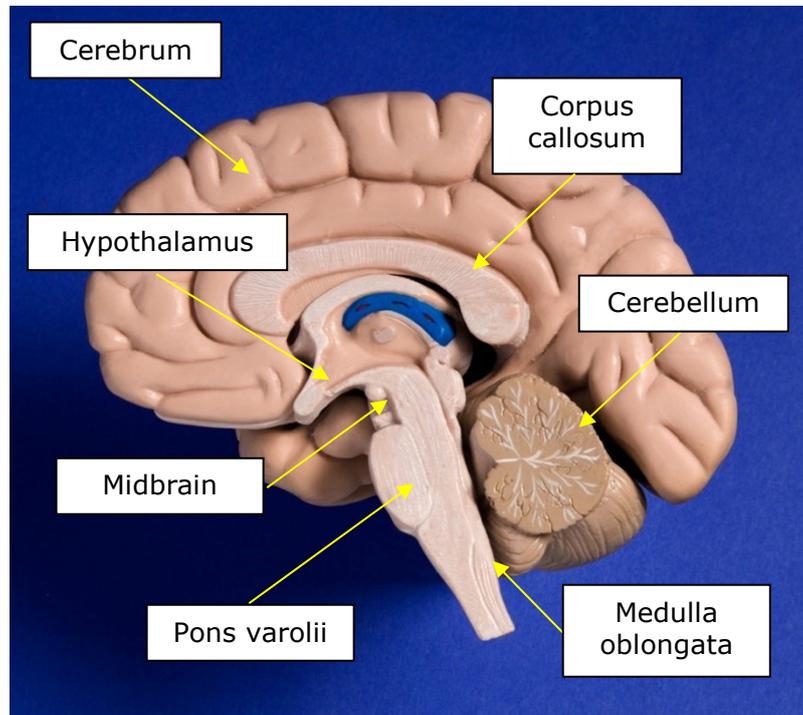
- Scalp
  - The outermost constituent part of the head is the scalp; the layer of 'skin' surrounding the skull.
  - *"The skin is a highly organized, stratified structure consisting of three main layers, called the epidermis, dermis and hypodermis."* (Hendriks, 2001)
  - *"Human skin is a non-homogeneous, anisotropic, non-linear viscoelastic material which is subjected to a prestress in vivo. Its properties vary with age, from site to site and per person."* (Hendriks, 2001)
  
- Skull
  - The skull is a network of bones formed around the brain, eyes, ears, nose, and teeth. Eight bones make up the brain case, with another 14 bones forming the face (excluding the teeth).
  - The eight bones comprising the cranium are the: ethmoid, sphenoid, frontal, two temporal, two parietal, and occipital bones (see Figures 2-1 and 2-2). The vault is the upper part of the cranium and the base of the skull is the lowest part of the cranium.
  - For the most part, the inner surface of the cranial cavity is smooth; however, at its base it contains depressions and ridges for supporting the different regions of the brain plus small holes (foramen) for arteries, veins, and nerves. The foramen magnum is the large hole through which the spinal cord passes as it transitions into the brain stem (medulla oblongata).



**Figure 2-1: Anterior view of skull**

**Figure 2-2: Lateral view of skull**

- Brain
  - The brain is the portion of the central nervous system that is contained within the skull (cranial cavity). Four principal parts comprise the brain; these are the cerebrum, cerebellum, pons varolii, and medulla oblongata (see Figure 2-3):



**Figure 2-3: Image of human brain model  
(as if sliced along the longitudinal fissure, sagittal<sup>1</sup> mid-line)**

- Cerebrum
  - This is the largest portion of the brain mass, and occupies a considerable part of the cranial cavity.
  - In general the cerebrum function is associated with thought and control of actions.
  - Its upper surface is an ovoid shape, broader behind than in front, convex in its general outline, and divided into two halves or hemispheres right and left, by the great longitudinal fissure.
  - The outer surface of the brain is often subdivided according to the bone under which each of four lobes lies.
  - The hemispheres are composed of a covering of grey matter, called the cerebral cortex and a central mass of white matter.

---

<sup>1</sup> The three orthogonal planes commonly used in anatomical descriptions are shown in the Glossary

- Beneath the cortex at the longitudinal fissure is the corpus callosum. This is a wide, flat bundle of neural fibres about 10 cm long. It connects the left and right hemispheres and is the largest white matter structure in the brain.
- Below the corpus callosum is the thalamus and hypothalamus. The hypothalamus is important from a functional perspective as it links the nervous system to the endocrine system, controlling hormones and metabolism.
- Midbrain
  - Between the forebrain (the cerebral cortex) and the hindbrain (including the cerebellum, pons varolii and medulla oblongata) lies the 'midbrain'. The midbrain connects these fore and hind portions. It serves important functions in motor movement, particularly movements of the eye, and in processing sensory inputs, for example auditory and visual information.
- Cerebellum
  - Situated on the inferior occipital fossa. It is separated from the cerebrum by the tentorium cerebella. Connection to the rest of the brain is via connecting bands, 'crura'. Of these, two ascend to the cerebrum, two descend to the medulla oblongata, and two blend together in front forming the pons varolii.
  - The function of the cerebellum is coordination of voluntary movements such as posture, balance, coordination, and speech.
- Pons varolii
  - This is the section of the brain that rests upon the upper part of the basilar process. It forms a sort of centre to the cerebrum and cerebellum receiving crura from both as well as being connected with the medulla oblongata below.

- The pons variolii has been linked with functions critical to life, such as control of breathing (depth and frequency). It has been linked to sleep cycle behaviour and also contains the origins of several nerves such as those for chewing and swallowing and for facial expressions.
- Medulla Oblongata
  - This portion of the brain extends from the lower border of the pons varolii to the upper part of the spinal cord. It lies beneath the cerebellum, resting on the lower part of the basilar groove of the occipital bone.
  - The medulla oblongata controls the involuntary functions; regulating respiration, heart rate and digestion.
- The membranes of the brain (also known as the meninges) surround the brain and central nervous system, descending from the cerebrum to encompass the spine down to the sacrum. The cerebrospinal fluid circulates between the meninges, in particular in the subarachnoid space between the arachnoid and pia maters. The three meninges are the:
  - Dura mater
    - The dura mater is a thick and dense inelastic fibrous membrane, which lines the interior of the skull. Its outer surface is rough and fibrillated and adheres closely to the inner surface of the skull bones. Its inner surface is smooth and epithelial<sup>2</sup>, being lined by the parietal<sup>3</sup> layer of the arachnoid. Bridging arteries and veins between the dura mater and the skull bones are numerous.

---

<sup>2</sup> Membranous (like a skin)

<sup>3</sup> Forming a wall

- Arachnoid membrane (like a spider's web, due to its extreme thinness)
    - This is the serous membrane which envelops the brain and is then reflected on the inner surface of the dura mater. It can be considered as a closed sack having a parietal and visceral layer.
  - Pia mater
    - A vascular membrane consisting of a minute plexus<sup>4</sup> of blood-vessels, surrounded by an extremely fine tissue. It invests the entire surface of the brain, dipping down between the convolutions and laminae.
- Ear
    - Three parts comprise the ear; the outer, middle and inner ear
    - The pinna or ear flaps form the outer ear together with the ear canal (the tube along which sound waves are transmitted from the outer to inner ear):
      - The ear canal or the external auditory canal consists of an outer cartilaginous portion, in which the first and second ear canal bends are located, and an inner bony portion that runs through the temporal bone and terminates at the tympanic membrane or ear drum. It is not a uniform tube and variations in the ear canal diameter can become significant with changes in ear canal cross-section or mechanical properties (Chen et al., 2010).

---

<sup>4</sup> A network of interlacing nerves or connecting/reconnecting blood vessels or lymphatics

- Little evidence was found regarding the material properties of the constitutive tissue of the ear and ear canal. Perhaps related to the ear; Rotter et al. (2002) found the equilibrium modulus of nasal cartilage, under compression, to be 234 kPa. Otherwise, the only direct information was that quasi-static tensile tests of human auricular specimens by Park et al. (2004) showed that the cartilage had an Ultimate Tensile Strength of 2.18 MPa and stiffness of 5.11 MPa.
- The middle ear consists of:
  - The inner end of the ear canal
  - The ear drum which is a thin layer of skin or membrane at the inner end of the ear canal
  - Beyond this the malleus (hammer) transmits vibrations from the drum to the incus (anvil), which vibrates the stapes (stirrup), which in turn passes vibrations onto the inner ear and the cochlea.
- In the inner ear is the cochlea and the associated semi-circular canals. These are filled with fluid and tiny hairs which are sensitive to the vibrations from the middle ear and are also responsible for balance.

As mentioned earlier, the face could be included in a description of the head, but for the purposes of this study, the head is taken to mean the cranium and its contents.

As presented, anatomical vocabulary provides a means to accurately describe parts of the head influenced and potentially injured during an event.

## 2.3 Head injury

As a result of considering the head anatomy and common injuries it was clear that several physical parameters could affect the expected forms of head injury or the tolerance of the head to that injury. These parameters are listed in a later section of this report, as confounding factors. Some examples of major factors are given below, for instance:

- Contact area
  - Broad or narrow
    - A narrow contact may be more likely than a broad contact to cause penetrating injuries.
- Stiffness of impacting object
  - Rigid or padded / deformable
    - Padding has been shown to have a protective effect against skull fracture and through absorbing the energy of an impact can reduce head accelerations for a given impact severity.

Based on the geometry and varying intracranial contents, the direction of impact is also expected to have a bearing on the injury outcome:

- Frontal
- Lateral
- Occipital
- Crown

### **2.3.1 Types of head injury**

Head lesions that develop at the time of the injury are called 'primary'; whereas, many develop over a period of hours to days after the triggering event and are called 'secondary' injuries (Hilton, 2009). Hilton states that a significant minority of patients with severe head injury develop progressive neurological deterioration several years later.

*"Primary traumatic brain damage is the result of mechanical forces producing tissue deformation at the moment of injury. These deformations may directly damage the blood vessels, axons, neurons and glia<sup>5</sup> in a focal, multifocal or diffuse pattern of involvement..."*  
(Blumbergs, 2005).

*"Secondary traumatic brain damage occurs as a complication of the different types of primary brain damage and includes ischemic and hypoxic damage, cerebral swelling, the consequences of raised intracranial pressure, hydrocephalus<sup>6</sup> and infection"* (also Blumbergs, 2005).

*"No specific pharmacological therapy is currently available that prevents the development of secondary brain injuries, and most therapeutic strategies have failed in translation from 'bench to bedside'"*  
(Beauchamp et al., 2008)

The precise classification of head injury forms can be subdivided to varying degrees depending on the author describing the injury. Descriptions of common head injuries are provided below. The injuries are grouped into four categories, skull, focal, diffuse, and external soft tissue injuries. The different types of injury comprising these categories are described within the group.

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<sup>5</sup> Glia, or the neuroglia, are the fibrous and cellular non-nervous supporting elements of the nervous system.

<sup>6</sup> Also called hydrocephaly, an increase in the volume of cerebrospinal fluid within the skull. *"The term is commonly applied to distensions of the ventricular system by cerebrospinal fluid which cannot escape into the subarachnoid space, is blocked in the subarachnoid pathways, or cannot be absorbed into the venous system."* (Hoerr and Osol, 1960)

- Skull injuries – two particular, and common, types of skull fracture are described below (linear and depressed). These fracture types can occur in any region of the skull; however, fractures to the main vault of the cranium or to the basilar region may have different consequences.

- Linear

- In a linear skull fracture, skull penetration does not occur.
- Gurdjian et al. (1949) reported that, "*linear fractures are, in general, initiated on the external surface of the skull due to outbending at a considerable distance from the point of impact. After initiation the fracture line runs toward the point of impact and also extends in the opposite direction.*"
- As a result linear skull fracture may not have much significance on the course of brain injury. However, it may be possible that if a cranial fracture results in the rupture of an underlying blood vessel, that the ensuing haematoma may exert pressure on the brain and lead to some brain injury.
- Also, skull fracture, of any kind, is indicative of a significant force having been applied to the head.
- Linear fractures occurring together may leave multiple fragments in which case the fracture is 'comminuted'.

- Depressed

- Similar to linear skull fractures, but tending to come about as a result of a more concentrated impact loading. This leads to more focussed contact effects, resulting in skull penetration. When occurring to the cranium, the skull fragments that are driven into the cranial cavity can lead to trauma to the underlying brain and blood vessels.

- Vault fracture
  - Fractures to the vault can cause meningeal and cortical injury when fragments of the fractured bone enter the cranial cavity.
- Basilar fracture
  - This part of the skull contains passages for the blood vessels, providing supply to the entire brain, and to passages for the neurological connections between the brain and the rest of the body.
- Focal injuries – *"Mechanical deformation due to compression, tension and shear can cause tearing of blood vessel walls and hemorrhage into the surrounding tissue provided there is sufficient moving blood in the circulatory system... The amount of the hemorrhage into the neural tissue depends on a number of factors including the nature of the blood vessel(s) damaged (i.e. whether capillary, venule, vein, arteriole, small or large artery) and systemic factors such as body temperature (hyper and hypothermia), shock associated with hypoxia, coagulation factor changes, blood pressure, age, acute alcohol intoxication, the effects of medications or illegal drugs, effects of accompanying injuries (multiple trauma) and prior or associated diseases such as arteriosclerosis."* (Blumbergs, 2005)
  - Epidural (or extradural) haematoma
    - Blood accumulation between the inner surface of the skull and the dura mater. It occurs as a result of trauma to the skull and the underlying meningeal vessels and not due to brain injury. The dura is adherent to the inner aspect of the cranial bones, particularly at the sutures and at the base of the skull, and contains several blood vessels. This injury is usually found in conjunction with skull fracture, but not always.

- Blood may accumulate over a period of hours (with an initial lucid interval for the patient) and the volume of the haematoma can be used as a predictor of outcome.
- Subdural haematoma
  - An extravasation<sup>7</sup> of blood located between the dura mater and the arachnoidal membrane.
  - Typically three sources – direct lacerations of cortical veins and arteries by penetrating injury, a large contusion bleeding into the subdural space, and tearing of veins that bridge the subdural space.
  - In association with rapid acceleration, subdural haemorrhages do not require direct impact (Hilton, 2009).
  - *"Subdural haemorrhage may present shortly after the head injury (acute subdural haemorrhage), 1-2 weeks later (subacute haemorrhage) or more than 2 weeks later (chronic subdural haemorrhage). Chronic subdural haematomas are particularly common in the elderly, alcoholics and patients with a low intracranial pressure,..."* (Hilton, 2009)
  - *"Subdural haematomas result from displacement of the brain relative to the dura sufficient to cause rupture of the bridging veins which course from the brain's surface to the overlying venous sinuses. Thus, unlike epidural haematomas which occur from focal impact injuries, subdural haematomas almost always result from angular deceleration of the head in which the brain continues to rotate relative to the more stationary skull and dura. This explains the high incidence of subdural haematoma seen in adults involved in motor vehicle accidents in which the head rotates around an axis in the lower cervical spine, often*

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<sup>7</sup> To pass bodily fluid out of its proper place

*decelerating abruptly as it impacts against a surface.*" (Duhaime et al., 1992)

- Subarachnoid haemorrhage
  - Blood extravasation in the subarachnoidal space, external to the pia mater.
  - *"Small collections of subarachnoid blood are fairly common after head injury, particularly in association with contusions and lacerations."* (Hilton, 2009). Blumbergs (2005) describes traumatic subarachnoid haemorrhage as being, *"the most common abnormality seen in head injury, although in most cases it is minor and of little clinical significance."*
  - Translational forces from falls or focal impact from a heavy moving object may be associated with focal contusions and usually localised subarachnoid haemorrhage. *"Such forces may also be generated when a child acts as a missile in a motor vehicle or pedestrian accident if the head and brain move in a line rather than rotate when they decelerate. These children often have a relatively benign clinical course even when fractures and focal contusions are extensive. Contusional brain swelling and focal neurologic deficit may complicate recovery or increase mortality when large forces are involved."* (Duhaime et al., 1992)
- Contusion
  - Essentially cerebral contusions, either coup-contusions or contrecoup-contusions (same or opposite side to the impact site); although intermediate contusions (intracerebral, between the impact and non-impact sides), fracture contusions (beneath the site of a fracture), and gliding contusions (from brain movement) are also documented.

- Consists of areas of necrosis, pulping, infarction, haemorrhage, and oedema.
- Adams et al. (1980) reported that contusions predominantly occur to the frontal and temporal regions which are impacted against the irregular floor of the skull at the frontal and middle fossae. However, the study supporting this statement considered a multitude of different impact types, so the regions of the brain injured most frequently could be related to the most frequent types of head impact.
- Contusions are relatively uncommon in young infants where the floor of the skull has a smoother contour (Hilton, 2009).
- Although contusions may be asymptomatic, they can be a cause of long-term epilepsy (Hilton, 2009).
- Being associated with acute subdural haematoma, this type of lesion may represent a serious life-threat (Musigazi et al., 2015).
- Neurological deficits usually correlate with the size and location of contusions (Blumbergs, 2005):
  1. Anterior temporal lobe contusions are often associated with delirium, disinclination to be examined or moved, expletive speech and resistance if disturbed.
  2. Inferior and fronto-polar frontal contusions may be associated with a quiet, disinterested, slowed mental state (abulia) with dull facial appearance, lying quietly with eyes closed when undisturbed.
  3. Medial temporal lobe contusions may be associated with anterograde and retrograde memory loss.

4. Convexity contusion – focal deficit such as aphasia<sup>8</sup> or hemiparesis<sup>9</sup>.
  5. Medial frontal contusions – confusion with inattention, poor performance on simple mental tasks, fluctuating or erroneous orientation.
- Intracerebral haematoma
    - Resulting from the rupture of a small blood vessel (capillary, vein, and/or artery) within the brain.
    - Haemorrhages tend to begin superficially, but extend deeply into the white matter of the brain.
    - The damage of the blood vessels begins a train of events from haemorrhage to breakdown of the blood-brain barrier and infarction. The severity of the changes can vary from focally dilated blood vessels to burst brain lobes (Blumbergs, 2005).
  - Intraventricular haemorrhage
    - *"A small amount of intraventricular hemorrhage (IVH) is frequently found in head injured patients that do not survive long enough to reach hospital."* (Blumbergs, 2005)
  - Following vascular injury, blood flow to the cerebrum can be altered. This can lead to reduced or absent perfusion of brain tissue and consequent ischemia if the blood circulation is inadequate. When limited to the region of the brain supplied by the affected vessel, the ischemia can be classed as focal, or regional. However, global ischemia (non-perfused brain) occurs when cerebral perfusion pressure drops below 6 kPa (Blumbergs, 2005). This can result in selective neural necrosis to pan-necrosis involving neurons as well as other cellular components of neural tissue.

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<sup>8</sup> Loss or impairment of the capacity to use words as symbols of ideas

<sup>9</sup> Paresis (slight paralysis, loss of muscular power) of one side of the body

- Diffuse brain injuries – consequences of diffuse damage can cause injuries that vary from concussion, without apparent neurological sequelae, to prolonged traumatic coma; with long-term, usually irreversible, neurological outcome commonly found in patients with severe head injuries.
  - Mild concussion
    - Does not involve loss of consciousness; however, confusion, disorientation, and posttraumatic / retrograde amnesia may be present.
    - Gennarelli recently proposed a symptomcentric concept of concussions in which, *““concussion” can occur and symptoms be generated, not only by dysfunction of the axon and other parts of the neuron (soma<sup>10</sup>, mitochondria<sup>11</sup>, dendrites<sup>12</sup>, synaptic networks, etc.) but also by mechanically induced dysfunction of vascular (causing symptoms due to vasoconstriction or vasodilatation of arteries, vein, capillaries), oligodendrocytic<sup>13</sup> (symptoms from demyelination or altered electrical conduction), astrocytic<sup>14</sup> (symptoms from gliosis) or microglial<sup>15</sup> (symptoms from inflammatory processes) components in single or multiple portions of the brain, not just the cerebrum.”* (Gennarelli, 2015)
  - Classical cerebral concussion
    - Involves immediate loss of consciousness. Clinically, the loss of consciousness should be less than 24 hours and be reversible (without detectable pathology).

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<sup>10</sup> The cell body of a neuron

<sup>11</sup> An organelle found in cells which is responsible for respiration and energy production

<sup>12</sup> Extension of a nerve cell along which the impulses are transmitted

<sup>13</sup> Relating to the oligodendrocytes which provide the myelin sheath for axons

<sup>14</sup> Relating to astrocytes which are star-shaped glial cells involved in support around the blood-brain barrier, provision of nutrients, maintenance of extracellular ion balance, etc.

<sup>15</sup> Microglia are a type of glial cell forming the active immune defence in the central nervous system

- Diffuse brain injury
  - At the transition between physiological dysfunction and anatomical disruption of the brain is immediate loss of consciousness lasting for over 24 hours. Often called diffuse brain injury in the absence of a more specific title. This can involve decerebrate posturing<sup>16</sup>, amnesia lasting for days, mild to moderate memory loss, and mild motor deficits.
  - *"Diffuse vascular injury – some patients who die immediately following a severe acceleration or deceleration type of brain injury have widespread petechial<sup>17</sup> haemorrhage throughout the brain due to shearing forces being exerted upon blood vessels. These patients do not survive long enough to develop any axonal changes."* (Hilton, 2009)
  
- Diffuse axonal injury
  - Immediate loss of consciousness lasting for days to weeks, decerebrate posturing, severe memory and motor deficits, and posttraumatic amnesia may last for weeks.
  - *"Brain injury characterised by prolonged traumatic coma not due to mass lesions that has dysfunction or structural damage to brain axons."* (Gennarelli et al., 1987)
  - *"Cerebral concussion and diffuse axonal injury appear to the less and more severe ends of a continuous spectrum of brain dysfunction characterised by increasing amounts and distribution of axonal damage throughout the brain and brainstem."* (Gennarelli et al., 1987)

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<sup>16</sup> An abnormal body posture that involves rigid extension of the arms and legs, downward pointing of the toes, and backward arching of the head

<sup>17</sup> A minute rounded spot of haemorrhage on a cross-sectional surface of an organ

- *"DAI [diffuse axonal injury] has been divided into three grades of severity on the basis of the combination of macroscopic and microscopic marker lesions. In grade 1 DAI widespread axonal damage is present in the corpus callosum, white matter of the cerebral hemispheres and brainstem. In grade 2 DAI, there are, additional focal abnormalities (usually small hemorrhages) in the corpus callosum and in grade 3 DAI there are, in addition to the findings of grade 2, small hemorrhages in the rostral<sup>18</sup> brainstem."* (Blumbergs, 2005; quoting Adams et al., 1989)
  - Brain swelling
    - Brain swelling, or an increase in intravascular blood within the brain, may occur alongside a diffuse brain injury. This contributes to secondary damage through there being increased intracranial pressure compromising cerebral perfusion leading to secondary ischemia. Additionally, deformation through herniation can have devastating consequences (Kochanek et al., 2007).
    - Brain swelling and oedema (increase in tissue fluid) are not the same but are often used almost interchangeably. Cerebral oedema is an increased water content of the brain. Congestive brain swelling is increased intravascular blood volume due to arterial dilation and/or venous obstruction. Brain swelling can be as a result of either one of these (cerebral oedema or congestive brain swelling), or a combination of the two.
  - Other diffuse pathophysiological changes
    - Glutamate release, which via the glutamate receptors activates other intracellular cascades and neuronal cell death (Beauchamp, 2008).

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<sup>18</sup> The top, or head, of the brainstem (i.e. towards the cerebrum, as opposed to the spinal cord)

- Elevation of cation concentration through elevation of intracellular calcium levels, contributing to cellular injury or death (Kochanek et al., 2007).
  - Balance of inflammation through pro- and anti-inflammatory mediators, as involved in the repair of injured tissue, but which are neurotoxic if they pass through the blood-brain barrier (Hill et al., 2016).
  - Release of reactive oxygen species and free radicals potentially leading to damage to cell membrane integrity, protein dysfunction and DNA damage (Hill et al., 2016).
  - Apoptosis (the process of programmed cell death) brought about via DNA damage, free radical interactions and mitochondrial damage (Büki and Povlishock, 2006).
- External soft tissue injuries
    - Bruise
      - A bruise occurs with blunt trauma of sufficient magnitude to cause the small blood vessels beneath the skin to extravasate blood into the surrounding tissue under the intact skin.
    - Abrasion
      - An abrasion is caused by a blunt object sliding over a body area with sufficient force to denude the superficial layers of the skin.
    - Laceration
      - A puncture wound or a longer incised wound. A puncture wound occurs when a sharp object applies enough force to the skin to penetrate it. When a sliding force is added to the penetration by a sharp object, a tearing or slicing produces a long opening in the skin.

- Avulsion
  - An avulsion is caused by a sharp object penetrating the skin with application of its force over a broad plane so that a peeling action occurs.
- Burn
  - A burn occurs when enough thermal energy is applied to the area to cause breakdown of the tissue.
- These external soft tissue injuries of the head are quite common and may result in severe haemorrhage, infection, or disfiguration. Even so, they usually cause much less morbidity than do skull and brain injuries (Douglass et al., 1968).

### **2.3.2 Head injury coding**

There are a few different strategies which can be used to classify head and brain injuries. Some of the more commonly used rating systems are shown in the following sections.

#### *ICD-10*

Chapter XIX of the International Classification of Diseases (ICD) covers: "Injury, poisoning and certain other consequences of external causes" (WHO, 2007). In particular the Block of injuries S00 to S09 relate to injuries to the head, the titles of the injury groups are given below. The S02 and S06 injuries have been expanded to show the different subcategories for fractures of the skull and intracranial injuries.

- S00 Superficial injury of head
- S01 Open wound of head
- S02 Fracture of skull and facial bones

The following subdivisions are provided for optional use in a supplementary character position where it is not possible or not desired to use multiple coding to identify fracture and open wound; a fracture not indicated as closed or open should be classified as closed.

0 = closed

1 = open

S02.0 Fracture of vault of skull

Frontal bone

Parietal bone

S02.1 Fracture of base of skull

Fossa: anterior, middle, or posterior

Occiput

Orbital roof

Sinus: ethmoid, frontal

Sphenoid

Temporal bone

Excludes: orbit NOS (S02.8), orbital floor (S02.3)

S02.2 Fracture of nasal bones

S02.3 Fracture of orbital floor

Excludes: orbit NOS (S02.8), orbital roof (S02.1)

S02.4 Fracture of malar and maxillary bones

Superior maxilla

Upper jaw (bone)

Zygoma

- S02.5 Fracture of tooth
    - Broken tooth
  - S02.6 Fracture of mandible
    - Lower jaw (bone)
  - S02.7 Multiple fractures involving skull and facial bones
  - S02.8 Fractures of other skull and facial bones
    - Alveolus
    - Orbit NOS
    - Palate
    - Excludes: orbital: floor (S02.3), roof (S02.1)
  - S02.9 Fracture of skull and facial bones, part unspecified
  - S03 Dislocation, sprain and strain of joints and ligaments of head
  - S04 Injury of cranial nerves
  - S05 Injury of eye and orbit
  - S06 Intracranial injury
    - S06.0 Concussion
    - S06.1 Traumatic cerebral oedema
    - S06.2 Diffuse brain injury
    - S06.3 Focal brain injury
    - S06.4 Epidural haemorrhage
    - S06.5 Traumatic subdural haemorrhage
    - S06.6 Traumatic subarachnoid haemorrhage
    - S06.7 Intracranial injury with prolonged coma
    - S06.8 Other intracranial injuries
    - S06.9 Intracranial injury, unspecified
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- S07 Crushing injury of head
- S08 Traumatic amputation of part of head
- S09 Other and unspecified injuries of head

It is worth noting that only the frontal and parietal bones constitute the 'vault' according to the ICD-10 coding. This reflects the division of the cranium where the vault is the upper section only.

### *Glasgow Coma Scale*

The Glasgow Coma Scale (GCS; <http://glasgowcomascale.org/>) is a 15 point scale for estimating and categorising the outcomes of brain injury on the basis of overall social capability or dependence on others.

The test measures the motor response, verbal response and eye opening response with these values:

#### I. Motor Response

- 6 - Obeys commands fully
- 5 - Localise noxious (painful) stimuli
- 4 - Normal flexion, withdraws from noxious stimuli
- 3 - Abnormal flexion, i.e. decorticate posturing
- 2 - Extensor response, i.e. decerebrate posturing
- 1 - No response

#### II. Verbal Response

- 5 - Alert and Oriented
- 4 - Confused, yet coherent, speech
- 3 - Words
- 2 - Sounds
- 1 - No sounds

### III. Eye Opening

4 - Spontaneous eye opening

3 - Eyes open to sound

2 - Eyes open to pressure

1 - No eye opening

The final score is determined by adding the values of I+II+III.

This number helps medical practitioners categorise the four possible levels for survival, with a lower number indicating a more severe injury and a poorer prognosis:

Mild (13-15):

Moderate Disability (9-12):

Loss of consciousness greater than 30 minutes

Physical or cognitive impairments which may or may resolve

Benefit from Rehabilitation

Severe Disability (3-8):

Coma: unconscious state. No meaningful response, no voluntary activities

Vegetative State:

Sleep wake cycles

Arousal, but no interaction with environment

No localised response to pain

Persistent Vegetative State:

Vegetative state lasting longer than one month

Brain Death:

No brain function

Specific criteria needed for making this diagnosis

A problem with the GCS is that the classification is based on a single assessment. Where this assessment may vary with time, the point of the assessment then becomes critical. Other complications can also produce issues; for instance, swollen eyes may affect the eye-opening response, and drug or alcohol use or intubation may affect verbal response.

#### *Glasgow Outcome Scale and Glasgow Outcome Scale Extended*

The Glasgow Outcome Scale (GOS) (Jennet & Bond, 1975) has become the most widely used scale for assessing outcome after head injury and nontraumatic acute brain insults. The eight-point extended Glasgow Outcome Scale (GOSE), develops the proposal of Jennett et al. (1981) by providing various criteria to subdivide the upper three categories of the scale into 'better' or 'worse' – "lower" or "upper". To accommodate this it adds specific sub-components for independence in the home, work and return to normal life (Wilson et al., 1998). Traditionally outcome on the GOS has been assigned after a short interview, usually unstructured, and not involving a written protocol. To address the limitations associated with this approach, Wilson et al. proposed a standard format for the interview and identifying specific criteria for assigning an outcome category. The questionnaires used to obtain the GOS and GOSE should therefore be identical apart from the inclusion of the additional items in the GOSE.

*"The advantages of the GOS remain its simplicity, wide recognition, and the fact that differences in disability are clinically meaningful. Provided that the purpose and limits, as well as the benefits, of the GOS are appreciated, it can continue to have a central place in the assessment of head injury outcome." (Wilson et al., 1998)*

### *Abbreviated Injury Scale*

The Abbreviated Injury Scale (AIS) can be used to score each injury a person may sustain on a six point scale, rating the threat to life associated with that injury. The six levels range from 1 (minor) to 6 (maximum – currently untreatable or incompatible with life). In its latest revision (AAAM, 2008), the AIS contains descriptions of almost 300 potential injuries to the head.

### *Functional assessment measure*

*"The functional independence measure (FIM) scores 18 functional activities on a seven level scale. In view of the prominence of communicative, cognitive, and behavioural disturbances after brain injury a further 12 items considering those issues were added to the FIM to construct the functional assessment measure. It has become accepted custom to use the abbreviation FIM+FAM for the complete 30 item functional assessment measure."* (Hawley et al., 1999)

### *Short Form 36 (SF-36)*

*"The SF-36 is a multi-purpose, short-form health survey with only 36 questions. It yields an 8-scale profile of functional health and well-being scores as well as psychometrically-based physical and mental health summary measures and a preference-based health utility index. It is a generic measure, as opposed to one that targets a specific age, disease, or treatment group."* (Ware, 2004)

Understanding of the terms presented in the injury and injury coding sections above enables precise descriptions to be given of the type of head injury sustained during an event.

## **2.4 Motion described through acceleration**

There has been much discussion in the literature regarding the importance of head acceleration to describe the motion of the head and provide a link to the mechanism of injury. Conventionally, acceleration modes are described either in terms of a rotation or translation of the head. This division has been adopted in the head injury literature and has become something of a focal point for much discussion as to which is the dominant measure: linear (translational) or angular (rotational) acceleration.

It seems that one key difference between translational and rotational motion for the head is that rotation implies varying tangential acceleration along the length of the radius. In pure translational motion all parts of the head will accelerate with the same magnitude (if a freely moving, solid body). This is likely to be of great importance when considering the occurrence of intracerebral injuries. Intrinsically rotational motion of the head (if assumed to be solid/homogeneous) will tend to induce more shear distortion in the brain than pure translation (which without edge effects would induce no shearing within the bulk of the head).

As we know that the head is not a solid body or homogeneous (and doesn't behave as such; Sabet et al., 2008), it becomes important to consider both translational and rotational accelerations. It is likely that the combination of both accelerations will be important in predicting the risk of a brain injury occurring from a given loading event.

## **2.5 Summary**

There are a variety of head injuries that can occur in general and in the transport setting. They can affect different structures within the head and can be of very different severity. Some can have immediate and severe consequences whilst others might be initially minor in nature but lead to long-lasting sequelae, and other might be minor in nature without the expectation of any secondary complications. Existing information, terminology and coding systems provide us with the necessary language to describe head injuries; the anatomical structures they affect and the loading to the head which caused them. This is important for documenting cases of head injury in sufficient detail to be of use in subsequent research. It also provides the opportunity whereby proposed mechanisms of injury can be discussed and exchanged between researchers and other interested parties. By complying with these conventional ways of describing head injuries there is the potential for research to be disseminated and used across various disciplines and research areas. It also makes head injury prediction techniques of interest, not only throughout the whole transport safety field; but also, sports injury and safety system research.

### 3 Head injury mechanisms

As King et al. (2003) set out, "the precise mechanisms of brain injury have not been fully established and methods of prevention cannot be fully effective if we do not know the cause." However, whilst this may be the case, it has not prevented many authors hypothesising as to the mechanisms behind head injuries.

There are a few general facts which seem uncontested.

*"Impact injury to the brain can be caused by forces applied to the head and by the resulting abrupt motions imparted to the head. These forces can either be external to the body and act directly on the head, or internal to the body and act on the head through the head/neck junction. (Melvin and Lighthall, 2002)..."*

*Motions of the head and torso quite often result in severe accelerations and large attendant velocity changes that, typically, are both translational and rotational in nature. Such abrupt motions result in inertial or body forces being developed in the brain tissue that in turn result in stresses and deformations throughout the brain. Thus, the state of deformation of the tissue in a region of the brain in a head undergoing impact loading will depend on*

- 1. Its location relative to the point of force application*
- 2. The nature of the distribution of the force*
- 3. The nature of the motion of the head due to the forces acting on the head.*

*In addition, abrupt head motion can also result in relative motion of the brain or parts of the brain with respect to the skull. Such relative motions can deform brain tissue due to impingement upon irregular skull surfaces or interaction with meningeal membranes and can stretch the connecting blood vessels between the surface of the brain and the skull. Finally, one last mechanism of deformation of brain tissue is that of local stretching of the brainstem and spinal cord due to motions produced at the head/neck junction. This motion can occur as a result of either head impact or head motion due to torso loading.”*

Linear and rotational head accelerations are hypothesised to be the primary risk factors for concussion during an impact. Both direct and inertial (i.e. whiplash) loading of the head may result in linear and rotational head acceleration, though neither acceleration would capture the severity of any strain applied over a prolonged period without some additional information about the duration of loading.

Head acceleration induces strain patterns in brain tissue, which may cause injury (Guskiewicz and Mihalik, 2011). Melvin and Lighthall state that injury will occur if the magnitudes of the deformations and stresses induced in the tissues are sufficiently great. Therefore in order to develop truly predictive injury criteria, tissue stresses and strains must be related to dysfunction in physiological processes.

Mao et al. (2006) used a finite element model of a rat brain in indicating that intracranial strains correlate best with experimentally obtained injuries. Unfortunately, the measurement of strain is almost impossible during an impact, particularly *in vivo* (King et al., 2003). It is worth noting that as the cranial contents are nearly incompressible materials, then the injury producing deformations will be due mostly to changes in shape rather than changes in volume (Bandak and Eppinger, 1994).

Current science has not identified an exact threshold for concussive injury, and direct measurement of brain dynamics during impact is extremely difficult in humans. Head acceleration, on the other hand, can be more readily measured; its relationship to severe brain injury has been postulated and tested for more than 50 years. Both linear and rotational accelerations of the head are likely to play important roles in producing diffuse injuries to the brain. However, the relative contributions of these accelerations to specific injury mechanisms have not been conclusively established (Guskiewicz and Mihalik, 2011).

Within the review of Melvin and Lighthall, they cite the work of Holbourn in 1943. Apparently, as early as 1943, Holbourn hypothesised that translational acceleration of the head would not produce significant deformations in the brain due to the incompressible nature of the confined brain tissue. Thus he concluded that shearing deformations, which produce no volume change, caused by rotational acceleration could develop the shear strains throughout the brain required to produce the diffuse effects needed for concussive brain injuries. This was an important turning point in moving away from other notions such as Kramer's contribution to the theory of cerebral concussion (1896). This had proposed a momentary increase of intracranial tension and consequent compression of the brain as causing interference with the blood supply to the entire brain and thus the primary symptoms of cerebral concussion.

Later, in 1974, Ommaya and Gennarelli proposed the centripetal theory of concussion. Their hypothesis for cerebral concussion was defined as,

*"a graded set of clinical syndromes following head injury wherein increasing severity of disturbance in level and content of consciousness is caused by mechanically induced strains affecting the brain in a centripetal sequence of disruptive effect on function and structure. The effects of this sequence always begin at the surfaces of the brain in the mild cases and extend inwards to affect the diencephalic-mesencephalic core at the most severe levels of trauma."* (Ommaya and Gennarelli, 1974)

*"It is suggested that rotational components of accelerative trauma to the head produce a graded centripetal progression of diffuse cortical-subcortical disconnection phenomena which is always maximal at the periphery and enhanced at sites of structural inhomogeneity. The translational components of such trauma are significant for the production of focal injuries only. In this hypothesis the rostral brain-stem (mesencephalon<sup>19</sup> and caudal<sup>20</sup> diencephalon<sup>21</sup>) is the least vulnerable part of the brain and its involvement in the paralytic coma of head injury is always associated with significant injuries to more peripheral parts of the brain."*

Viano put forward a "central" theory for the biomechanics of brain injury (Viano, 1988). Within this theory he suggested that, "rapid motion of the skull causes displacement of the hard bony structures of the head against the soft tissues of the brain, which lag in their motion due to inertia and loose coupling to the skull." The relative displacement between brain and skull brought about by this behaviour, and the resulting deformation of brain tissue and stretching of bridging veins, was proposed as the tissue-level cause of brain injury. However, it is not clear how this suggestion accounts for contre-coup injuries (those occurring on the opposite side of the head to direct loading).

However, in a review of previous work, notably of Gurdjian et al. (1955), Hodgson et al. (1969) stated that, "The evidence indicates that concussion is due to involvement of a specific area in the brain, mainly the brain stem..." Hodgson et al. note that,

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<sup>19</sup> The mesencephalon, or 'mid-brain' is a part of the brain stem. It is the short, constricted portion which connects the pons and cerebellum with the diencephalon and cerebral hemispheres. It is associated with vision, hearing, motor control, sleep/wake, arousal (alertness) and temperature regulation.

<sup>20</sup> Of, at, or near the tail or hind parts. It can mean posterior or inferior depending on the axis or body part being described.

<sup>21</sup> The diencephalon is located deep in the brain underneath the cerebrum and above the pituitary gland. It is the caudal (posterior) part of the forebrain which contains the thalamus and the hypothalamus. It is the link between the nervous system and the endocrine system.

*"In concussion, cell chromatolysis [Disappearance of Nissl granules from nerve cells, showing cell damage] was found primarily in the brain stem and in lesser amounts in the cortex of the dog but almost exclusively in the brain stem of the monkey... Although the motion of the head involved both angular and translational acceleration, the preponderance of affected cells found in the brain stem and the almost complete absence of chromatolysis in the cortex, makes it appear likely that translational acceleration is the most important mechanism."*

In contrast to previous findings related to concussion, Ommaya and Gennarelli (1974) put forward the hypothesis that, "rotational components of accelerative trauma to the head produce a graded centripetal progression of diffuse cortical-subcortical disconnection phenomena which is always maximal at the periphery and enhanced at sites of structural inhomogeneity." In essence, this is an extension of the original hypothesis made by Holbourn (1943) regarding rotational motion. Ommaya and Gennarelli proposed that the translational components of such trauma would be significant in the production of focal injuries only. On that basis they suggested that the rostral brain-stem is the least vulnerable part of the brain and its involvement in the paralytic coma of head injury would always be associated with significant injuries to more peripheral parts of the brain. However, it seems as though this simple way of considering inertial injuries could not be maintained. Instead, it is likely that the influence of rotational or translational accelerations will be dependent on the type and severity of injury caused and its location within the head.

Based on sagittal plane acceleration impulse loading to the head of rhesus monkeys, and consideration of the resulting incidence of brain contusions, Gennarelli et al. (1979) drew the following conclusion:

*"In the case of the frontal lobes, the tangential component and in the case of the inferior temporal lobe, the radial component of the brain movement appear to be the injurious factors."*

A total of 11 mild traumatic brain injury cases were replicated by Zhang et al. (2001) with the Wayne State University Brain Injury Model (WSUBIM) I. The input used for the model was based on reconstructions of head impacts which occurred during games of American football. When looking at the correlation between intracranial stresses with the translational and rotational accelerations, Zhang et al. found that:

*"... the maximum shear stress is highly sensitive to rotational acceleration with highest correlation coefficients and appears to be not directly related to translational acceleration or angular velocity." (Zhang et al., 2001)*

In a later study using vehicle impact test data and parametric finite element analysis, Zhang et al. (2006) found that rotational accelerations contributed to more than 80 percent of the strain in the brain. On this basis, Zhang et al. concluded that rotational accelerations appeared to be the major contributor for the brain strain and, hence, the major cause of strain-induced brain injury. They, therefore, proposed that rotational accelerations should be quantified for improved injury assessment.

Strain rate has also been hypothesised to be a key biomechanical parameter to explain the cause of brain injury and concussion by King et al. (2003).

### **3.1 Car occupants**

Grabow et al. (1984) associated approximately 49 per cent of the total cost of head trauma in the U.S. (precisely Olmstead County, Minnesota) with automobile injuries. They stated that,

*"It would seem plausible from the standpoint of potential economic benefit that the automobile and its occupants should be the primary target of preventative measures." (Grabow et al., 1984)*

Whilst the importance of car occupant injuries will change from place to place and over time, it is likely that a substantial proportion of head trauma is generated by car crashes. Through improvements in restraint system design and the introduction of airbags in cars, there has been a reduction in the prevalence of head injuries for car occupants. Therefore, improvements in car crash safety have had an effect on this 'primary' target. However, occupants can still receive head injuries in car crashes. For example a third of car driver fatalities in car-to-car or car-to-light goods vehicle impacts will receive a head injury (Richards et al., 2010).

Davcera et al. (2012) analysed a series of eighty cases with fatal closed head injuries. In this series, cases where the survival time was at least 2 hours were also analysed for the occurrence of diffuse axonal injury. There were 49 cases from traffic accidents and 25 of these showed evidence of diffuse axonal injury (diagnosed with clinical and pathological criteria, e.g. visualised via immunohistochemical staining). Within this group there were five drivers all of whom had diffuse axonal injury. Acute subdural haematoma occurred for only 1 of 12 drivers or passengers.

Alongside the development of the National Highway Traffic Safety Administration's (NHTSA's) test procedure for evaluating small overlap and oblique impacts, a variety of impact configurations were considered. Saunders et al. (2011) compared injury assessment reference values from seven such tests. Those authors reported that the rotational injury measure, BRIC, was exceeded in four of seven tests, whilst the linear acceleration based measure, HIC, was not exceeded in any test. In this instance, the BRIC measure was intended to represent a 30 percent probability of diffuse axonal injury occurring. As noted in Section 3.3.5 and Carroll (2010), the HIC<sub>15</sub> limit of 700 represented a risk of skull fracture between 5 % and about 30 % depending on the literature source and the analysis technique used to define the risk function. As such there is a continued need to protect the head of car occupants. Additionally, the safety industry may need to respond to the threat posed by diffuse injuries as well as focal injuries.

Seacrist et al. (2010) remarked that as restraint systems restraints become more effective at limiting peak loads applied to car occupants, it may become increasingly important to evaluate occupant behaviour in loading environments that have historically been considered as being relatively benign. Through a small series of low severity sled tests they went on to consider the biofidelity of a six year old child crash test dummy in simulations of low speed crash events (forward facing, 4 g, 120 ms duration). During these tests they observed that paediatric volunteers could undergo 31 to 49 degrees of head rotation at angular rates up to 6 to 10 rad.s<sup>-1</sup>. Whilst these values represent non-injurious motion, it is clear that conventional vehicle occupant seat belt restraint systems allow substantial movement of the head during an impact event.

## **3.2 Vulnerable road users**

According to EU Injury Database (IDB) estimates, 4.2 million road injuries per year have to be treated in EU hospitals. On average in the EU almost two-thirds of these road injury victims are vulnerable road users (hospital admissions and outpatients): 6 % pedestrians, 18 % motorized two-wheelers, 40 % pedal cyclists. Head injuries have a high share in these road traffic injuries. They account for 36 % of the injuries for pedestrians, 34 % for bicyclists, and 24 % for two-wheeled motor vehicles (EuroSafe, 2013).

### **3.2.1 Pedal cyclists**

In the Katholieke Universiteit (KU) Leuven cyclist database, brain contusions were the second most frequent bicycle-related head injury (63 %) and associated acute subdural haematoma was seen in two-thirds of these cases (Musigazi et al., 2015). Also a forensic review of 80 cases of closed head injury in Skopje, Macedonia (Davceva et al., 2012) revealed that acute subdural haematoma was mostly found in cyclists, simple falls and cases of assault. Of the brain contusions that were associated with a coup or contrecoup or fracture mechanism, by Musigazi et al. (2015), nearly 96 % occurred at the frontal and temporal lobes. The inferior lobe surface was the

primary site for sagittal impacts and the lateral temporal lobe surface in lateral impacts. Musigazi et al. were led to the hypothesis that the majority of frontal and temporal contusions in pedal cycle accidents result from the forceful contact of the cortical surface against the skull interior; from compressive strains. They assert that the magnitude of such compressive strains is associated with the amplitude of the head rotational acceleration. This is unlike prolonged exposure to rotational acceleration where brain deformation can be induced and diffuse axonal injury can be produced.

### **3.2.2 Pedestrians**

For pedestrians struck by a car, the risk of that person being killed increases slowly with increasing impact speed until speeds of around 30 mph. Above this speed, the fatality risk increases rapidly by between 3.5 and 5.5 times from 30 mph to 40 mph (Richards, 2010). Whilst the risk of a pedestrian being killed at 30 mph is comparatively low, approximately half of all pedestrian fatalities occur at this impact speed or below, because most car collisions with pedestrians occur at speeds up to 30 mph.

In a study by Reith et al. (2015) a dataset documented in the TraumaRegister DGU® (a large multi-centre database for anonymous and standardized documentation of severely injured patients that was initiated by the German Trauma Society in 1993.) of the years 2002 to 2012 was analysed. Those authors characterised injured pedestrians (n = 4435) and compared the findings with a control group of motor vehicle occupants involved in road traffic collisions (n = 16,042). The main findings were that, firstly, compared to motor vehicle occupants more women, children and elderly citizens were involved in pedestrian accidents. Secondly, in motor vehicle to pedestrian collisions, victims head, pelvis and lower extremities were more commonly and more severely injured than in the motor vehicle occupant group. Thirdly, injured pedestrians showed a higher mortality compared to motor vehicle occupants in spite of a shorter rescue time and nearly similar ISS (Injury Severity Score: a score combining the AIS from the three most severely injured body regions). They comment

that it is the injuries themselves and not the underlying collision mechanism which mostly influence mortality for those involved in road traffic collisions. Age and severity of head injury exerted the strongest impact on prognosis and mortality. These were the two factors which differed most between pedestrians and motor vehicle occupants in their observations.

Pedestrians struck by a pickup truck, van or SUV (sports utility vehicle) are at greater risk of severe injury or death than pedestrians struck by cars. Also, risks are higher for an older pedestrian struck at any given speed than for a younger pedestrian struck at the same speed (Tefft, 2012).

Richards and Carroll (2012) analysed Hospital Episode Statistics data (from April 2008 to March 2009) selecting pedestrians in impacts with cars who had a head injury as their primary injury. The primary injury was defined as the main condition treated, or the main symptom where there was no definitive diagnosis. Richards and Carroll found that superficial injuries, open wounds of the head, and unspecified injuries are the most frequently occurring for pedestrians in impacts with cars, but it is intracranial injuries which lead to the longest average stay in hospital, followed by fractures. When ranked by cumulative duration of stay, the three top injuries were diffuse brain injury, traumatic subdural haemorrhage, and traumatic subarachnoid haemorrhage. These are the injuries which account for the greatest burden to hospitals using the metric of duration of stay.

In reviewing discharge data for pedestrian casualties from eight European countries, Arregui-Dalmases et al. (2010) conducted an extensive analysis of the mechanisms responsible for head injuries. Their sample size in this analysis was 3403 pedestrians with head injuries. Diagnoses related to traumatic brain injury were classified according to an algorithm to identify whether the injuries related to translational accelerations, rotational accelerations or either. The basis for the classification was:

*"Focal injuries are caused mainly by direct impacts to the head and they encompass contusions, lacerations and hemorrhages that produce hematomas in the extradural, subdural or intracerebral compartments of the head. Diffuse injuries are often caused by inertial mechanisms in*

*which there is relative motion of the cranial contents. If the inertial acceleration is translational, the most common injuries are vasculature injuries such as countercoup contusions and subdural hematomas. However, if the inertial motion is caused by a rotational acceleration, the injuries associated are caused by strains that cause Diffuse Axonal Injuries (DAI) associated to cerebral concussions, post-traumatic coma and unconsciousness. Therefore, it is possible to associate a specific brain injury to a specific type of acceleration.”*  
(Arregui-Dalmases et al., 2010)

After applying their algorithm, Arregui-Dalmases et al. reported that more than half of the injuries to the brain are associated exclusively to a rotational acceleration. However, it should be noted that classification basis is an assumption. The analysis did not prove any link between rotational or translational accelerations and the incidence of a particular type of injury. Identifying such a link is the aim for future research and the validation of a head injury criterion and its relationship to injury risk.

Kimpara and Iwamoto (2012) applied existing head injury criteria to finite element model impact data from two pedestrian accident reconstructions, as reported by Dokko et al. (2003). In both cases head injuries occurred; but Kimpara and Iwamoto found that in one of the cases there was a high risk of injury predicted by the linear motion metrics and in the other case the rotational metrics dominated. They propose the use of HIC, RIC and PRHIC for assessing head injury risk; criteria which are described in the following section.

### **3.3 Existing criteria**

To be effective in its purpose, a head injury criterion must provide correlation between the criterion score and risk of injury for a head exposed to the same event conditions. It then follows that for increasing impact severity, the risk of injury would also increase, as would the measured criterion. Conventionally, injury risk is determined by conducting a suite of impact tests using PMHS (Post-Mortem Human Subjects) or animal subjects around the injury threshold severity, or by having detailed information

on real world accident cases (where that information is often obtained through reconstructing the accident under controlled conditions and using instrumentation to assess the loading to the test subject). Parameters describing the impact conditions, or criteria derived from those parameters can then be used to describe how injury risk changes with changes in impact severity. Statistical analysis will show to what extent the impact severity measure (injury criterion) is able to describe the variation in injury risk.

Unfortunately, the surrogates for living humans used conventionally in injury risk determination are not ideal to the investigation of brain injury. A summary of the reasons why was reported by Newman (1986):

- *"PMHS experiments can, at best, provide insight into [some] brain injuries of AIS  $\geq 3$ . Physical disruption of brain tissue may be observed in PMHS autopsies by the extravasation of fluid dyes injected into the arterial system before impact. However, diffuse axonal injuries associated with brain cell damage, which might appear as concussion or generalised diffuse brain injury are not visually evident."*
- *"Animal experiments do permit the observation of the effects of an impact resulting in minor injuries. Animals will exhibit concussion and/or temporary brain dysfunction. However, of course, except through dubious methods of scaling, such data cannot provide numerical limits directly applicable to humans. Trends (if they exist) however, can be discerned and can lend support to (or discredit) a particular model."*
- *"Experiments with volunteers are always limited to non-injurious situations but, as such, can possibly provide lower bounds on tolerance limits."*
- *"Accident victims can be subject to the entire range of brain injury but, except for very special cases, are associated with too many unknowns to be of much value in a validation exercise."*

Newman was very critical of the potential options for determining brain injury tolerance in humans, perhaps overly so. For instance, accident investigation techniques have been improving continually since they started. Information and data provided through accident reconstructions can now offer very useful information regarding the loading to an injured person (e.g. in the development of injury criteria for children and child dummies; Johannsen et al., 2012), although there will always remain some unknown details of exactly what occurred at the time. Furthermore, recent advances in technology are suggesting even better data will be available in the future. Now black box technology can provide crash data from accidents themselves and the level of information available continues to improve as the technology matures.

### **3.3.1 Peak translational acceleration**

The simplest form of head injury criterion is to consider purely the peak linear acceleration. The acceleration of a head surrogate is routinely monitored in most forms of test work with specifications in place for sampling conditions and post-event processing. By simply specifying a threshold which cannot be exceeded, engineering solutions to reduce the peak linear acceleration can be encouraged or enforced. Often the peak value is set for the peak resultant acceleration. Alternatively, where coordinate systems are important and defined, it is possible to set direction-dependent thresholds. However, peak translational acceleration limits tend to be used in a relatively pragmatic, rather than quantified injury risk analysis, way. As such they tend not to be related to a specific type and severity of injury but rather are used to express a general severity of loading to the head.

One application is in the approval of motorcycle helmets for motorcycle and moped riders and passengers. In UN Regulation No. 22, the impact test criterion is that the resultant acceleration measured at the centre of gravity of the headform at no time exceeds 275 *g*, and the Head Injury Criterion does not exceed 2400.

Laboratory reconstructions of video-recorded concussions using helmeted Hybrid III dummies were used to suggest that an injury threshold of 70-75 *g* exists for sustaining concussion based on the translational (linear) acceleration of a American

football player's head (Guskiewicz and Mihalik, 2011). This conflicts with the results from reconstructions of NASCAR accidents out of which 44 from 4,071 impact events led to mild concussions with or without loss of consciousness and higher injury thresholds were proposed. Following numerical simulation of the impact event reconstructions, Somers et al. (2011) proposed the risks of injury shown in Table 3-1.

**Table 3-1: Mild concussion risk, with and without loss of consciousness (AIS ≥ 1, AIS ≥ 2, respectively) (Somers et al., 2011)**

Risk of injury	AIS ≥ 1 (g)	AIS ≥ 2 (g)
5 %		133.2
25 %	176.6	188.5
50 %	215.0	226.9

### 3.3.2 Wayne State Tolerance Curve

The Wayne State Tolerance Curve may be considered as the foundation of research on human head injury criteria. To study the head impact mechanism and evaluate the characteristics of the injury inflicted, Ford Motor Company sponsored a series of head impact studies at the Medical College of Wayne State University, the first of which was undertaken in 1954.

The tolerance curve was derived on the key principle that higher acceleration and longer duration events were more likely to be injurious (i.e. high acceleration events are tolerable if the duration is short). The exact relationship was derived on the basis of:

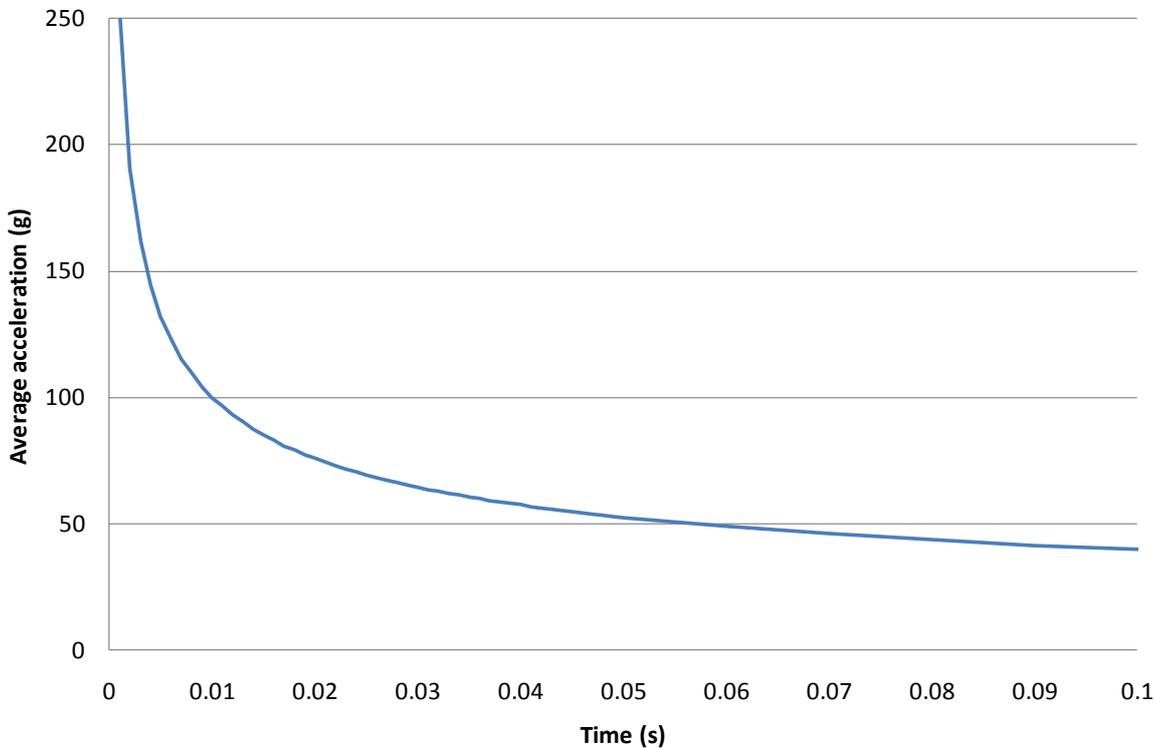
- PMHS tests, where the subject was dropped onto their head
  - In the preliminary series of medical tests, heads alone were used. Later studies employed the complete cadaver to simulate better the head impact of an automobile occupant (Haynes and Lissner, 1961). Skulls were X-rayed after impact to determine when a fracture occurred. The first tests were impacts on solid steel blocks, clamped panels of sheet metal and production instrument panels.
  - In the testing reported by Patrick et al. (1963), PMHS were dropped a known distance and the forehead impacted a heavy steel block. The brain was replaced by gelatine of the same specific gravity to allow pressure measurements to be made.
- Animal experiments
  - Blasts of air administered to the exposed brains of dogs:
    - Air pressure pulses of varying amplitude and duration were investigated by applying air pressure blasts to the dura mater of dogs.
  - Impact to the heads of dogs with a rotary hammer provided further information on the cranial pressure and acceleration-time tolerance to concussion in dogs.
- Volunteer sled tests

Tolerance is expressed in terms of average head acceleration and the period at which that level is maintained during the event. The maximum product of the acceleration and time period can then be compared against a threshold at which it may be considered that head injury is likely (see Equation **3-1**). In this equation, ' $\bar{a}^m$ ' is a peak, average acceleration measured (raised to the power 'm') for the period, 'T'. Versace (1971) reported that the power should be between 1.9 and 3.2 depending on which part of the Wayne State Tolerance Curve was considered a priority for close fitting.

$$\bar{a}^m T = N$$

**Equation 3-1**

The following figure (Figure 3-1) shows this relationship with the exponent, 'm', set to 2.5.



**Figure 3-1: Tolerance of the human head expressed in terms of average acceleration and duration of the event**

The short duration part (two to six milliseconds) of the Wayne State Tolerance Curve was obtained from PMHS head impacts on rigid steel plates. Skull fracture was chosen as the injury criterion in these tests on the assumption that PMHS skull fractures would correlate well with living human concussion. The dog experimental concussion-pressure-time relationship along with head acceleration-pressure-data obtained from the series of PMHS drop tests onto automobile dash panels were used to construct the intermediate time domain of the curve, up to 10 ms (Hodgson et al., 1970).

As stated by Patrick et al. (1963), the Wayne State Tolerance Curve is based upon impact to a hard, flat surface and consequently is a severe condition. For low velocity impacts the high acceleration, short time portion of the curve is significant. At higher velocities the time can be so long that the nearly constant acceleration portion of the

curve prevails. The Wayne State Curve for long pulse duration is based on sled runs with volunteers. All of these sled runs were found to be below any head tolerance levels. Patrick et al. suggest that for times in excess of 50 ms, use of constant acceleration is justified. They also comment that the 42  $g$  (where  $1 g = 9.8 \text{ m}\cdot\text{s}^{-2}$ ) asymptote has been exceeded without serious injury when there was no impact. Consequently, when an impact to a padded surface occurs, a higher effective acceleration limit would be in order. They estimated that a 60 to 80  $g$  limit would be reasonable. Reconstruction data from NFL American football impacts suggested a value of about 70 to 75  $g$  for concussion in padded impacts (Pellman et al., 2003). The MSC curve for front head impacts was found to be approximately 80 percent above the Wayne State Curve for most pulse durations greater than 10 ms (Stalnaker et al., 1971). The MSC curve for front impact raises the tolerance level from 50  $g$  for long pulse duration (greater than 20 ms) to 90  $g$  for the same pulse duration.

It has been confirmed that the criterion employed in the Wayne State Tolerance Curve is nearly the same as the concussion threshold level of the JARI Human Head Tolerance Curve (JHTC; Ono, 1998). However, the most well-known expression of this form was proposed by Gadd as the Severity Index (Gadd, 1966).

### **3.3.3 Severity Index**

The work of Gadd (1966) adopted the relationship expressed in Equation **3-1** but proposed the use of 2.5 as the exponent for internal head injury. This value was derived from the animal data taken from the Wayne State University work and was said to represent dangerous concussion. According to Gadd, the value was approximately the same as that selected by Eiband (in a NASA memorandum) for anterior-posterior acceleration of the seated human. Gadd also reported that he and his colleagues at General Motors Corp. had been using a value of 1000 for the threshold for serious internal head injury in frontal impacts. Hence the resulting Severity Index became the expression shown in Equation **3-2**, where 'ã' is the 'effective' linear acceleration of the head (acting through the centre of gravity) and 'T' is the duration of that acceleration.

$$\bar{a}^{2.5}T < 1000$$

**Equation 3-2**

Following the proposal by Gadd, Versace (1971) wrote a critical review of the Severity Index. This identified a number of flaws in the assertions made by Gadd.

1. The Wayne State University data used by Gadd was based on head acceleration; whereas, the data suggested as confirming the relationship (from Eiband) was based on sled accelerations. Acceleration data from the two measurement sources may not be directly equivalent. The sled-based acceleration is liable to have a higher peak value for a shorter duration than the head acceleration.
2. Different durations of head acceleration may be expected to induce different injury mechanisms. Approximating the Wayne State Tolerance Curve to a linear relationship based on all impact durations gives equal priority to all impact durations. Versace suggested that a focus should be given to typical duration impacts from car crashes (if car crashes were being investigated predominantly).
3. The Severity Index uses the average acceleration of the impulse duration. This would give a very different result from consideration of the peak value when comparing, for instance, a square-wave and a triangular-wave acceleration pulse.

Of course, this last point may be an advantage if average, rather than peak, acceleration related most closely to the risk of injury.

One problem with the Gadd Severity Index (GSI), raised by McElhaney (2005) is that, "it keeps on integrating and must be arbitrarily ended. Thus one  $g$  for 1000 seconds yields a GSI of 1000 seconds." However, we all experience one  $g$  (through gravitational acceleration) all of the time, without substantial head injury risk.

Versace suggested that ' $\bar{a}$ ' should be expressed as a mathematical average and proposed the alternative equation shown below (Equation **3-3**).

$$\left(\frac{1}{T} \int a(t) dt\right)^{2.5} T < 1000$$

**Equation 3-3**

### 3.3.4 Time-limited acceleration

To deal with the time dependent nature of brain injury, one approach has been to specify time durations, at particular levels of acceleration, which if exceeded would be expected to be injurious. For example:

$$a_{\text{maximum}} < 400 \text{ g}$$

$$\text{time at } 200 \text{ g} < 2 \text{ ms}$$

$$\text{time at } 150 \text{ g} < 4 \text{ ms}$$

These limits are used in the U.S. Federal Motor Vehicle Safety Standard (FMVSS) 218 for motorcycle helmets. The biomechanical basis of the FMVSS 218 helmet criterion is not well understood (Rigby and Chan, 2009).

In UN Regulations concerning vehicles and vehicle safety, acceleration exceeding 3 ms duration is often used as an assessment criterion. Examples of this are in the UN Regulations No. 44 and 129 on child restraint systems and No. 94 and 137 for adult occupant protection in frontal impacts. Limits are placed so that the acceleration cannot exceed either 75 or 80 g (for young children or older children and adults, respectively) for more than 3 ms.

### 3.3.5 Head Injury Criterion

In order to address the criticisms of the Severity Index made by Versace, and making use of some of the suggestions made in that review, in 1972 the US National Highway Traffic Safety Administration (NHTSA) adopted a slightly modified Severity Index. The modified parameter was called the Head Injury Criterion (HIC) and can be expressed as in Equation 3-4. In this equation, 't<sub>1</sub>' and 't<sub>2</sub>' are any two temporal points in the impact (in seconds), and 'a' is the resultant acceleration of the centre of mass of the head (in g).

$$\left[ \frac{\int_{t_1}^{t_2} a \cdot dt}{t_2 - t_1} \right]^{2.5} (t_2 - t_1) < 1000 \quad \text{Equation 3-4}$$

To make use of the HIC in head contact events, Hodgson and Thomas (1972) put forward the hypothesis that;

*"If a head impact does not contain a critical HIC interval of less than 15 ms, it should be considered safe as far as cerebral concussion is concerned."*

The implication of this statement is that impact events lasting longer than 15 ms, regardless of the acceleration level, will not cause cerebral concussion. This was suggested on the basis that the Wayne State Tolerance Curve was more reliable for impacts with a short duration.

As the HIC has become increasingly widely used, it has become subject to extensive evaluation and some criticism. For example, Newman (1998) identified that:

- The HIC contains no reference to rotational kinematics
- Acceleration of the head, as a whole, may not be very relevant for a deforming, multi-modal structure composed of fluids and solids about to fracture
- For time durations greater than seven ms, rather than basing the HIC on Equation **3-2**, a better fit to the Wayne State Tolerance Curve would be that shown in Equation **3-5**

$$\bar{a}^{3.2}T < 9580$$

**Equation 3-5**

- Concussion, as well as other types of brain injury, can occur when linear head acceleration pulses and HIC intervals happen to exceed 15 ms in duration

Expanding on the second of these issues, the short-duration part of the Wayne State Tolerance Curve is based on unidirectional translational accelerations, measured at the back of the head and assumed to be representative of measurements at the centre of mass of the head. This assumption is incorrect as the head is not a rigid body but can deform during an impact (particularly when associated with skull fracture).

McElhaney, 2005 identified many factors which are important to head injury but not included in the HIC.

*"These factors include location of the impact, area of contact, stiffness of the impacting surface and angular accelerations induced by off CG [centre of gravity] blows and torso restraint." (McElhaney, 2005)*

As Koch (1988) reported, HIC data is a "mixture of apples and bananas". This is because a skull fractured in an experiment is likely to give a lower HIC value due to the increased time of deceleration during the deformation and fracture of the skull. Therefore, "HIC values for fractured and non-fractured skulls do not measure the same property." That is to say, a skull that fractures would not be expected to produce the same HIC value as a skull that did not fracture under identical impact conditions. The fracture case will tend to result in a lower HIC value. This causes increased uncertainty in developing risk functions for head injury where both skull fracture and no fracture cases are included. Around the threshold input level, it may not be the case that risk of injury increases with applied impact severity, if severity is assessed by the HIC or translational acceleration.

When comparing measurements made with a rigid, non-frangible headform, it should be taken into consideration that the headform will tend to give higher translational acceleration and HIC values than would be the case for an identical impact with a human head. Such a difference is likely to be profound for impact conditions where the human head would fracture, thereby attenuating the acceleration.

An analysis of field accident data by ISO Working Group 6 indicated that there were no cases of brain injuries to the three-point belt restrained car occupants whose head did not impact forward interior components (Prasad and Mertz, 1985). Therefore, HIC was not developed or suggested for use in non-head impact cases. All of the biomechanical data used in the development of the HIC were head contact related.

Whilst HIC was principally developed to investigate the risk of head-contact related injuries (often including a skull fracture), Prasad and Mertz (1985) report on the use of HIC in predicting brain injury. A limitation of the injury data is noted in that since only the arterial system was pressurised for the original head impact tests, damage to the venous system was not measured. Therefore venous ruptures, which result in subdural haematomas (AIS 4), would not have been recorded. Neither would it have been possible using these test methods to detect diffuse axonal injuries due to brain cell damage that may result in concussion (AIS 2 to 5); or other brain injuries like cerebral oedema and swelling.

Prasad and Mertz comment that the risk curves drawn by them using the "Mertz/Weber Method"<sup>22</sup> for skull fracture and brain damage were virtually identical. They infer that for a given level of HIC, skull fracture, brain damage, or both, are equally likely to occur. When analysed using an appropriate statistical technique, this inference is not supported by their dataset.

In his 2005 Stapp memorial lecture, McElhaney relayed a conversation with Dr. Gurdjian (one of the original developers of the Wayne State Tolerance Curve); apparently the HIC was such a considerable extrapolation of the original data that it far exceeded the authors' expectations.

In summary:

- The Head Injury Criterion (HIC) was developed based on the short duration (two to six milliseconds) part of the Wayne State Tolerance Curve. The application of HIC tolerance thresholds to long duration events has not been validated.
- HIC does not account for rotational head motion.
- For the PMHS impact tests used in the determination of HIC risk functions, accelerometers were not mounted at the centre of gravity of the head. This may create a systematic error when relating HIC values to dummy measurements.

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<sup>22</sup> Mertz and Weber (1982) used a modification to the Median Rank method where the input parameter values are ranked from the lowest value associated with specimen failure to the highest value associated with no failure. A line is then drawn between the two end points to provide an estimate of the cumulative distribution of the failure thresholds for the tested specimens.

The potential for this method of generating risk curves to be misleading was demonstrated by Hertz (1993). She showed very similar risk functions for HIC vs. skull fracture for Normal, logNormal and Weibull parametric methods, giving approximately 40% to 45% risk of skull fracture at a HIC of 1000; but when using the Mertz/Weber function for the same data, it gave a risk of <20% at the same HIC level.

The Mertz/Weber Method for developing risk functions was reviewed again by Hynd *et al.* (2006) and not recommended for further use.

- Despite inferences by some authors, the HIC risk function does not relate to brain injury in the manner those authors claim. That is not to say that HIC cannot be used to predict brain injury, only that the existing functions have not been defined in an appropriate way to do so.
- The Mertz/Weber and Certainty methods of analysing data were used in developing published, and accepted, HIC risk functions. These functions are not supported statistically by the data used to define them.
- The data used in validation of the HIC represents an assortment of impact conditions. It may be that, rather than trying to apply HIC broadly to all conditions; it would give greater significance to specify new injury risk functions for each.

### **3.3.6 Maximum rotational acceleration**

Because activities in the midbrain and upper brainstem are responsible for alertness and responsiveness, rotational mechanisms of TBI are believed to more likely result in loss of consciousness than predominantly linear types of impacts (Guskiewicz and Mihalik, 2011).

Severe whiplash injuries were produced in monkey subjects throughout the 1960s, as reported by Ommaya et al. (1967). Ommaya et al. proposed a scaling strategy for converting the monkey tolerance to a concussion threshold for man. This scaling was derived following discussions with Holbourn, who had proposed a scaling based on brain mass and the original hypothesis linking rotational motion with head injuries caused through inertial loading (Holbourn, 1943). One of the key assumptions used in the scaling process was that there is a three-dimensional geometric similarity between the brains of a series of subhuman primates and the human brain. However, many other assumptions and caveats are listed by Ommaya et al.

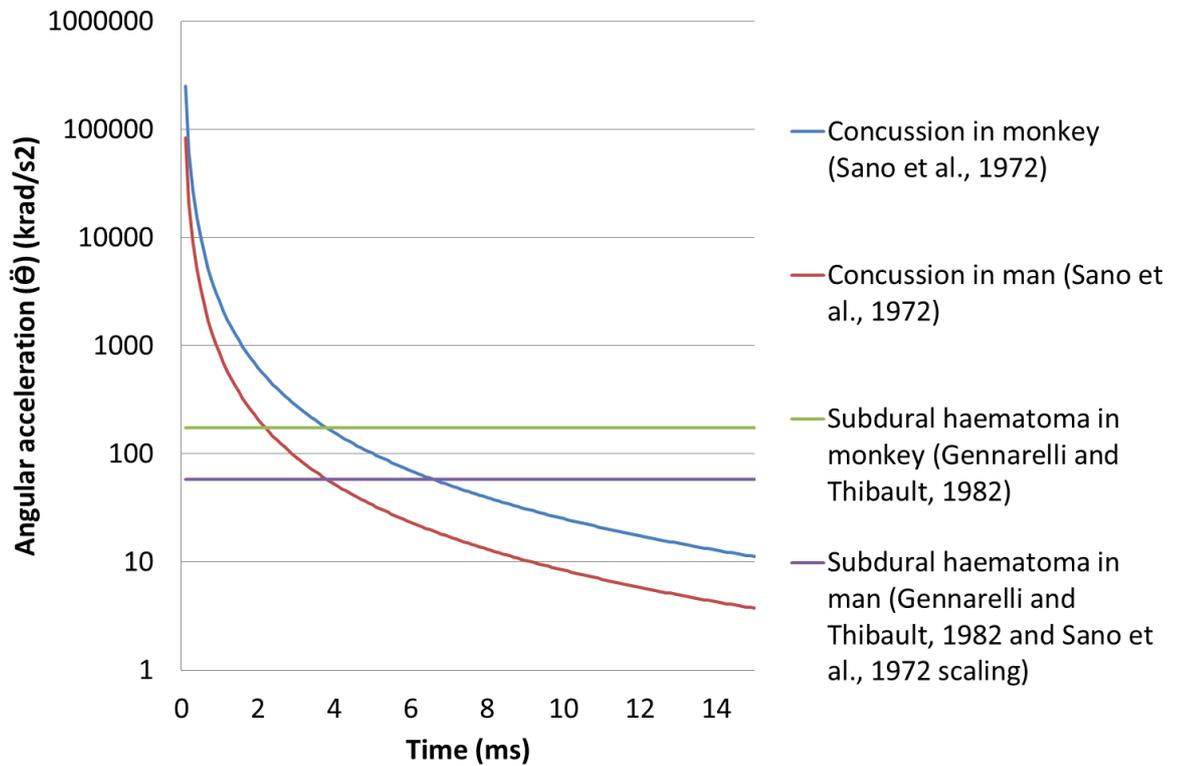
The result of the work by Ommaya et al. was to suggest that the cerebral concussion tolerance of about  $40,000 \text{ rad}\cdot\text{s}^{-2}$  observed with rhesus monkeys in sagittal plane rotations (note that this tolerance is time dependent) equated to  $7,500 \text{ rad}\cdot\text{s}^{-2}$  for a human.

Ommaya and Hirsch (1971) also proposed that, "levels of head rotation during whiplash, in excess of  $1,800 \text{ rad.s}^{-2}$ , will probably result in cerebral concussion in man."

However, Ewing (1975) notes that human volunteer exposures have shown no adverse effects at  $38 \text{ rad.s}^{-1}$  head angular velocity, nor with head angular acceleration of  $2,675 \text{ rad.s}^{-2}$ . Ewing suggests that the reason for the discrepancy may lie in the lack of direct measurement by instrumentation on the primates.

Unterharnscheidt (1971) reported on test results investigating tolerance to rotational acceleration. Effects of rotational acceleration on the central nervous system were reported based on experiments with squirrel monkeys. The monkeys were subjected to rotational accelerations of 101 to  $386 \text{ krad.s}^{-2}$ . A continuum of responses was observed, from no signs of central nervous system damage through concussion to death.

The lowest rotational accelerations employed were 101 to  $150 \text{ krad.s}^{-2}$ . These experiments caused no primary or secondary alterations in the cerebrum. The next highest accelerations, up to  $197 \text{ krad.s}^{-2}$  produced subarachnoid haemorrhages in 10 out of 13 animals. Accelerations of more than  $200 \text{ krad.s}^{-2}$  caused severe primary traumatic haemorrhages in the cortex and white substance. Rotational acceleration of more than  $300 \text{ krad.s}^{-2}$  was not survived. These monkeys were the only animals to show additional haemorrhages in more central regions of the brain. Gennarelli and Thibault (1982) proposed that a rotational acceleration exceeding  $175 \text{ krad.s}^{-2}$  would produce subdural haematoma in the rhesus monkey. These authors noted the expectation that there would be a relationship between acceleration and duration. This follows the work of Sano et al. (1972) who (citing Hayashi et al., 1969) proposed the product of angular acceleration and duration as a criterion for concussion with a limit of 2.52. An equivalent curve is shown in Figure 3-2. Sano et al. also suggest that the tolerance in man would be about one third of that in the monkey.



**Figure 3-2: Concussion threshold relating maximum angular acceleration,  $\ddot{\theta}$ , with impact duration (Sano et al., 1972)**

From Figure 3-2 it appears that a rotational acceleration of about 100 krad/s<sup>2</sup> applied for a duration of 3 to 5 ms would be about the threshold of both concussion and subdural haematoma, broadly. Applying 100 krad/s<sup>2</sup> for 3 ms would generate an angular velocity change of 300 rad/s. Maintaining an average rotational velocity of 150 rad/s for 3 ms would generate an angular change of 0.45 radians; 300 rad/s would correspond with 0.9 radians. Such an angular displacement is entirely plausible for the movement of a vehicle occupant's head in an impact event or for a pedestrian. It may be less likely for a top-level driver in motorsports, where their head is restrained by a HANS (head and neck support) device. In that application the physical constraint placed on head motion may keep the rotational accelerations and velocities below proposed thresholds. Based on finite element modelling of a Hybrid III dummy head in a motorcycle helmet, Aare et al. (2003) concluded that:

*"When measuring the strains in the brain tissue in the Finite Element model of the human head subject to oblique impacts, it could be seen that injury thresholds should include rotational parameters as well as translational parameters."* (Aare et al., 2003)

From the body of literature reviewed it seems clear that one should expect a relationship between rotational acceleration and haemorrhages within the head as well as further damage to the substance of the brain (whether that is bleeding or axonal disruption). Testing with rats, Stemper et al. (2015) identified that increasing rotational acceleration magnitude produced longer unconsciousness times, which were used as an assessment of acute injury severity. However, they also determined that longer duration rotational accelerations produced changes in the 'emotionality' of the rats, as measured using the Elevated Plus Maze assessment. This suggests that it is important to monitor the duration as well as the peak magnitude of the rotational acceleration if behavioural sequelae are to be investigated as well as the acute injury severity.

### **3.3.7 New Mean Strain Criterion**

The Mean Strain Criterion (MSC) was originally put forward as a head injury criterion in the early 1970s. It was based on a series of mechanical impedance experiments which allowed the conceptual characterisation of the head with two masses, coupled by a spring and dashpot. However, as noted by Stalnaker et al. (1985), because the MSC was developed using separate lateral and frontal head models, and was based on limited PMHS information, the MSC soon gave way to the HIC due to confusion and misunderstanding. To solve this confusion and misunderstanding the criterion was updated to include four directional models:

- Anterior-Posterior (A-P)
- Posterior-Anterior (P-A)
- Superior-Inferior (S-I)
- Left-Right (L-R)

The model inputs and outputs were also redefined and the injury criterion updated with additional primate and PMHS data. This led to the New Mean Strain Criterion (NMSC).

The outputs from the NMSC are strain and strain rate values which provide linear correlation to AIS values. However, it should be noted that the AIS uses discrete levels and so interpolation of levels is not an appropriate way of considering such data.

An interesting finding from the work of Stalnaker et al. is on the sensitivity of the head to impacts from different directions. Using a sine wave input pulse to the NMSC model, Stalnaker et al. showed that the L-R direction may be more vulnerable than the A-P, P-A, or S-I directions.

### **3.3.8 GAMBIT**

The Generalised Acceleration Model for Brain Injury Threshold (GAMBIT) was proposed by Newman (1986). The concept behind the GAMBIT was to have a criterion for brain injury which took into account the combined effects of both translational and rotational kinematics.

The original GAMBIT equation is shown in Equation **3-6**.

$$G(t) = \left[ \left( \frac{a(t)}{a_c} \right)^n + \left( \frac{\alpha(t)}{\alpha_c} \right)^m \right]^{\frac{1}{s}} \quad \text{Equation 3-6}$$

In the equation 'a(t)' and 'α(t)' are the instantaneous values of translational and rotational acceleration respectively and 'n', 'm', and 's' are empirical constants selected to fit the available data. Newman comments that setting n=m=s=1 provides a simple linear weighting of the translational and rotational components. Alternatively, n=m=s=2 provides an elliptical function for the two kinds of motion. The terms 'a<sub>c</sub>' and 'α<sub>c</sub>' are the limiting "critical" values. These limits need to remain valid when the brain is loaded by the combined effects of both forms of motion.

Newman noted that it had been suggested that the tolerance of the head to impacts from different directions is different. To account for this within the GAMBIT, he suggested that one might have “critical” accelerations for each of the six degrees of freedom. The equivalent translational and rotational acceleration terms would then take the form shown in Equation 3-7. The  $a_c$  values represent the orthogonal critical values and  $a_x$ ,  $a_y$ , and  $a_z$  are the component values in the three principal directions. A similar expression can be written for the rotational terms. However, Newman commented that whilst such refinement is technically appropriate, it would be beyond the scope of the currently available data to define the necessary intercept values.

$$\left(\frac{a}{a_c}\right)_{equiv} = \left[ \left(\frac{a_x}{a_{cx}}\right)^2 + \left(\frac{a_y}{a_{cy}}\right)^2 + \left(\frac{a_z}{a_{cz}}\right)^2 \right]^{\frac{1}{2}} \quad \text{Equation 3-7}$$

Another limitation of the GAMBIT noted by Newman is that it does not account for the time-dependence of tolerable accelerations. He suggested that a third dimension, taking account of time, could be added to the GAMBIT function to define a three-dimensional surface limit, rather than a line. It was noted that, “Whether this ‘improvement’ could ever be validated is a matter for future speculation.”

Newman concludes that a straight line intersecting the translational acceleration axis at 250 *g* forms a triangular region within which injuries to PMHS are typically not found when the rotational acceleration intercept is approximately 10,000 rad.s<sup>-1</sup>. Therefore the resulting GAMBIT formula proposed by Newman remains relatively simple and is shown in Equation 3-8.

$$G = \frac{a_{max}}{250} + \frac{\alpha_{max}}{10,000} \leq 1 \quad \text{Equation 3-8}$$

### **3.3.9 Brain Compliance Model**

As an extension to the New Mean Strain Criterion, Viano (1988) suggested the Brain Compliance Model. This used a similar description of the head in terms of lumped masses, springs, and dashpots; however, the compliance characteristics were changed to represent a rigid skull and displaceable brain. The basic premise was to consider the viscous response of the brain-skull motion, in a similar manner to that which Viano had implemented for considering other organ injuries (e.g. thoracic Viscous Criterion). This viscous response used the strain (or 'C' compression) multiplied by the strain rate ('V' velocity of the deformation).

Outputs of HIC, strain criteria, and the brain compliance model were compared from a series of sled tests using a Hybrid III dummy. Viano concluded from these tests that, "the viscous mechanism of brain injury may lead to a better interpretation of injury risk from lumped-mass modelling of Hybrid III dummy head dynamics."

### **3.3.10 Maximum rotational velocity**

From their parametric investigation of regional brain strain responses using the Dartmouth Head Injury Model, Zhao and Ji (2015) suggested that peak rotational velocity is a more relevant kinematic indicator for regional strain responses than peak rotational acceleration alone.

The culmination of several parallel research paths was reported by Margulies and Thibault (1992) as a proposed tolerance criterion for diffuse axonal injury (DAI) in man. Animal studies, physical model experiments, and analytical model simulations were used to help the understanding of the kinematics of the head associated with DAI. According to Margulies and Thibault, the physical models of anatomically correct geometry allowed them to develop a relationship between the dangerous inertial loads of the animal studies and the magnitude of the resulting intracranial deformations in a surrogate brain. This relationship was used to scale the DAI tolerance for baboons to the tolerance for humans.

For the animal model, and to simulate inertial loading, the heads of anaesthetised baboons were moved suddenly in a rotational manner without impact. Lesions were often located in the deep white matter of the cerebral hemispheres. Margulies and Thibault note that DAI was produced most readily by lateral noncentroidal rotations of the head. The thresholds for the lateral rotational acceleration and velocity which produced moderate to severe DAI in baboons with a 145 g brain mass were  $100,000 \text{ rad.s}^{-2}$  and  $260 \text{ rad.s}^{-1}$ , respectively.

Assuming baboon and human material properties were the same and had a constant value of critical strain, the tolerance values from the human physical model (brain mass of 1,067 g) were  $16,000 \text{ rad.s}^{-2}$  and  $46.5 \text{ rad.s}^{-1}$ . However, Margulies and Thibault commented that the rotational velocity tolerance value was deliberately deduced from loading below the injury threshold and was therefore overly conservative.

### **3.3.11 Translational Energy Criteria**

The Translational Energy Criteria {Criterion} (TEC) was proposed by Stalnaker et al. (1997). The TEC is based on the Translational Head Injury Model (THIM), which is composed of two masses connected by a spring and damper in parallel. It is modified from the Mean Strain Criterion (Stalnaker et al., 1985) through the addition of a damper in series with the spring, as well as the damper in parallel.

The procedure for validation of the TEC involved a series of impact tests to the heads of six fresh unembalmed PMHS. All impacts were limited to below the fracture tolerance of the skull, thus characterising only the brain contusion portion of the TEC. A total of six impacts were carried out, three lateral and three frontal. A HYGE pneumatic impactor was used in the study, weighing 9.05 kg. The cadaver was positioned so that the 15 cm diameter impacting plate contacted high on the forehead with the line of force directed through the centre of gravity of the head. The impact therefore, contained a superior-inferior component with respect to the anatomical coordinate system, as well as an anterior-posterior component. For one test in the

frontal configuration and one lateral test, a 2.5 cm layer of Ensolite foam was placed in the face of the impactor.

In deriving the TEC, it seems as though Stalnaker et al. try to predict equivalent AIS levels. However, the obvious issue with this approach is the discrete nature of the AIS levels. Treating the data in this manner does not seem appropriate.

### **3.3.12 Maximum Power Index**

Newman et al. (2000) noted that if the exponent of the head acceleration used in the Gadd Severity Index (of 2.5) was set to 2 instead, it would produce a curve closer to that of the Wayne State Tolerance Curve in the 5 to 30 ms duration range. This led Newman et al. to the observation that such a function (as shown in Equation 3-9) would have similar physical correlation to the rate of change in kinetic energy for a rigid object, or power; where  $\Delta v$  is the change in velocity of the head and T is the period of that change.

$$\frac{\Delta v^2}{T} = 6737 \quad \text{Equation 3-9}$$

Based on a general expression for the rate of change of translational and rotational kinetic energy for any rigid object, but setting coefficients equal to the mass and appropriate mass moments of inertia (varying by axis) for the human head, Newman et al. provided an equation for the Head Impact Power. Here the 'a's represent linear acceleration terms (in the three orthogonal axes) and the '\alpha's, rotational acceleration terms.

$$HIP = 4.5a_x \int a_x dt + 4.5a_y \int a_y dt + 4.5a_z \int a_z dt + 0.016\alpha_x \int \alpha_x dt + 0.024\alpha_y \int \alpha_y dt + 0.022\alpha_z \int \alpha_z dt$$

However, Newman et al. continued to discuss whether the different tolerance to power absorption of the head in different directions needed inclusion. They suggested that the Maximum Power Index would need to include additional coefficients reflecting directional tolerance. An investigation of American football head impact cases in their mild traumatic brain injury database supported the authors' assertions regarding directional tolerance. It also seemed to support the conclusion that the maximum

head impact power appeared to correlate better than existing head injury assessment functions (i.e. HIC) with the mild traumatic brain injury data. They therefore proposed the maximum head impact power  $HIP_m$  as a new head injury assessment function, though they did not propose the directional tolerance coefficients. As such, this criterion still needs validation before it can be used more widely.

### **3.3.13 Skull Fracture Correlate**

Vander Vorst et al. (2003) commented that:

*"The growing understanding of head injuries indicates that a comprehensive assessment of impact-induced injuries involves many modes, each requiring a specific criterion."*

Their research then sought to address one mode of head injury only, impact-induced linear skull fracture. The Skull Fracture Correlate (SFC) is the average acceleration evaluated over the HIC interval. It was developed to relate skull fracture with the tensile strain in the outer table of the cranial bone. Fracture occurs when the ultimate strain is exceeded. However, skull strain data at the location of fracture is difficult to measure in an impact test and is not available for historical test data. This is why a SFC is needed to bridge between impact event measurements (head acceleration) and injury outcome.

A database of PMHS skull fracture outcomes with corresponding risk factors from Hybrid III headform drop tests was constructed by Vander Vorst et al. for statistical analysis. Data from 30 PMHS tests at the Medical College of Wisconsin and 76 tests selected from the open literature were entered into the database. Vander Vorst et al. found that the headform change in velocity over the time period used in the HIC calculation (exact period not specified by the authors) was the best correlate to skull fracture. This parameter was then defined as the SFC.

For a 15 percent, or less, probability of skull fracture the criterion is  $SFC < 120 g$  (95 percent confidence interval:  $88 < SFC < 135 g$ ).

### 3.3.14 Combined probability of concussion

The study by Rowson and Duma (2013) introduced a new injury metric which considered both linear and rotational head acceleration. The metric was derived using a multivariate logistic regression analysis of American football head impact data obtained using the Head Impact Telemetry System (HITS) for instrumenting helmets. The dataset consisted of peak linear and rotational accelerations for 62,974 sub-concussive events and 37 impacts where concussion was diagnosed. However, Rowson and Duma also make an adjustment to account for underreporting.

Based on this data and using a generalised linear model, the combined probability of concussion was defined by the following equation (Equation 3-10).

$$CP = \frac{1}{1+e^{-(\beta_0+\beta_1a+\beta_2\alpha+\beta_3a\alpha)}} \quad \text{Equation 3-10}$$

Where  $\beta_0$ ,  $\beta_1$ , and  $\beta_2$  are regression coefficients,  $a$  is peak linear acceleration,  $\alpha$  is peak rotational acceleration, and  $CP$  is the combined probability of concussion. The regression coefficients for the combined probability of concussion equation were determined giving the values ( $\beta_0 = 210.2$ ,  $\beta_1 = 0.0433$ ,  $\beta_2 = 0.000873$  and  $b_3 = -0.000000920$ ) as shown in Equation 3-10.

$$CP = \frac{1}{1+e^{-(-10.2+0.0433a+0.000873\alpha-0.000000920a\alpha)}} \quad \text{Equation 3-11}$$

### 3.3.15 Strain rate

Concussive events that occurred during US American football (National Football League) games were quantified and duplicated in the laboratory using helmeted dummies. Linear and angular accelerations measured during the reconstructions were used as inputs into the finite element Wayne State University Head Injury Model (WSUHIM). A variety of brain response parameters were computed for both the concussed and non-concussed players. A total of 53 cases were studied of which there were 22 cases of concussion, as diagnosed by the team physician on site. Strain rate was manually calculated by differentiating the maximum principal strain versus time curves for elements with the highest values of strain. The rate varied from 23 to 140  $s^{-1}$  with an average value of 84  $s^{-1}$  for injury cases and from 11 to 67  $s^{-1}$  with an

average value of  $38 \text{ s}^{-1}$  for non-injury cases. The product of strain and strain rate was also suggested as a local tissue response measure that could be a mechanical parameter for neural injury.

Based on values from three significance tests, the product of strain and strain rate at the midbrain region provided the strongest correlation with the occurrence of mild traumatic brain injury in the WSUHIM. Strain rate was also a good injury predictor in this model.

### **3.3.16 Strain**

In mathematics there tend to be two ways of expressing strain: in natural (or Eulerian) terms it is the instantaneous change in length divided by the instantaneous length; however, the often more familiar expression is known as the Lagrangian strain, and it is the difference between the current and original length, divided by the original length. Morrison et al. (2003) applied mechanical injuries to organotypic hippocampal slice cultures and quantified the resultant cell death. They concluded that:

- Cell injury is dependent on the magnitude and rate of application of tissue deformation
- Mechanical deformations  $\leq 0.1$  Lagrangian strain are not injurious when applied at strain rates between  $5$  and  $50 \text{ s}^{-1}$
- Mechanical deformations  $\geq 0.2$  Lagrangian strain induce significant levels of cell injury, noting that the time course for the damage was dependent on the strain rate of the applied deformation.

More recently, Sahoo et al. (2016) used a state-of-the-art finite element head model to look at diffuse axonal injury predictions using a variety of predictive measures. The study used the Strasbourg University Finite Element Head Model after it had been enhanced using new constitutive material laws for the brain and the skull. Importantly, the brain model was improved by implementing anisotropy based on fractional anisotropy and fibre orientation extracted from medical imaging (diffuse

tensor imaging). With this head model, 109 head trauma cases were simulated. This included 70 pedestrian cases and 39 cases from other impact types. In total 27 % of the casualties sustained a diffuse axonal injury and 73 % did not.

Sahoo et al. computed nine intracerebral mechanical parameters as potential predictors of a diffuse axonal injury outcome: brain axonal strain rate, axonal strain, first principal strain, von Mises strain, first principal stress, von Mises stress and cumulative strain damage measure (CSDM; as described in the following section, and at three different levels, 0.1, 0.15 and 0.2). Based on statistical analysis, axonal strain had the highest correlation with correct prediction of diffuse axonal injury. The proposed tolerance limit for a 50 % risk of diffuse axonal injury is 14.65 % of axonal strain. As noted by Sahoo et al. this agrees with other axonal strain limits proposed in the literature.

### **3.3.17 SIMon injury metrics**

The Simulated Injury Monitor (SIMon) finite element head model was developed by NHTSA to predict acceleration-induced traumatic brain injury (Takhounts et al., 2003).

#### *3.3.17.1 Cumulative strain damage measure*

Three injury metrics are calculated by SIMon; the first of these is the Cumulative Strain Damage Measure (CSDM), a correlate for diffuse axonal injury. The Cumulative damage strain measure was first proposed by Bandak and Eppinger (1994). They suggested that, "a mechanical measure to evaluate the extent and severity of strain-related damage within the brain..." would be useful in the evaluation of deformation-related brain injuries. This measure was based on an association between DAI and the cumulative volume of brain matter experiencing tensile strains exceeding a critical level at some point during an impact event. The measure therefore calculates the volume of model elements that experienced a strain above a prescribed threshold value for each time increment and gives a maximum cumulative value after the strain-causing event.

To select the critical values of strain and volume for the CSDM injury metric, data from animal experiments were used to relate the CSDM levels to the observed occurrence of DAI. Takhounts et al. (2011) produced a risk curve for predicting AIS  $\geq 4$  brain injury or diffuse axonal injury based on CSDM which was generated via the animal experiment data. The 95 % confidence intervals are very wide below a CSDM of about 0.5 and a 50 % risk of this level of injury.

#### *3.3.17.2 Dilational damage measure*

The second injury metric documented with the SIMon head model is the Dilational Damage Measure (DDM); as an estimate of the potential for contusions. This measure assesses regions where stress states in the brain model lead to negative pressures. The concept for this criterion is that negative pressures may reach values large enough to cause contusions and tissue damage (as may be found with contre-coup injuries {opposite to the side of the head being loaded directly}). The DDM monitors the volume of the brain exceeding a threshold level in a way similar to the CSDM calculation. The result from the measure is the percentage of the brain volume that has exceeded that specified threshold value. The threshold pressure was defined as -100 KPa by Takhounts et al. (2003) to align with the vapour pressure of water (the pressure at which steam is saturated). Presumably, it is envisaged that at lower pressures bubbles can form leading to the risk of cavitation.

#### *3.3.17.3 Relative motion damage measure*

The third of the three injury metrics calculated with the SIMon head model is the Relative Motion Damage Measure (RMDM), which is a correlate for acute subdural haematoma. This measure is used to evaluate the motion of the brain with respect to the interior surface of the cranium. Fourteen pairs of nodes on the cranium and brain surfaces are tracked throughout the loading event to quantify the tangential motion between the brain and skull. The strain and strain rate is then compared against experimental data from Löwenhielm (1974).

### 3.3.18 RIC

Kimpara and Iwamoto (2012) proposed two new criteria for predicting traumatic brain injury based on angular accelerations. The first of these was called the Rotational Injury Criterion (RIC). It uses the same formulation as for HIC, but with the resultant linear acceleration substituted for the resultant angular acceleration, as shown in Equation 3-12.

$$RIC = \left[ \frac{\int_{t_1}^{t_2} \alpha \cdot dt}{t_2 - t_1} \right]^{2.5} (t_2 - t_1) \quad \text{Equation 3-12}$$

### 3.3.19 PRHIC

The second new criterion proposed by Kimpara and Iwamoto (2012) was called the Power Rotational Head Injury Criterion (PRHIC). This started with the rotational elements of the HIP, as shown in Equation 3-13.

$$HIP_{rot} = \sum I_{ii} \alpha_i \int \alpha_i dt \quad \text{Equation 3-13}$$

Where,  $I_{ii}$  is moment of inertia and  $\alpha_i$  is angular acceleration. For a mid-sized male, Newman had already proposed MOI values for the x, y and z axes of 0.016, 0.024 and 0.022 kg.m<sup>-2</sup>, respectively.

Just as for RIC, the PRHIC substitutes out the linear acceleration of the head and replaces that with the  $HIP_{rot}$  (Equation 3-14).

$$PRHIC = \left[ \frac{\int_{t_1}^{t_2} HIP_{rot} dt}{t_2 - t_1} \right]^{2.5} (t_2 - t_1) \quad \text{Equation 3-14}$$

Kimpara and Iwamoto found that the time durations for angular acceleration obtained from American football head impact data were greater than 15 milliseconds. Therefore, they proposed that the maximum time integral duration used with RIC and PRHIC should be set to 36 ms. The new criteria therefore became RIC<sub>36</sub> and PRHIC<sub>36</sub>.

The validation for RIC and PRHIC came from using concussive and non-concussive head acceleration data obtained from American football head impacts. These were taken as inputs for a human brain finite element model which predicted the CSDM and hence some estimate of the risk of diffuse axonal injury. The results identified that

RIC<sub>36</sub> was significantly correlated with CSDM 10 %, a potential indicator of mild traumatic brain injuries such as concussion. PRHIC<sub>36</sub> was correlated with CSDM 30 %, which may predict severe traumatic brain injuries. Therefore, Kimpara and Iwamoto recommended that RIC<sub>36</sub> and PRHIC<sub>36</sub> be used as the different injury predictors for mild and severe traumatic brain injuries, respectively.

### 3.3.20 BrIC

The Brain rotational Injury Criterion (BrIC) was proposed by Takhounts et al. (2011). They ran crash test dummy tests with 50<sup>th</sup> percentile male dummies to generate head kinematic data from crash events. They then used the outputs from the head instrumentation of the dummies as inputs to the SIMon finite element model to produce a CSDM value for each test. The critical values for the BRIC equation (Equation 3-15) were then chosen as design variables to optimise the linear correlation between CSDM and BrIC. Another constraint was that the BrIC had to be equal to 1 when CSDM was equal to 0.425. This was set to indicate a 30 % probability of diffuse axonal injury (or an AIS ≥ 4 injury) occurring.

$$BrIC = \frac{\omega_{max}}{\omega_{cr}} + \frac{\alpha_{max}}{\alpha_{cr}} \quad \text{Equation 3-15}$$

In 2013, Takhounts et al. published a revision to BrIC that incorporated some major modifications. Takhounts et al. determined that BrIC correlated well with CSDM and the maximum principal strain, but that it was the angular velocity and not angular acceleration which was important for injury prediction. Therefore the angular acceleration term was dropped from the criterion. They also noted that the critical values for angular velocity are directionally dependent, and are independent of the crash test dummy used for measuring them. Therefore the formulation of the criterion became that shown in Equation 3-16. The suggested critical values for maximum angular velocity about the x, y and z axes are 66.25, 56.45 and 42.87 rad.s<sup>-1</sup>, respectively.

$$BrIC = \sqrt{\left(\frac{\omega_x}{\omega_{xC}}\right)^2 + \left(\frac{\omega_y}{\omega_{yC}}\right)^2 + \left(\frac{\omega_z}{\omega_{zC}}\right)^2} \quad \text{Equation 3-16}$$

### 3.4 Confounding factors

One of the recurring issues in brain injury research is the transferability of results obtained with animal subjects to the human case. Even in 1963, alongside the presentation of their tolerance curve, Gurdjian et al. noted factors which were likely to affect significantly the findings from animal experimentation:

1. Differences in the size of the brains may be important with regard to the production of concussion
2. The use of anaesthesia may affect the results obtained
3. The different configurations of the skull in different animals and in the human may be of importance with regard to production of concussion

According to Stalnaker et al. (1975), "Headache, dizziness and concussion play an important role in determining the AIS code for accident victims, but it is difficult to get this kind of information from experiments on non-human primates and human cadavers."

*"... the difference between AIS 2 and AIS 3 for "cerebral concussion with or without skull fracture," is that the former is assigned when unconsciousness lasts for less than 15 minutes and the latter for more than fifteen minutes ... it is very difficult to accurately determine the state of consciousness in anesthetized animals. As such, it is quite easy for the investigator to give an AIS 2 rating when it should actually be 3. Similar arguments can be made for any injury level ... This would be even more the case for human cadaver experiments." (Stalnaker et al., 1975)*

The appropriateness of using PMHSs as surrogates for humans in head injury research (particularly into brain injuries) is often questioned. However, through tests where a mass was dropped onto the exposed brain in both living and unembalmed cadaveric canine specimens, Smith (1979) investigated whether living and cadaveric cerebral vessels had a similar propensity to rupture. Smith's study demonstrated that, given the artificial repressurisation parameters employed and the intravascular marker used, the post-mortem cerebral vessels were able to reflect injury in a fashion similar to the living vessel.

However, Nusholtz et al. (1979) reported that, "during the contact time of direct impact, the motion of the brain of the unembalmed cadaver can be only partially constrained by the skull; the degree of constraint can depend on the time after death and the preparation of the cadaver. This partial de-coupling may have marked effects on kinematic time history of the head during and following an impact."

Other confounding factors include:

- Impact conditions
  - Impact velocity, effective masses, energy, etc. will have a bearing on the injury outcome for a head impact. These parameters can vary readily from head drop tests to head strikes from a pendulum or linear impactor.
  - Impact angle (whether directed through the head centre of gravity, or a glancing blow) will lead to varying levels of head linear or angular motion. The balance between the two types of motion is often considered as being important, particularly when investigating a particular type of head injury.
  - Thibault and Gennarelli (1990) made an important comment regarding the relation between existing head injury criteria and soft tissue head (brain) injuries:

- *"In the complex environment of the automotive collision the head of an occupant can experience forces which are applied with variation in the following parameters: direction, magnitude, waveshape, and effective anatomic location." Thibault and Gennarelli (1990)*
- The side of the head struck
  - Skull thickness varies around the cranium and therefore one expects tolerance to vary by region of the head impacted (e.g. as observed by Hodgson and Thomas, 1973). The skull thickness will also affect stiffness of the head response and therefore the ability of the head to absorb the energy of the impact.
  - It should be noted that, as determined with small milled bone test specimens (1.1 x 2.5 x 11.1 mm), skull bone does not vary in tensile breaking stress, breaking strain, or energy absorbed to failure based on age, side of body or bone type (Wood, 1971).
  - In his work with monkey subjects, Ommaya (1966) observed that:
    - *"It did not appear that the actual site of impact on the head was a crucial factor in the production of concussion for occipital and frontal blows. The important factor was the efficiency of impact {presumably the transfer of momentum to the head of the subject}... The site of impact was crucial, however, for the production of skull fracture. Thus blows to the vertex produced fractures most easily, while blows to the occipital region (inion<sup>23</sup>) were least likely to do so. The frontal and temporal regions were intermediate between the other regions in this regard." Ommaya (1966)*

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<sup>23</sup> The most prominent projection of the occipital bone low down at the back (posteroinferior) of the skull.

- Guskiewicz and Mihalik (2011) speculated that top-of-helmet impacts might result in a coup-contrecoup mechanism occurring in a superior-to-inferior direction, causing the cerebellum to impact the base of the skull and recoil superiorly into the cerebellar tentorium. Interestingly, their data indicated that top-of-helmet impacts typically result in relatively lower rotational acceleration values compared with injuries after impacts to the other areas of the head. However, these impacts to the top are at least six times more likely to result in impact magnitudes with a peak linear acceleration greater than 80 g than side or front impacts.
- *"These findings bring into question the notion that rotational acceleration is the leading precursor to injury and are suggestive that the type of acceleration, in combination with impact location, may be a better determinant for both onset and severity of injury."* (Guskiewicz and Mihalik, 2011)
- Impact surface shape
  - Different shapes of impacted object (e.g. flat plate, cylinder, kerb-like edge) will produce different impact pressures for a given impact force. This will influence the propensity of the skull to fracture under a particular force and may also contribute to other aspects of the head motion; for instance, the energy absorbed by the head and the rotation induced by the impact.
- Impact surface stiffness
  - The rigidity of the impact surface will control the amount of the impact energy absorbed by the surface and transmitted to the head. It will also determine the duration of the impact and will influence the area of contact to a certain degree.

- The direction of induced head motion
  - The skull and the brain inside the skull are not spherical. As a result one would expect the stresses and strains caused by motion in one particular direction (about one axis) to be different to those caused by an equivalent motion in another direction or axis.
  - *"The location of the damaged axons in DAI determines the specific neurological sequella of the injury and are very dependent on the direction that the head moves during injury."* (Gennarelli, 1985)
  - Testing using anaesthetised monkey subjects was used by Gennarelli et al. (1987) to demonstrate that the direction of brain motion is important in the amount of axonal brain damage produced by inertial loading. They reported that for comparable angular acceleration and velocity levels, the brain is most susceptible to axonal damage in coronal plane acceleration (see the Glossary for a description of the anatomical planes), while horizontal and sagittal plane accelerations produce less damage.
  - *"The direction of the head motion has a clear effect on the resulting brain injury patterns and neurological deficits, thereby suggesting that these factors need to be considered in man."* (Miller et al., 1998)
  - In their finite element analyses, Zhou et al. (1996) found that because of the moments of inertia about the head, the same loading would result in higher shear stresses in the brain in lateral rotation than in sagittal rotation. This finding supports the assertions made by Gennarelli et al. (1987). However, Zhou et al. note that the maximum shear stress in the genu<sup>24</sup> is still higher in sagittal rotation.

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<sup>24</sup> The anterior end of the corpus callosum

- In their finite element analysis, Bandak and Eppinger (1994) found that anterior-posterior rotations appeared to be somewhat more severe for their model in that they resulted in higher values of cumulative damage than medial-lateral rotations.
- However, in their recent finite element analysis, Ghajari and Sharp (2015) found that, based on axial stretch of white matter tracts, the white matter was more vulnerable to lateral impacts than to rear impacts.
- Also recently, in their finite element analysis Zhao and Ji (2015) found the following regional sensitivities:
  - Whole-brain and cerebrum; largest regional average peak strains in rotations about the vertical axis
  - Brainstem; lowest response in rotations about the vertical axis, largest average peak strains in sagittal rotations
  - Corpus callosum; largest average peak strains in a slightly oblique coronal rotation.
- Furthermore, some proposed mechanisms of brain injury depend on relative movement between the brain and the skull. The brain-skull interface differs around the head with variable smoothness or protrusions to the internal surface of the skull and varying depths of separating tissues and fluids between the brain and skull. These differences are likely to provide some influence on the ability of the brain to move with respect to the skull and therefore the resulting strains created in the tissue.
- Based on their finite element numerical investigation of the relative movement of the brain with respect to the skull and strain in bridging veins, Kleiven and von Holst (2001) observed greater motion between the brain and the skull in occipital impacts compared with frontal or temporal impacts. This led to the following comments:

*"The larger amount of motion between the brain and the skull for the occipital impact compared to results of the frontal impact might also be explained by the anatomical difference between the frontal and occipital region of the skull. Following a frontal impact the sharp edges of the sphenoidal bone between the middle fossa and the anterior skull base, as well as the anterior skull base itself acts as a restriction of the motion. For the corresponding occipital impact the main support is the tentorium with its compliant properties. In the same way, when enduring a temporal impact, the lower values of relative motion compared to the occipital impacts could probably be explained by the supportive properties of the falx membrane."*

- *Zhou et al. also used numerical simulations to make the following comments regarding the incidence of subdural haematomas.*

*"The low incidence of subdural haematomas in vehicular accidents may be due to the fact that frontal impacts predominate. The importance of the impact direction in causing subdural haematoma was demonstrated in these simulations." Zhou et al. (1995).*

- **Helmet wearing**
  - Helmets are generally considered to be beneficial in reducing the likelihood of head injury for a given impact (Javouhey et al., 2006). Interestingly, helmets are principally designed to reduce linear accelerations and distribute forces over the head. However, the benefit of wearing a helmet is often attributed to a reduction in concussion (e.g. in American football); an injury thought, by some, to be caused primarily by rotational head motion (although as cited earlier, Hodgson et al. (1969) reported that it was likely for translational acceleration to be the most important mechanism in concussion). On this basis, King et al. (2003) concluded that, "the mechanism of head injury may not be linked to rotational acceleration as strongly as that suggested by early researchers." This led them to question, "Just how does the helmet

protect the brain if the prevailing thought is that angular acceleration is responsible for brain injury?”

- Pressurisation
  - Some testing with PMHS has been conducted without any vascular pressurisation. Other authors noted that the vascular system may be more prone to injury when pressurised as in the living human. Therefore, in much PMHS test work vascular pressurisation is performed. This is usually carried out by pumping fluid into the arterial system through a catheter. The success of this strategy is not always guaranteed. As such, the pressurisation of the vascular system will fall in the range from none, to very good (life-like pressure).
  
- Preservation
  - Following death, PMHSs can be treated in different ways in order to keep them relatively fresh for test work. Traditionally some preservation techniques involved embalming the PMHS; however, the embalming process causes changes to the behaviour of the biological tissue. Therefore, differences in response can be expected between embalmed and living tissue. In recent PMHS work it is more often the case that the PMHS will be refrigerated or frozen until being brought back to room temperature shortly before testing. This avoids gross chemical changes in the tissue (other than the cell structure changes caused by freezing), though there could be material property differences where laboratory temperature cannot be brought up to body temperature.
  
- Age
  - Biomechanical properties of organic tissues vary with age. This tends to affect properties such as moduli of elasticity and failure strength, which can be fundamental in the biomechanical response to impact.
  - Increasing age is associated with poor outcome following traumatic brain injury (Shukla and Devi, 2010).

- Kleiven and von Holst (2001) explained an observed increased risk of subdural haematoma in elderly people as being due to the reduced brain size. Through finite element modelling, they showed reduced brain size to be associated with larger relative motion between the skull and the brain resulting in distension of bridging veins.
- Soysal et al. (2005) added that, "The elderly are predisposed to bleeding because normal atrophy<sup>25</sup> related to aging occurs, stretching the bridging veins from the dura. These stretched veins damage more easily."
- Gender
  - Gender seems to have a less pronounced effect on biomechanical properties than age. However, it does tend to govern body shape and size, etc. to a certain degree.
  - After evaluating multiple years of concussion incidence data in comparable sports, Dick (2009) reported that the evidence indicated that female athletes may be at greater risk of concussion than their male counterparts.
- Size
  - The size of a head will have an effect on its tolerance to applied load. The radius of curvature for the skull will affect its ability to deform during an impact (affecting stiffness and energy absorption) and also the stress concentration during bending (affecting failure load). Also, the relative balance between the mass of the brain and skull will change with head size and is likely to play a part in the head's tolerance to applied loading. A proportionally smaller brain will have greater contact/edge effects per unit size, whereas an absolutely smaller brain will have less inertial mass than a larger brain. These features will tend

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<sup>25</sup> The physiological process of wasting away, through the breakdown and reabsorption of tissue.

to be protective for smaller heads, as is shown in differences between humans and smaller primates (ignoring species-based differences).

- Other individual variations (Guskiewicz and Mihalik, 2011)
  - Cerebrospinal fluid levels and function
  - Vulnerability to brain tissue injury
  - Relative musculoskeletal strengths and weaknesses
- Anticipation of an oncoming event (Guskiewicz and Mihalik, 2011)
- Measurement tools
  - Typically, the impact severity for the head is assessed based on instrumentation attached to the skull. For example, pressure transducers, angular rate sensors and accelerometers have been used in this way.
  - The applied loading to the head as measured by the instrumentation will be sensitive to impact location. Measured HIC might vary from the centre of gravity to the non-struck side of the head by a factor of three (Got et al., 1978). It is therefore quite clear that similar measurement positions must be compared between studies. Dummies tend to measure accelerations close to the head centre of gravity. On this basis it is important to consider how the subject-based measurements can be related to the dummy measurements.
  - How rigidly the instrumentation is attached to the head will affect the natural damping of the measurement. This should be kept in mind, particularly as mounting of instrumentation to test subjects is not always ideal.

- Following the recording of a measured signal, post-processing will be carried out to remove unwanted signal artefacts (e.g. high frequency noise). The filtering techniques commonly used in biomechanical applications have changed over the years. Early studies often used heavy (low band pass) filters, which by today's standards are harsh. They can substantially alter reported peak values. Whilst rigorous reporting of studies will include a description of the filtering process used, this is not always the case for the literature documenting head injury testing.
- In reconstructions of head impacts, a dummy head or headform is used typically. The ability of the headform to deform as a human skull would, during the impact will influence strongly the measured loading to the head.



## 4 Comments on existing criterion effectiveness

*"Research on the influence of various biomechanical factors for predicting outcomes after sport-related concussion is inconclusive, but new technologies may lead us to more answers."* (Guskiewicz and Mihalik, 2011)

### 4.1 APROSYS Project findings

Within the APROSYS project (EC Integrated Project on Advanced PROtection SYStems), Deck et al. (2007) used two numerical head models to assess the injury prediction capabilities of different head injury criteria. Head impact conditions that occurred in 68 motor sport, motorcyclist, American football and pedestrian accidents were re-constructed with a state of art finite element human head model (from the Université Louis Pasteur, Strasbourg) and a simplified head model (the NHTSA SIMon model). The input parameters for the head models were obtained from reconstructing each of the accidents using physical test tools. It should be noted that the dummy heads used in the reconstructions have certain limitations when used in this respect. Of primary importance is that the headforms are non-frangible, unlike a human head. Typically a dummy head, such as the Hybrid III head will have a relatively simple, hairless, rubber skin fitted over a metal skull structure; avoiding many of the complexities associated with a human head.

Deck et al. then carried out statistical analysis on the head loading parameters from the accidents (e.g. peak linear and rotational acceleration of the head) and predictions from the head model (e.g. von Mises stress<sup>26</sup> and pressure in the brain). This led to the determination as to which of the investigated parameters provided the most accurate predictor of the injuries sustained in the accidents. The recorded injuries from the analysis were mild or severe diffuse axonal injury (DAI), sub-dural haematoma, and skull fracture.

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<sup>26</sup> The von Mises - Hencky criterion is a formula for calculating whether the stress combination at a given point will cause failure. Three orthogonal stresses are combined into an equivalent stress (index number), which is then compared to the yield stress of the material. If the "von Mises stress" exceeds the yield stress, then the material is considered to be at the failure condition (definition from various internet sources).

Within the head impact cases considered by Deck et al., helmet wearing was found to have a significant influence on severe DAI and skull fracture. Based on the odds ratio of persons with or without that injury type, the risk of severe DAI was four times greater for a non-helmet wearer than a helmet wearer; and the risk of a skull fracture was 28 times greater for a non-helmet wearer than a helmet wearer. However, as indicated by Deck et al., this result depends strongly upon the accident data set used in the analyses, and the similarities and differences between helmet and non-helmet cases. Once helmet wearing was taken into account, the NHTSA head model parameter 'peak negative brain pressure' had the highest  $r^2$  value for both severe DAI and skull fracture.

Deck et al. concluded that there was no significant benefit in using either a simplified or a state of the art head FE model to predict severe head injury. The models introduced substantial additional complexity and were not better than HIC or the input linear acceleration in predicting the presence of absence of severe head injury. They recommend that the models are not beneficial for regulatory testing as long as severe ( $AIS \geq 3$ ) injuries are concerned. However, if injury data from a specific type of accident are being analysed, or if a specific mild head injury is being investigated, use of a head FE model may be of benefit.

## **4.2 New analysis of published data**

To investigate the effectiveness of the existing criteria further, a database of head impact cases was constructed by the author based on the information reported in the literature. This database collated information from a variety of historical references and included an assortment of impact types, impact directions, contact surfaces, species (human or monkey), etc. These diverse sources of head impact data were considered to include a broad range of factors that may affect the ability of certain measured criteria, or parameters, to predict head injuries. Research was chosen to include literature sources where the head loading environment was described in detail, although the loading measurements were not necessarily reported in a consistent

manner. There were 630 documented cases of head impact testing collated into the database used for analysis.

A report describing the analysis has already been published (Carroll, 2010). Some of the main results are reproduced below.

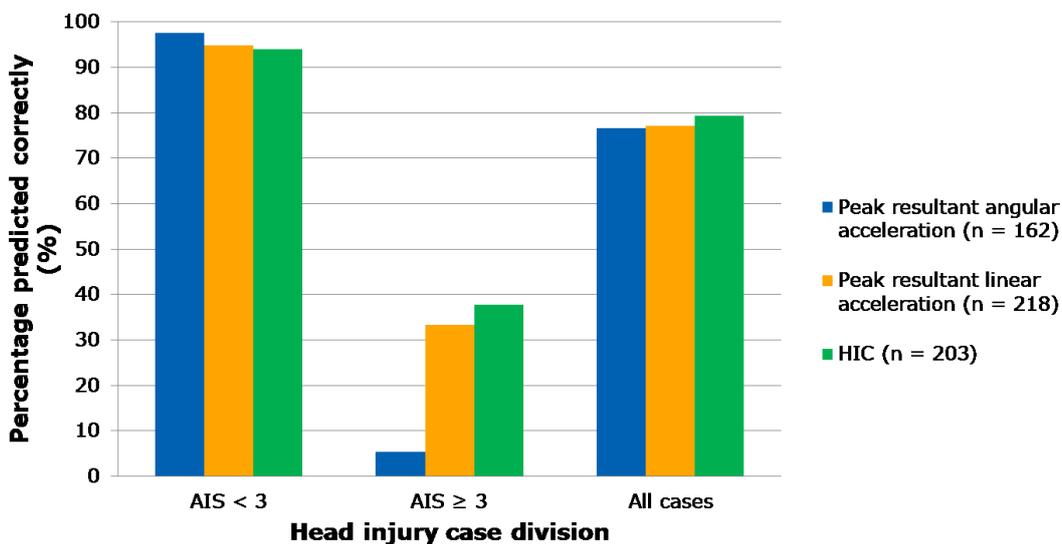
For 431 cases, the head injury was reported in sufficient detail to determine if it was severe in nature (using an Abbreviated Injury Scale (AIS; AAAM, 2008 {and former revisions}) score of at least 3). For 277 of the cases that were coded as being either  $AIS < 3$  or  $\geq 3$ , a peak resultant acceleration value was also reported (this included cases from boxing, data from the EC sixth framework Aprosys Project, pedestrian accident reconstructions, constrained/forced head motion tests, impactor tests, and full-body drop tests). Based on logistic regression of these 277 cases, the resultant acceleration was shown to be a statistically significant predictor of severe head injury. However, whilst this could predict correctly the absence of a severe head injury in over 70 percent of those events (where most resulted in a non-severe injury or no injury), it could only predict severe head injury in less than half of the severe head injury cases.

A HIC value was reported for 240 of the 431 cases for which an AIS coding was possible (the cases came from the following sources: boxing, the Aprosys Project, pedestrian accident reconstructions, impactor tests, and full-body drop tests). Again logistic regression showed that HIC was a significant predictor of severe head injury. However, this significance was again slightly misleading; as whilst it is true that the HIC could be used to predict the correct result in over 75 percent of the cases, over 80 percent of the severe head injury cases would have been predicted as false negatives.

An obvious issue that was noticed with the conglomeration of head injury cases was that the monkey subjects had been exposed to much higher acceleration levels than the human subjects. Therefore the logistic regression analyses were repeated looking at human subject cases only.

For the human subjects (data from PMHS tests as well as accident / accident reconstruction data) the peak resultant linear acceleration was again significant in predicting the severity of head injury. Again, whilst it correctly predicted an AIS < 3 result in almost 95 percent of those cases, the AIS ≥ 3 result was correct for only about one third of the severe head injury cases. The HIC gave only a very slightly improved predictive ability over the raw resultant linear acceleration. Despite predicting more of the AIS < 3 cases correctly, the peak resultant angular acceleration was the measure least able to predict the AIS ≥ 3 injury outcome correctly.

A graphical comparison of the predictive capabilities of the peak resultant linear acceleration, resultant angular acceleration, and HIC is shown in Figure 4-1, for the human head impact cases.



**Figure 4-1: Percentage of human head impact cases where the outcome was predicted correctly by the linear or angular accelerations, or HIC**

Of the human cases, 162 had both an AIS coded head injury and a peak resultant angular acceleration value (the cases came from the following sources: Aprosys Project, boxing, pedestrian accident reconstructions, impactor tests, and full-body drop tests). For the 125 cases of AIS < 3 head insult the rotational acceleration value could be used to predict that result correctly for all but three of these (almost 98 percent). However, in only two of the 37 AIS ≥ 3 cases would the rotational acceleration have given the correct result, instead suggesting AIS < 3.

One criticism raised regarding the use of the HIC is that skull fracture can affect heavily the measured peak linear acceleration and HIC value. A fracture will allow greater deformation of the skull than would occur without a skull fracture, thereby attenuating the acceleration under equivalent impact conditions. Taylor (1967) cited the work of Pott and Earle (1808) who had observed that a skull fracture might protect a person against the general effects of brain deceleration. Taylor commented that,

*"A great deal of energy is absorbed in fracturing the skull, which is then diverted from the production of shearing stresses in the brain... One expects the patient with a fractured skull to suffer less from concussion effects than his unfractured counterpart who absorbed the same amount of energy on impact." (Taylor, 1967)*

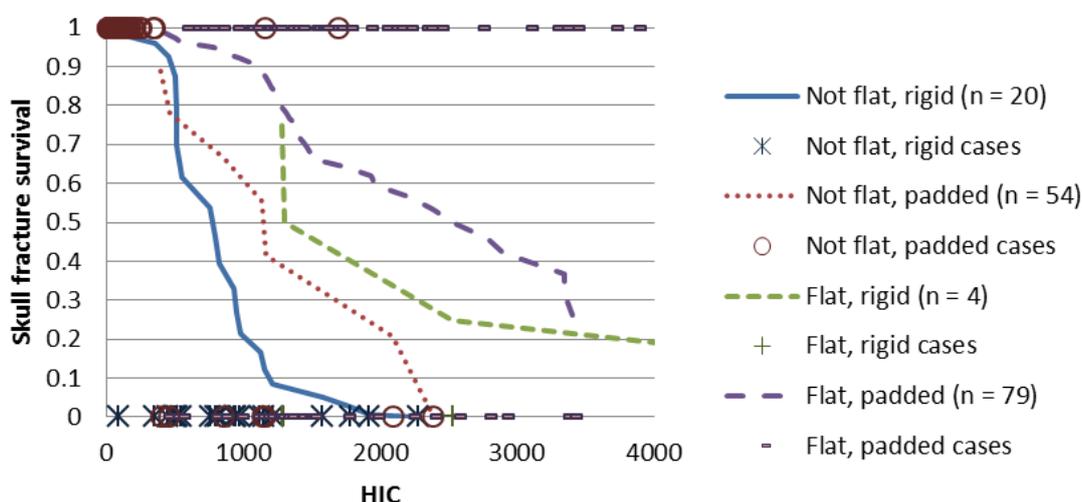
To investigate this effect in the published data, the presence or lack of a skull fracture was included as a covariate in the logistic regression of HIC with AIS  $\geq 3$  head injury (the sources of data for these cases were: boxing, Aprosys Project, pedestrian accident reconstructions, impactor tests, and full-body drop tests). The results suggested a significant influence of skull fracture, and an improvement in the predictive capability accounting for skull fractures. However, for the no skull fracture cases, the HIC still only predicted one of the nine AIS  $\geq 3$  cases correctly.

The influence of a skull fracture on the ability to predict AIS  $\geq 3$  injuries suggests that one must know the skull fracture outcome for a case before considering the brain injury outcome. This will obviously cause issues when applying HIC to prospective cases (e.g. research testing with a crash test dummy); although, this would not be expected to cause a problem where the head loading was not going to cause a skull fracture, for example in non-contact cases or at sub-fracture loading levels.

The likelihood of a skull fracture occurring for a given HIC value was investigated using a Cox regression survival analysis, as shown in Figure 4-2 (human head impact data were taken from boxing impacts, and full-body drop tests). As is the case for each of the four curves shown (flat or shaped impact surface and padded or rigid impacts), the probability of a head surviving without a fracture decreases with

increasing HIC. Each experimental outcome influences the risk of skull fracture, which is the reason for the many changes in gradient, exhibited by each of the lines.

For the purpose of this investigation, the helmeted conditions used by Got et al. were considered as padded (Got et al., 1978, and Got et al., 1983). From Figure 4-2 it can be seen that there is a protective effect of padding, and impacts with a flat rather than curved surface. This is likely to be due to those two conditions spreading the impact force over a greater area of the skull; assuming that impact pressure or stress is related to the likelihood of fracture. Padding will also help to attenuate impact force, to some extent, through energy absorption.



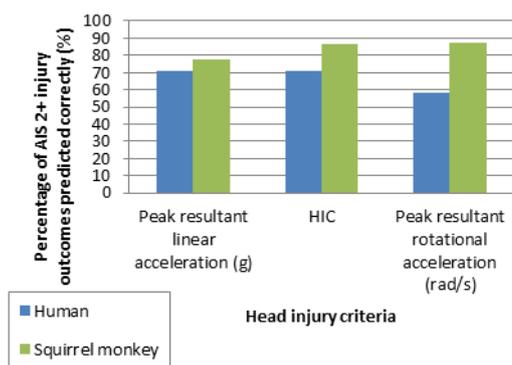
**Figure 4-2: Cox regression plot for probability of avoiding a skull fracture with HIC, for human head impact cases**

Impact site (i.e. frontal, side, rear, etc.) did not significantly affect the ability of the HIC, or peak resultant accelerations to predict AIS  $\geq 3$  head injuries. However, impact site was significant when looking at subsets of the data. For instance, the data of Ono et al. (1980), based on tests with monkey subjects, showed that frontal or rear impact directions were a significant ( $p = 0.02$ ) factor when using the peak resultant linear acceleration to predict AIS  $\geq 3$  head injury.

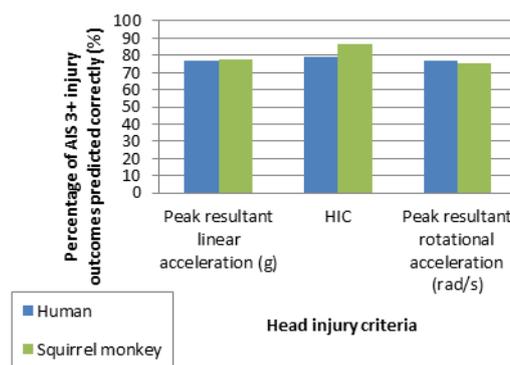
### 4.2.1 Discussion

From the historical data it was not possible to prioritise factors contributing to the head injury risk because not all factors are controlled for, or reported in, all of the studies. This means that when considering all contributing factors together, little, if any, statistical power has been gained through combinations of the data in the literature.

Throughout the analysis of data from the literature, there has been no clear advantage shown when using either peak linear resultant acceleration, HIC, or peak rotational resultant acceleration to predict head injury outcome at the AIS  $\geq 2$  or AIS  $\geq 3$  levels. The following summary figures (Figures 4-3 and 4-4) show the prediction efficacy of the linear or rotational acceleration, or HIC for either AIS  $\geq 2$  or AIS  $\geq 3$  head injuries to humans or squirrel monkeys.



**Figure 4-3: Head injury outcome prediction efficacy for AIS  $\geq 2$  injuries**



**Figure 4-4: Head injury outcome prediction efficacy for AIS  $\geq 3$  injuries**

One of the main limitations with the assessment of injury outcome predictions is the data used for the evaluation. The injury information is only what can be obtained from the reference sources reviewed. Typically, within the historical data the focus has been on the more severe of the spectrum of head injuries. This is particularly true when considering the PMHS impact cases where it is difficult to identify clearly anything but severe brain injuries and skull fractures. Hence from Figure 4-4, the human AIS  $\geq 3$  prediction outcomes are closer to those from the squirrel monkey tests than in Figure

4-3 for AIS  $\geq 2$  head injuries. It should be remembered, at this stage, that relatively simple closed fractures of the skull are AIS 2 injuries. The human dataset used to derive Figure 5 will include many cases of skull fracture (an impact loading injury), whereas the squirrel monkey injuries at the AIS 2 level are more likely to be intracranial injuries, from inertial loading of the head. Rotational acceleration is not expected to be a good predictor of skull fracture. This may explain part of the reason why the prediction of human AIS  $\geq 2$  head injury outcome, based on rotational acceleration, is not as good as the predictive ability shown with the squirrel monkey data.

Based on logistic regression of a subset of cases from the full dataset, the resultant linear acceleration was shown to be a statistically significant predictor of severe head injury. Peak linear head acceleration and the Head Injury Criterion (HIC) were found to be reasonable predictors of serious (Abbreviated Injury Scale, AIS  $\geq 3$ ) head injury occurrence. However several features of the impact conditions were shown to contribute to the injury outcome. Therefore, whilst these criteria seemed useful if applied in a very general way, specific thresholds would have to take account of the confounding factors such as the impact conditions. This supports the real-world observations that using linear head acceleration based criteria to control the loading to the head in a general way is a useful first step in the mitigation of head injuries, but does not account for all the detail that may be necessary in more focussed future safety advances.

It is clear, considering all sorts of biomechanical data, that limiting the force, concentration of force, and the linear and rotational accelerations to which a head is subjected is a sensible approach when trying to reduce injury risk. These basic engineering principles have guided the development of: test tools to assess the severity of head impacts, test procedures, and basic injury criteria for the use in testing potential road-safety related head impacts. Such testing has facilitated the development of injury prevention countermeasures such as padded vehicle interiors, airbags, cycle and motorcycle helmets, etc. As a result, since the implementation of testing, through regulatory or consumer testing schemes, they have driven improvements in head protection levels. These head injury countermeasures have been effective. For instance, the incidence of severe head injuries in frontal impact car crashes has reduced substantially since the widespread introduction of airbags (e.g. Knack et al., 2003).

The proposals to use peak linear head acceleration or HIC to protect against head injuries were based on testing relevant to a historic need, such as the head of a car occupant contacting interior structures. This generated technical solutions which have been successful in addressing particular contact or injury types (e.g. skull fractures for car occupants). Moving on from that situation, it is likely that head injury prevention will need to focus on the remaining injuries and address diffuse injuries occurring away from the point of impact. Based on their derivation and limits of application, peak linear acceleration and HIC do not seem sufficient for that purpose. This may be particularly true for moderate severity injuries or in predicting precisely the type of brain injury. Therefore, additional risk assessment tools are likely to be necessary.

The rotational acceleration components of a head impact are likely to have a greater bearing on intracranial head injury than skull fracture (e.g. Gennarelli et al., 1972). Also, an inference from the APROSYS Project work (Deck et al., 2007) could be that it is important to consider the combined loading to the head (rotational and linear accelerations) when estimating what type of intracranial injury is likely to occur. This is a reasonable assumption, that simple consideration of the linear acceleration of the head (peak resultant value, or HIC) will not provide an accurate prediction of

intracranial head injury occurrence. Consideration of the linear head kinematics alone is unlikely to be sufficient to judge the most likely type and location of potential intracranial head injuries. This is particularly important when considering those injuries with a low threat-to-life, such as concussion and mild diffuse axonal injuries. Reasoning such as this has fuelled, and continues to support, the development of advanced finite element models of the head. However, having an injury criterion based on global head motion and relating to intracranial injuries should still offer a substantial benefit. It would bridge the gap from routine use of the HIC and peak resultant linear acceleration to the time when advanced head models can be used as regular tools for head injury risk testing.

In order to be able to develop advanced head injury criteria, or validate a detailed head model irrefutably, for assessing brain injury risks, it is necessary to have head injury data sources that consider both, not either, rotational and translational head kinematics. Peak acceleration values alone are not sufficient as the timing, duration, and strain rate of the forces to which the head is subjected are also likely to be important. Having full acceleration time-histories is likely to be particularly important given the assumption that combinations of rotational and translational accelerations are necessary for improved injury criteria. The combinations should consider criterion values at each point in time, not just the isolated peak values from any time throughout the acceleration pulse.

As biomechanical research continues to investigate the detailed aspects of head injuries it becomes apparent that the traditional criteria, or indeed any new criterion, would need additional validation (beyond that available from the published data). This validation would have to account for the conditions of loading to the head and the particular head injury under consideration.

#### **4.2.2 Conclusions**

Published head injury research from the past 70 years was reviewed to:

- Collate existing propositions with respect to mechanisms for head injuries
- Draw together a dataset of head injury case data based on published information
- Compare this dataset with expected results based on previously published assertions (making use of statistical techniques for this analysis)

Even with a broad dataset of over 600 head injury cases collated from the literature, it has not been possible to prioritise, according to their importance, the impact factors that contribute to head injury risk. Neither has it been possible to develop an improved estimate of head injury likelihood, which takes into account these contributory factors. One reason for this is that not all factors were documented for all studies conducted previously. Due to the extent of the missing data only limited advantages have been observed through combining the historic data.

Peak head acceleration and the HIC are reasonable predictors of serious ( $AIS \geq 3$ ) head injury occurrence, based on my analysis of historical test data. However several features of the impact conditions have been shown to contribute to the relationship between the injury predictor and the injury outcome. These are:

- Occurrence of a skull fracture
- Impact surface padding
  - Padding affected risk of skull fracture
- Profile of the object impacted
- Impact site
  - Demonstrable with a small subset of animal test data only

To address other diffuse injuries occurring away from the point of impact, additional risk assessment tools are likely to be necessary. Other authors have already proposed the analysis of metrics reflecting the linear and angular motion during an impact event with respect to the occurrence of diffuse injuries.

My analysis of the published head injury case data was not able to show the benefit of including a rotational acceleration component in the injury risk assessment. This was due to a number of reasons:

- Unreliable brain injury detection within previous research
- Inadequate reporting in the literature to determine the occurrence of less life threatening injuries
- Linear and rotational acceleration data not always reported in conjunction with head loading events
- When reported, only peak data values are routinely given in the literature

The need to consider both linear and rotational data in understanding loading to the head during an impact event is not contested. However, my analysis indicates that it is difficult, if not impossible, to assign relative importance to either component with the existing data. Additional and more complete data are needed to define injury risk relationships to criteria isolating particular types of injury and confounding factors.

### 4.3 Summary of current position

In their description of a SIMon (Simulated Injury Monitor) release, Takhounts et al. (2008) gave the following introduction:

*"Many attempts have been made in the past to reduce the occurrence and severity of TBI {Traumatic Brain Injury} as a result of automotive crashes... The process of further improvement of head injury protection systems is limited, however, by the degree of sophistication of currently used head injury criteria... to take the next step forward in protecting automobile occupants from TBI, a better understanding of the physical, biochemical, physiological and biomechanical processes within the traumatically injured brain is necessary (Takhounts et al., 2008)."*

Within the Aprosys project Deck et al. found limited additional benefit of using numerical simulations of the head to predict severe head injury when compared with traditional considerations of the head acceleration and HIC. This finding was based on accident reconstructions where head impacts from a variety of different sources were considered (motorcycle, motorsport, pedestrian, and American football). The accelerations measured during physical reconstructions were used as inputs for the numerical simulations. Clearly, using data from this range of sources will introduce many confounding factors into the prediction of severe head injury. The confounding factors mentioned earlier go some way to explaining the poor result when simulating the head motion numerically over the predictive ability of the accelerations alone. There is also some justification as to why we may not expect a greater predictive ability from the head accelerations.

From the historic data reviewed and re-analysed by me, it has not been possible to prioritise factors contributing to the head injury risk because not all factors are controlled or reported in all of the studies.

Based on logistic regression of a subset of cases from the full dataset, the resultant linear acceleration was shown to be a statistically significant predictor of severe head injury. Peak linear head acceleration and the Head Injury Criterion (HIC) were found to be reasonable predictors of serious (Abbreviated Injury Scale, AIS  $\geq 3$ ) head injury occurrence. However several features of the impact conditions were shown to contribute to the injury outcome. Therefore, whilst these criteria seemed useful if applied in a very general way, specific thresholds would have to take account of the confounding factors such as the impact conditions. This supports the real-world observations that using linear head acceleration based criteria to control the loading to the head in a general way is a useful first step in the mitigation of head injuries.

It is clear, considering all sorts of biomechanical data, that limiting the force, concentration of force, and the linear and rotational accelerations to which a head is subjected is a sensible approach when trying to reduce injury risk.

The proposals to use peak linear head acceleration or HIC to protect against head injuries were based on testing relevant to a specific historic need. Moving on from that situation, it is likely that head injury prevention will need to address diffuse injuries occurring away from the point of impact. Additional risk assessment tools such as an advanced head injury criterion or the use of a numerical head model are likely to be necessary for this purpose.

Consideration of the linear head kinematics alone is unlikely to be sufficient to judge the most likely type and location of potential intracranial head injuries. Reasoning such as this has fuelled, and continues to support, the development of advanced finite element models of the head. Though, having an injury criterion based on global head motion and relating to intracranial injuries should still offer a substantial benefit over the current position.

Until such time as updated injury criteria are available it is recommended that reporting of head impact events needs to consider not only the peak acceleration, but also some aspect of the duration of loading as well. Ideally time-histories should be made available for analysis. Where brevity dominates, then as a minimum, a peak value and duration must be described. Additionally, existing criteria such as HIC are

still a useful reference with respect to general severity of linear acceleration and historic safety countermeasures. Equally, an exceedance value, such as 3 milliseconds, incorporates duration of loading into a single peak acceleration value and could potentially offer additional insight into injurious potential of a loading event. This is the framework for reporting data adopted in the subsequent sections of this thesis, where as far as possible; a peak value, total event duration, HIC and a 3 ms exceedance value have been considered and reported.

In order to be able to develop advanced head injury criteria, or validate a detailed head model irrefutably, for assessing brain injury risks, it is necessary to have head injury data sources that consider both, not either, rotational and translational head kinematics. This investigation has shown that existing data sources are not sufficient for this purpose. It seems as though data from throughout the full impact time history, in the six degrees of freedom, three orthogonal linear and rotational axes) will be necessary to use these criteria effectively. This is essential if the criterion depends upon, or is, the output from a numerical model of the head. New sources of information that contain such complete data together with detailed descriptions (including extent, location, and severity) of any injuries are not readily forthcoming at this time. This view is supported by researchers in the U.S., who after analysing data from six degree-of-freedom (three orthogonal linear and rotational) acceleration data from American football, boxing and mixed martial arts concluded that,

*"... more 6DOF {six degree-of-freedom} data is needed to confirm this evaluation of existing injury criteria, and to develop new criteria that considers directional sensitivity to injury."* (Hernandez et al., 2014)

Until these criteria are developed it is not possible to attribute injury risk in automotive impacts to a particular component of the kinetics or kinematics.

#### **4.3.1 Use of headforms in assessing head injury risk**

One of the key limitations in using a dummy head or headform to assess injury risk in automotive safety testing is that whilst they have some compressible skin element, they are not frangible. In this way they differ from the human case, in which skull fracture can occur. When a skull fractures, the measured acceleration will be attenuated. Therefore, one can expect a systematic capping of head acceleration values in cases of skull fracture at lower levels than in cases without skull fracture. However, when taking measurements using a non-frangible headform such a reduction in acceleration does not occur. This causes inaccuracies when trying to relate the risk of injury based on human acceleration values to those observed in testing with a headform. To address this complication, dummy-specific injury risk functions should be developed. This would involve replicating the injury causing (and non-injurious) conditions with the dummy head or headform in question. The measured acceleration or injury criteria could then be related directly to the injury outcome for the original human subject. By testing with the headform, the effects of skull fracture and any other potential systematic errors (such as the headform biofidelity) would be negated.

## **5 Advances needed**

Based on the literature reviewed and the published head injury case information, evaluation of the predictive ability of each head injury criterion has been severely limited by the level of information provided for each case. This meant that there were often many missing fields when comparing data from one test series with another. The cause of missing data in most cases is simply a result of the authors focussing on one aspect of head injury research at that time.

As such there is no existing possibility, with the data already published in the literature, to provide validation of proposed advanced head injury criteria or head injury models.

*"The lack of detailed test data for model validation is another major hindrance for finite element modelling of the head. Current head models have not been fully validated and are only a qualitative simulation of a simplified surrogate." (Zhou et al., 1995)*

### **5.1 Detailed case data**

To address the limitation surrounding the existing head injury data, it is proposed that additional complete data series are sought. There are a few avenues that may be able to provide such information. These avenues are investigated throughout the remainder of this thesis.

It should be noted that this is not the only research directed towards this goal. Other research groups around the globe are attempting to derive and use head injury case data for various safety considerations (for example, American football helmet improvements and military helmet effectiveness studies). These efforts are complimentary and could be drawn on for technical (head instrumentation) solutions, where appropriate.

It should also be noted that recent evaluations have suggested that helmet-based accelerometer systems as used in American football and ice hockey, may not offer the data fidelity required for finite element model based injury threshold development (Allison et al., 2013).

## **5.2 Research questions**

On this basis it seems that some head injury research questions remain to be answered.

1. Is it possible to validate advanced head injury criterion and head models using additional (new) head injury case data so as to make their application more robust in efforts to mitigate future injuries?
2. On the basis of head injury case data, can a criterion be proposed which takes into account the dominant confounding factors which currently limit the usefulness of existing criteria?
3. Can an advanced criterion be implemented in a test tool in order to drive future vehicle safety improvements in a way that can mitigate particular types of intracranial injuries effectively?

Clearly, the latter two points identified here are a natural progression from the first, relying on the provision of new head injury case data. This seems to make the provision of detailed case data a priority in this field.

Therefore the direction I chose to pursue for my studies was to investigate the hardware, the sensor systems, that would facilitate collection of new head injury case data.

### **5.2.1      *Advanced head injury criterion***

Without detailed head impact case data, it has not been possible to validate advanced head injury criteria that have already been proposed. Equally, to develop new criteria, detailed case data would also be needed. Therefore it is suggested that the obtaining of detailed head impact data is set as a priority in the development and implementation of advanced head injury criteria.

For this reason it seems imperative that new data are sourced in order to investigate the real potential for improved prediction of particular types of head injuries. Specifically, to look at those intracranial injuries which may have had historically a low severity, based on an immediate threat to life, but which can also lead to severe impairment and important societal costs, six degree of freedom head motion data seem to be important. This is proposed as the key step in the way forward with this research.

### **5.2.2      *Potential data sources***

Two potential sources of detailed head injury case data were identified: Those arising from accident reconstructions of either motorsport accidents, or accidents involving a pedal cyclist.

In the case of the motorsport reconstructions, each would be a physical reconstruction test making use of a crash test dummy as the human surrogate. It is expected that test work would need to be performed in conjunction with research by the FIA Institute. A fundamental limitation with reconstructing motorsports accidents is that only the top formulae of motorsport require the collection of sufficient accident scene information to reconstruct the events accurately. As severe accidents are not 'commonplace' in the events at such levels, this means that the number of accidents ready to be reconstructed is small. For this reason accident reconstruction information is unlikely to yield the volume of head injury case data points required for statistical analysis investigating advanced head injury criteria. That is not to say, that information from accident reconstructions cannot be used to aid investigations into

new criteria. Any data of this quality should provide useful validation points to assess a prospective criterion.

Pedal cyclist accident reconstructions are ongoing as part of other postgraduate research (e.g. at the University College Dublin). The exact level of detail available from the accident cases, and the level of accuracy in the reconstruction technique are not known and may limit the usefulness of this data source. Such limitations and the exact specification for the collaboration, including the accidents to be reconstructed and the manner of modelling, are yet to be defined.

Whilst both of these options are being pursued, it is not expected that they will produce sufficient quantities of data to support head injury criterion development within the timescale of this study. For this reason the author has decided to investigate what can be done to generate new head injury case data in a more immediate timeframe.

### **5.2.3 Future work**

As mentioned earlier, development work has been carried out alongside American football and military studies to make use of small helmet mounted accelerometers. The progression from this technology was to shrink the accelerometers so that they can be mounted in the ear. Prototype designs of in-ear accelerometers are available in these two research domains; however, their price and some other technological features (such as, for example, measurement precision and sampling rate) limit their use to specific research studies.

*"Real-time accelerometer data collection is a novel method available to researchers who are attempting to better understand the biomechanics of mTBI, but the earlier study designs were limited and unable to provide a realistic and meaningful interpretation of the data."* (Guskiewicz and Mihalik, 2011)

Data collection from far more head injury cases would be possible if these accelerometer devices could be made small enough so that they could be worn comfortably in a wider variety of sports and if they were cheaper. At the moment a price barrier exists whereby only special studies can afford to equip sports stars. A much wider take-up and use of the technology would be expected if a sports' governing body could buy the sensors without such a financial barrier to overcome.

The next phase of this study was to develop an in-ear accelerometer design that can be used more widely. Once prospective devices were sourced, I then evaluated the designs to assess their 'fitness for purpose'.

In discussions with the FIA Institute, avenues for future research were identified. These were all subject to constraints associated with the provision of hardware and support.

1. The FIA Institute are in the process of commissioning some miniature MEMS (MicroElectroMechanical Systems) accelerometer components. The first batch of these components will have a tri-axial measurement range of  $\pm 24 g$  peak (linear) acceleration. This is about the level of acceleration measured at the head of the Hybrid III crash test dummy during Formula 1 'nose cone' approval testing. Therefore the opportunity was identified to run the new accelerometers in parallel with existing dummy sensors. The back to back data collection allowed analysis of the sensitivity of the new accelerometers and the reliability of the captured data to be compared with the established dummy sensors. The author took responsibility for the parallel data acquisition and analysis post-test.
2. The second task was to assess the coupling between the device and the head that can be achieved using the latest in-ear accelerometer mounting. For this assessment it was required that PMHS head drop tests were conducted. This work was intended to extend previous testing which used a former design of ear mounting. However, the test programme was not commissioned.

3. As mentioned earlier in this report, measurement capabilities to assess rotational velocity alongside the linear accelerations, in a head mounted sensor, are desirable. However, no robust technical solutions have been found for this purpose, yet. One sensor which has become available now may represent potential for development, although a limitation has already been identified: the measurement is not stable in large acceleration fields. The FIA Institute has an interest in quantifying the extent of this issue. Therefore, the author devised a programme to obtain some sensors and develop an evaluation programme to investigate the measurement accuracy during impact events (laboratory-based tests).

Armed with a new device, efforts were to be made to contact a particular motorsports group to arrange a trial deployment. The British SuperBikes championship was identified as a key potential data source. Subsequently, the British Touring Car Championship offered an alternative. However, the author was not involved in any efforts to roll-out data acquisition in either arena.

Given these avenues for research and limitations, it was determined that the development of a new criterion was likely to be outside the scope of this research. However, to be of use in this pursuit, the new data needed to address the limitations identified with previously reported head injury case data (for instance, offering six degree of freedom time-series data with detailed information on the exact injuries sustained) had become a priority to pursue.

Therefore the aim of the study was to:

Assess the performance of the FIA Institute miniature in-ear accelerometer system with respect to the ability to capture valuable head injury case data from future impact events.

Beyond that, an objective was to:

Develop an alternative system, based on the same technology, with the view to make it more widely accessible for collecting head injury case data.

These two aspects comprise the rest of the research documented in this thesis.

With the limited availability of solutions for collecting rotational velocity and acceleration data at this point in the research, attention was specifically directed to investigating, what could be done with linear acceleration? The rotational metrics identified in Section 3.3 are still considered by the author to be important for assessing the severity of a head loading event and should be included in future research.



## **6 Head acceleration measurement possibilities**

### **6.1 Summary of existing solutions**

#### **6.1.1 High fidelity solutions**

There has been early interest in American Football towards the need for collection of head injury loading data from matches and practice sessions. In this application, the helmets worn by all players are used as a mounting location for an accelerometer array. The acceleration data is transmitted to the sidelines in case a significant head contact is detected. The system used is known as HIT or HITS (Head Impact Telemetry System). This is a device capable of recording acceleration from six linear accelerometers at 1,000 Hz. It was developed with researchers at Virginia Polytechnic Institute and State University and has been in use since the early 2000s. The system is based around six sensors located in the padding of the helmet from which linear and angular acceleration components can be computed. If a pulse of over 10 *g* is detected, the sensors are activated and transmit a packet of data to the receiver. The HITS is used by College football teams in the U.S.

In a cohort study of 188 American football players in three national collegiate teams, Crisco et al. (2010) considered the number of head impacts greater than 10 *g* and location of those impacts sustained by the participants. All players wore Riddell (Riddell Inc.) football helmets instrumented with the HIT System. From the acceleration time-histories, the severity (linear and rotational) and duration of the head acceleration and location of the impact are computed. Crisco et al. used these data to look at the head impact exposures by impact location, player position and session type.

Whilst it is conceivable that the HITS would be adaptable for use in other sports where a helmet is worn, it is a proprietary system and hence there might not be sufficient competition to make widespread adoption financially realistic. However, trials have already been completed in a boxing setting.

The HITS was used in Instrumented Boxing Headgear (IBH) by Stojsih et al. (2010). The IBH data included 55 participants with 1930 impacts. For male participants the four 2 minute rounds produced mean values of 42 impacts (of at least 9.6 *g*), Head Injury Criterion (HIC<sub>15</sub>) of 43, Gadd Severity Index of 66, peak translational acceleration of 30 *g* and peak rotational acceleration of 2,571 rad/s<sup>2</sup>. The peak values recorded in this group were 104 impacts, HIC<sub>15</sub> of 1,652, Gadd Severity Index of 2,292, translational acceleration of 191 *g* and rotational acceleration of 17,156 rad/s<sup>2</sup>. Neurocognitive assessment data were collected after the bouts within 30 minutes and at 24 hours, as well as prior to the bouts when the baseline performance was established. Comparing pre and post bout assessments, and regardless of the acceleration measurements obtained, delayed memory was significantly different. Therefore whilst the majority of impacts were below established injury thresholds, they were of a sufficient severity so that immediately after sparring retention of new information was compromised. Studies such as the one by Stojsih et al. clearly indicate how the potential data that can be collected from helmet or head mounted systems can be used in assessing injury risk.

Higgins et al. (2007) proposed that as most field measurements assess the acceleration experienced by the player with accelerometers attached to the helmet and as helmets are designed to help mediate the amount of acceleration experienced by the head; then, accelerometers placed on the helmet may not reflect acceleration of the head. They developed a mouthpiece accelerometer and compared its performance in drop tests with a helmet-mounted accelerometer system. The *r*<sup>2</sup> value relating the peak acceleration at the headform centre of gravity to the peak acceleration from the helmet was only 0.245; whereas, with the mouthpiece accelerometer it was 0.664. Higgins et al. reported that their experimental results supported the hypothesis that an accelerometer placed intraorally would measure acceleration to the head more accurately than would an accelerometer placed on the helmet. They state that their findings suggest that placement of the accelerometer on the helmet is not a valid measurement of head acceleration.

Guskiewicz and Mihalik (2011) studied how American football players performed on concussion clinical measures after a game or practice session in which they sustained an impact exceeding 90 *g*. Athletes were tested only in the absence of a concussion diagnosis within 16-24 hours after the session. The most important finding was that non-concussed football players did not exhibit a decline in balance and cognition after an exposure in which they sustained at least one high impact greater than 90 *g*, which is a proposed theoretical injury threshold. These findings suggest that clinicians should not expect a single impact greater than 90 *g* to necessarily result in immediate symptoms of a concussion or subsequent balance or cognitive deficits that would suggest the impact affected their overall function 24 hours later. Football players are concussed by impacts to the head that occur at a wide range of magnitudes (60.51g-168.71g linear acceleration), and that clinical measures of acute symptom severity, balance, and neuropsychological function all appear to be largely independent of impact magnitude and location. There was no relationship between impact magnitude or location, and clinical outcomes of symptoms, balance, or neuropsychological performance. In short, the concussions sustained as a result of lower end magnitudes tended to present with just as many clinical deficits as those with higher end magnitudes. Thus, despite the literature suggesting that high magnitudes of head impact, particularly with high-angular acceleration, result in more serious clinical outcomes in cases of moderate or severe TBI, the magnitude and location likely do not predict clinical recovery in cases of mTBI. The findings would seem to contradict the notion that a rigid threshold for concussion can be set, given that all 22 players in the high-impact condition sustained impacts well above the proposed threshold of 70-75 *g*. This may also support the concerns over fidelity of measurements, as raised by Higgins et al. (2007).

A company called X2 Biosystems has been active with ST Microelectronics (manufacturer of MEMS accelerometers) to introduce a small accelerometer package which combines sensitive element, power and wireless communication into a small device which can be attached with adhesive to the head. The "xPatch" system is designed to be stuck to the hairless region of the head just behind and below

(posterior and inferior) to the ear. The efficacy of the coupling between the accelerometer and the skull when mounted in this way is not known. However, it is interesting to note that this is the preferred attachment over a helmet based system (as is also offered by X2). It is understood that the University of Michigan is performing comparison tests with the HITS and X2 Patch (Moore, 2014).

Initial comparative results between the xPatch and reference sensors “fixed rigidly to the foramen magnum” {presumably, fixed across the foramen magnum} have been presented recently (Siegmund et al., 2015). Tests were conducted with three post-mortem human subject heads fitted with an American football helmet and freely dropped onto the forehead, side and rear from heights of 3 to 142 cm. The Siegmund et al. results show that the xPatch overestimates peak linear acceleration by  $64 \pm 41$  %. Although, those authors included in their discussion that the linearity of the pooled data may still be of use for population estimates of acceleration exposure; given r-squared correlation coefficients between the xPatch peak acceleration and that of the reference accelerometers of 0.85. However, the angular acceleration data were not similar in this regard; varying by  $370 \pm 456$  % across all tests, and are likely to be unusable.

### **6.1.2 Lower fidelity solutions**

Other examples of sensors used to identify when a head impact event has occurred are available. The Sports Legacy Initiative has a certification programme for products offering the Hit Count<sup>®</sup> function (The Sports Legacy Institute, 2014). This is where the device counts all hits to the head over 20 g. Currently certified Hit Count<sup>®</sup> products or sensors are:

- GForce Tracker (GForceTracker.com)
- Triax (TriaxTec.com)
- Shockbox (theshockbox.com)

The GForce Tracker™ and Shockbox® products are designed to be fitted into a helmet and are therefore not readily adaptable for use without a helmet. However, in contrast, the Triax™ specifically markets two options for attachment; either in a helmet or at the back of the head in a head band.

It should be noted that the ambition for these devices is different to that anticipated for the high fidelity accelerometer systems (e.g. HITS) as described above. For these commercial products they are designed to be able to detect a potentially concussive impact, log that event and count multiple events, and commend the sports person to seek medical advice and follow concussion assessment (and subsequently, return-to-play) procedures. The fidelity of the measurement is not critical as long as the event is correctly identified as being over a specified threshold and with there being a nominal risk of concussion occurring. They are not necessarily intended to accurately assess the acceleration sustained by the head. The need for additional head impact data being obtained (as identified in Section 3) is in response to a lack of information from head impact events which could be used to identify criteria or validate detailed finite element models of the head for predicting the risk of injury (perhaps a variety of different head injury types). For this purpose, the fidelity of the measurements is vital as this will have a direct influence on the efficacy of correlation between any future criterion and the risk of an injury occurring. Furthermore, the research on head injury predictions is not limited to concussive events only. Therefore an appropriate sensor system needs to be accurate at higher loading severity levels than those associated with concussion monitoring/warning systems.

Whilst limitations have been explained regarding the lower fidelity head impact sensor solutions, it is important to note that these systems may have redundant capabilities. For instance, the Brain Sentry Impact Sensor integrates a triaxial STMicroelectronics' MEMS accelerometer into a helmet with the goal of, "identifying players who should receive further evaluation..." (Electronic design, 2013). In principle, access to the raw time-history data from this sensor could aid a more detailed review of the impact conditions leading to potentially injurious events, etc. It is just that the immediate use of the product doesn't require a detailed review of the time history and operational

requirements, such as the low power consumption (allowing use for a full year without charging) may lead to a lower than maximum data sampling and recording regime.

Impakt Protective Inc. developed a system which avoided the need for accelerometers and used force switches mounted in a helmet instead. These were implemented in an ice hockey helmet together with a Bluetooth transmitter for communication with a smartphone 'app'. When tested by Foreman and Crossman (2013) in linear impacts at 2.0, 2.5 and 3.0 m/s the sensor gave an aggregate difference of 8.9 % at the front, front boss, side, rear boss and rear impact locations, with regard to the force measured compared with the force applied.

### 6.1.3 Data fidelity assessment

To provide context for comparisons of sensor outputs, the repeatability of test measures is usually assessed according to the coefficient of variation (the standard deviation of repeated measures divided by the mean). For the approval of crash test dummies for use in regulatory applications, the U.S. National Highway Traffic Safety Administration (NHTSA) set out the coefficient of variation assessment bands shown in Table 6-1.

**Table 6-1: Assessment of coefficient of variation scores (Rhule et al., 2005)**

<b>Coefficient of variation (%)</b>	<b>Assessment</b>
0 to 5	Excellent
> 5 to 8	Good
> 8 to 10	Marginal (acceptable)
> 10	Poor (unacceptable)

These assessments relate to the output response from sensors within a dummy during laboratory and sled testing. They therefore incorporate, intrinsically, variation due to the test conditions (for instance, temperature, humidity, impact location, etc.) and to the other dummy hardware (controlling for instance, kinematics, stiffness, etc.). Whilst the scores above are applicable for both repeatability and reproducibility assessments, the equivalent ISO rating scale was applied by Bortenschlager et al. (2007) for repeatability assessments only (Table 6-2).

**Table 6-2: Rating scale to assess repeatability (Bortenschlager et al., 2007)**

Coefficient of variation (%)	Assessment
0 to 3	Good
> 3 to 7	Acceptable
> 7 to 10	Marginal
> 10	Not acceptable

Taking into account the expected sources of variation associated with these requirements, then any sensor system should really be operating in the 'good' region for repeatability (i.e. 0 to 3 %, according to ISO); otherwise, it is unlikely to be 'acceptable' when incorporated into a complete test system. For this reason subsequent assessments of sensor system performance will use the conservative repeatability framework of a lower than 3 % variation to be 'good', extending to 5 or 7 % for considerations of more complex testing scenarios and the levels where it would be difficult to recommend such a sensor under usual impact testing expectations and assessment criteria. Assessment against the NHTSA repeatability and reproducibility ratings would be more appropriate after consideration of prototype systems

With regard to required levels of accuracy for a measurement, then it is not possible to say what is appropriate without understanding the formulation and practical meaning of a future injury criterion. This means that statistical tests for equivalence between two sensor systems are difficult to apply (they usually dictate some practical understanding of the required measurement accuracy before setting acceptance limits). To facilitate some interpretations within later sections of this thesis, broad interpretive limits on accuracy (denoted with the symbol ' $\pm$ ') are suggested for peak linear acceleration, 3 ms exceedance and HIC; as below. Given the need to consider differences between measurement systems, the practical interpretation of any differences and an initial assumption that the difference does not vary systematically over the measurement range, the limits are given in terms of the unit of measurements rather than a proportion.

- Peak linear acceleration
  - $\pm < 5 g$  = useful
  - $5 \leq \pm < 12 g$  = marginal
    - A range of 70 to 75  $g$  was identified as being the threshold for concussion in padded impacts by Guskiewicz and Mihalik (2011). Therefore, uncertainty of 5  $g$  could take a prediction for a concussion outcome from one of the two possible options to the other (i.e. no concussion to concussion, or *vice versa*).
  - $\pm \geq 12 g$  = unacceptable
    - Twelve ' $g$ ' takes the mild concussion risk estimates of Somers et al. (2011) from AIS  $\geq 1$  to AIS  $\geq 2$  severity levels.

- 3 ms exceedance
  - $\pm < 5 g$  = useful
  - $\pm \geq 5 g$  = marginal
    - Limit values of 75 or 80 *g* exist in the regulations for child or adult occupants, respectively. Therefore, uncertainty of 5 *g* would be of significance assuming the estimates used to derive these limits support the two different values.
  
- HIC
  - $\pm < 125$  = useful
    - In the middle of the HIC injury risk function, where the gradient is steepest, a 5 % change in probability of injury (which seems reasonable from the point of view of having confidence in the result) coincides with a change in HIC of about 125 (Prasad and Mertz, 1985).
  - $125 \leq \pm < 400$  = marginal
    - The confidence limits around the HIC limit of 1,000 are at least 400 wide (as shown in Carroll, 2010). Therefore, this seems to indicate a point at which that level of precision has been accepted to date. It should be noted that the limits are not substantially smaller at lower HIC values.
  - $\pm \geq 400$  = unacceptable

## **6.2 Forthcoming applications**

### **6.2.1 Application**

The FIA Institute has made its plans to use earpiece accelerometers public for several years. In 2005 the miniature ear accelerometer project was considered to be a key aspect of a programme to develop an integrated ear accelerometer system for immediate use in Formula 1 and British Super Bikes, BSB (Mellor, 2005).

“The FIA Institute has been working on the development of a miniature ear accelerometer system for potential use by Formula One drivers. The accelerometer would measure the acceleration of the driver’s head during an accident and provide vital data for crash investigators.” (FIA Institute, 2010)

However, there are some stringent requirements being placed on the accelerometer technology for use in this application.

“This system is already well developed, with one using technology similar to that of the potential Formula One ear accelerometer, having previously been developed and used in the Indy Racing League. The technology used in this device was examined with a view to using it in Formula One; however it was not suitable for the specific requirements of Formula One drivers who needed a small and compact device, which above all, would sit comfortably in the driver’s ear.”

“The Formula One ear plug would be located in the driver’s earplug; initial ear accelerometer prototypes, although remarkably small proved slightly too large and uncomfortable. Developing a smaller more compact version of the ear accelerometer was a logical step.”

“In 2005 motor sport engineers and medical professionals began to examine the possibilities of making improvements to the design and fit of the accelerometer, enabling it to sit deeper in the ear canal, providing

better comfort and a closer coupling with the drivers head.” (FIA Institute, 2010)

“Following research from Wayne State University that concluded the US Racing League’s accelerometers were too bulky and could detach at increased frequencies, the FIA Institute designed an advanced sub-miniature model measuring no more than 3 mm cubed.” (FIA Institute, 2009).

- Accuracy and linearity: 5 %
- Frequency response: 20 to 1,000 Hz
- Range: 0 to 300 *g*
- Operating temperature: 10 to 50°C
- Excitation voltage: 5V (Mellor, 2005)

As noted by Mellor (2005), an open cockpit race car is a relatively harsh environment as regards noise levels and Electro-Magnetic Compatibility and electric field strength. To combat the environmental features, it was considered desirable to house the amplifier for the accelerometer, or perhaps the analogue to digital converter within the body of the ear-piece. At the very least, the pre-amplifier signal cables would have to be shielded.

Another requirement of any ear-piece accelerometer is that the data from the sensor would have to be acquired via the Event or Accident Data Recorder (EDR or ADR). This would be mounted to the car in race car motorsport applications, but could be a miniature (mobile phone sized) data logger worn in clothing or a race suit for motorcycle or karting applications. Typically the data logger would be able to record two hours of data at 1,000 Hz, though there is potential to increase the sampling rate to 10,000 Hz for one second around an impact event.

### **6.2.2 Previous performance evaluation**

The initial earplugs used in the Indy Racing League were developed especially by Endevco, model 7269 (Knox et al., 2008). While the 7269 accelerometers worked they were more expensive than the race teams or sponsors were willing to pay and the 7269s required a preamplifier to interface with the Delphi ADR2 crash recorder. A new version of the earplugs was made using less expensive sensors procured from Delphi. These were adopted by the Indy Racing League and Championship Race Teams (CART) in 2003. However, the mounting of the sensitive component in the ear-piece allowed a decoupling and phase shift of the ear-piece acceleration from that of the head itself (Begeman et al., 2006). To address too much decoupling, mini-triaxial accelerometers were evaluated, positioning the sensor deep within the ear canal portion of the earplug. The dimensions of the Endevco 7273GT are 1.5 x 1.8 x 2.0 mm. It has a specified range of  $\pm 300 g$ , with a sampling rate of 0 to 1,000 Hz. The results showed that the mini-triaxial accelerometer, which was small enough to fit in the canal portion of the earplugs, worked well in recording impact events. Knox et al. concluded that, "through careful earplug design and application of modeling it will be possible to obtain a sufficiently accurate estimate of skull (head) acceleration to be of use in designing better head protection and perhaps recommendation for further clinical evaluation for suspected head injury."

Remaining issues at this stage of development were:

- Most of the phase shift in the impact response observed in the results of Knox et al. (2008) was due to the rubber ear used in that testing. The exact performance of ear-pieces with a miniature accelerometer mounted in the ear canal of human ears is not known.
- The earplug material causes slight phase shift. Alternative materials (such as a stiffer epoxy) may help to reduce this component.

- There will always be a thin layer of compressible tissue lining the ear canal that prevents the earplug from being completely coupled to the skull. Knox et al. demonstrated that the influence of this feature of the ear on the resulting acceleration measurements could be reduced through the use of a transfer function. In their approach an average response for the ear piece was converted to the frequency domain using a Fast Fourier Transformation (FFT). The ratio of earpiece to reference acceleration value (as a frequency response) was then used to normalise all further measurements taken with the ear-piece. For this the new signal would be converted via a FFT, undergo the normalisation and be converted back to the time-domain.
- The largest issue at this point remained the cost of the sensor. In developing a bespoke prototype miniature accelerometer for this test work, Endevco had exceeded (presumably unknowingly) the price limit set for use of the sensor in Formula 1 and BSB. For this reason alternative miniature accelerometers were sought.

### **6.3 Summary regarding existing sensor systems**

It seems that there are two broad classifications of sensor systems available at the moment, either expensive high fidelity systems or cheap low fidelity systems. The issue with the first is that the price prevents widespread adoption of the technology. It may be perfect as a research tool or to demonstrate the technology, but it cannot provide sufficient data quantity to support the generation of reliable findings regarding head injury mechanisms and metrics. Alternatively, the low cost systems are becoming increasingly popular. This is particularly true given the ease of data interpretation via a smartphone and parental concerns over child and adolescent concussions. However, in this case, the access to and fidelity of the data being generated prevents it from being available to generate high quality research findings regarding head injury metrics. The HITS system used in American football goes some way to bridge the gap and provide a large quantity of data of research quality.

However, measurements obtained via the helmet may be of dubious accuracy in some instances. Also, there is the issue regarding the levels of acceleration sustained by the players routinely and the lack of observable injuries. This may reflect problems with the instrumentation, injury determination, analysis (for instance in the consideration of impact duration) or sample of the population being studied. However, until this issue is investigated none of the available technological solutions seems to address the need for high quantities of high fidelity head motion data associated with detailed injury information. Therefore, it seems that a need remains to provide a low-cost system capable of providing accurate head loading information that can be used widely to generate a sufficient quantity of data that can be used in developing new head injury criteria in a reasonable timescale.

## **7 System design specification**

### **7.1 Design specification**

To measure the acceleration of a head during an impact event it is necessary to have instrumentation attached to the head. The data from the instrumentation would have to be sampled and recorded for later analysis by a researcher. If this is to happen for a broad range of sporting events, it then becomes necessary to have the complete instrumentation system worn by the sports participant. Requirements for such a system are discussed and proposed in the following sections of this document. Each section gives regard to a particular aspect of the specification with the associated issues and approaches taken or potential approaches that could be taken.

#### **7.1.1 Installation**

At the time of specifying the mini-accelerometer, it is not known in exactly what applications such a system could be used. Primarily the intention is to make the technology available for use in sporting events. Therefore, the installation must be viable for the examples considered below. However, it seems important to try, wherever reasonable, to keep options open for use in other applications so as to increase the potential for the system to be used as widely as possible. This will generate the data needed to develop new, or validate existing, head injury metrics sooner rather than later.

There are two components of the installation which need to be considered in each instance. Firstly, the sensing component of the accelerometer, which needs to be mounted on the head of the subject and attached as rigidly as possible to the skull. The second is the supporting components of the device, expected to incorporate such things as the data logger and battery.

### *Sensor*

As mentioned, it is the intention with the sensitive element of the accelerometer to mount it to the head of the subject or sports participant. For greatest fidelity of data, the accelerometer should be mounted rigidly to the skull. However, this is unfeasible with volunteers. Instead, any mounting should emulate this goal as far as is possible without overly inconveniencing the wearer or increasing the risk of injury. At the moment, reasonable options for mounting seem to be:

#### *On the skin*

Electromyography (EMG) sensors are routinely mounted to the skin of a subject. Similarly the accelerometer could be attached directly to the skin with adhesive, either glue or sticky tape. For this approach, the sensor would have to be as low in mass as possible to minimise inertia relative to the head and the strain on the skin.

Negative aspects of direct attachment are that the skin is relatively free to slide over the surface of the skull and this might cause problems with the accurate measurement of head acceleration (particularly rotational acceleration). Also gluing or taping to the skin sounds simple in concept, but time taken to set up the sensors for each use and potential complications with removal may make this process unviable (or at least make it somewhat undesirable).

#### *In the ear*

The FIA Institute approach is to mount the accelerometer in the ear of the sports participant. This takes advantage of the device being small enough to fit into the bony part of the ear canal and that bespoke ear pieces are already worn by the top level motorsports participants. For a low-cost solution, bespoke ear pieces are unlikely to be an option. In which case, the accelerometer may have to be pushed into a simple ear plug. The requirements for this would still be to keep the size as small as possible, ideally fitting within the ear canal (staying as close as possible to the 3 x 1 mm profile of the chip).

Negative aspects of use in an ear plug are the reduction in hearing for a sports participant. While this might be considered an advantage in motorsports, it is likely to adversely affect communications in team sports, and orientation and reactions in general. Also, there is the chance that a hard component in the ear could exacerbate the risk of injury under certain circumstances, for example during a rugby tackle or boxing punch to the side of the head.

#### *In a skull cap or helmet*

As mentioned, in the American football setting it was possible to use accelerometers mounted in the helmets of the players. This means that slightly larger and more massive devices can be used but creates other requirements for the system. For instance, as the sensors might be more remote from the centre of gravity of the head, there is a greater need to be able to reconstruct rotations as well as linear acceleration measurements. This will take some development to derive the mathematical relationships depending on the exact orientation and arrangement of the sensors. It also means that at least three tri-axial sensors could be needed for each subject.

Negative aspects of using helmets are that the fidelity is dependent on how well the helmet is coupled to the head. In tightly fitting helmets the error between measured and skull acceleration should be small. However, a user may not always tightly attach the helmet (depending on convenience and comfort, etc.) and there can be significant movement in some directions even with a tightly fitting helmet. Also, not every sport requires helmets.

A positive aspect would be that the sensors are installed in the helmet in a known location and with a known orientation. This means that any processing of the raw data to generate orthogonal acceleration responses based on the centre of gravity will not vary from use to use. There should be no need to adjust the process each time the helmet is worn and there should be little or no inconvenience for the wearer.

### *In a bite-bar or gum shield*

A system which incorporated an accelerometer in a gum shield was shortlisted in 2013 for the Dyson award for innovative design (Dublin Institute of Technology, 2013). Bite bars have also been used historically when assessing accelerative loads and vibrations to volunteers. Using a gum shield requires the sensor to be extremely small, as with placing it in the ear canal. Using a bite bar would allow a larger sensor to be used, but may preclude use in aerobic sports.

A negative aspect of using a mouth mounted sensor is that of trailing cables. Whilst this might be acceptable in a laboratory, it is unlikely to be acceptable on a sports field. Instead, the sensor needs to be connected (within the mouth-mounted components) to something which will transmit the data to a receiver and data logger.

#### Summary

For each of these it seems that general concepts are consistent. The sensor needs to be:

- as low a mass as possible within practical constraints
- as close as possible to 3 x 1 mm in profile
- potentially combined in a cluster so that rotational accelerations can be recreated from the primary linear measurements

Optional extra functionality would be in the radio transmission of data to the logger, rather than having a wired connection, see the 'Other ideas and notions' section (Section 7.1.5).

#### **7.1.2 Battery/logger location**

Unlike motorsports, where the logging equipment can be mounted in the vehicle, for other sports (particularly field sports) there is a need for the logging equipment to be carried by the participant. This means that the hardware has to be small enough so that it can be carried comfortably and without presenting an increased risk of injury to the wearer.

In practice it is expected that this means if a small unit is to be strapped to the body it is in a smooth, rounded package. This should be no bigger than a mobile phone, so that it can easily fit into a pocket or pouch in the sportswear.

Ideally, it would be possible to design the battery and logging equipment so that it fits in something small enough to be worn on the wrist or around the bicep as with the current trend for wearable body monitors. Where helmets are worn, these may present a suitable volume in which the logger could be encapsulated.

### **7.1.3 Logging**

The requirements for logging data are set by the expected use of the system. Aspects that need to be defined include: the number of data channels needed, the rate at which these need to be sampled during an impact event, the duration of the logging period and the time for which the data needs to be stored before it can be accessed by a researcher and transferred to robust archive.

#### *Channels*

Whether three single axis or a tri-axial accelerometer unit is used it is important to capture the acceleration in three orthogonal directions. Therefore it should be expected that each accelerometer consists of a tri-axial accelerometer unit, meaning that there are three channels of data to be sampled.

Given the comments about resolving rotational as well as linear accelerations, one could imagine a system needing to incorporate two or three of these units. With two units mounted either side of the head, it may be possible to use differentials from the measurements to determine rotation about two axes. With three units it is, in principle, possible to resolve all three axes of rotation.

Three units with three axes each, equalling nine channels in total. Availability of angular rate sensors instead of only linear sensors may reduce the required number of channels.

### *Sample rate*

In typical short duration impact events, it is possible to see a pulse of 100's *g* occurring in fewer than 10 milliseconds. Even higher peak acceleration and shorter duration events are likely to occur and could offer useful data. To facilitate detailed measurements of such pulses, the TRL laboratory data acquisition system regularly operates at 10 or 20 kHz and can be set to sample at up to 100 kHz.

Such sampling rates would be useful for investigating head injury risk. However, it is also understood that the system used by the FIA Institute can manage only 1 kHz. Therefore, a comparable system must obtain at least 1 kHz and where possible, the faster the sampling, the better.

### *Duration (logging period)*

The required logging period for most sporting activities in which head contact can be imagined will be defined by the length of the event. For instance, a boxing match will be the length of each round multiplied by the number of rounds with some allowance for the breaks between rounds. For a rugby match it will be two halves of 40 minutes each plus the pre-match warm-up and half time break. It is generally expected that a logging period of about two hours should be sufficient to capture most sporting events.

Another proposed application for small portable accelerometers is via a project looking at placing them on child restraint systems during trials to determine the accelerations sustained in normal use. For this application, the logging period might be as long as one week. It is not clear whether this could be achieved reasonably using the same components as that used for a sports match. Alternatively, the facility should be possible to exchange batteries for a small unit which lasts about two hours, or a large unit which may last about one week.

Finally, another solution to extending the available logging periods without a large consumption of battery or data storage would be to incorporate a threshold triggering system. This is where the device only records data after a limit has been exceeded. Once that spike in acceleration has been measured, a rolling buffer of data should be

written to the data storage. In this case, provision should be given to have a buffer of perhaps 30 seconds pre and post event and to be able to record 10 or maybe 20 impact events before downloading the data. This recording of a short window of data would reduce the need for continuous access of the data storage medium, though it does not avoid the need to constantly monitor the sensors to check the last measurement value.

#### *Data storage time*

It is expected that a researcher would have access to the device within two hours of any event being concluded. Experimental design might influence this, therefore the longer available window before data are lost, then the better.

#### **7.1.4 Calibration**

As fundamental as it may seem, there is some need to offer assurance as to the accuracy of the measurements provided by the accelerometer. Calibration of the sensor is to be used in response to this need. The two features of calibration to be incorporated are to determine the direction of measurement axes and also to provide some confidence that a given value relates to that level of acceleration.

#### *Orientation*

With an ear-piece solution it may be possible to have some control over how the sensor is mounted and how the ear piece fits within the ear. However, this is not perfect and the sensitive axes may not align with anatomical axes. Some accounting for deviations can be made through software analysis of the results; however, this requires knowledge of the true orientation of the sensor.

For current ear-piece installations, it has been suggested that when the wearer is stationary with their head held broadly level, gravity can be used to judge the vertical axis position. However, in practice, the noise (random fluctuations in output, up to  $\pm 0.97 g$ ) associated with the accelerometers precludes an accurate estimation being offered without additional data or measurements being taken.

Alternatives would be to create calibration loads for the head along precise lines. This approach would require further investigation to set up a suitable fixture with a process to which subjects won't object.

Otherwise, more care would be necessary to align the sensors with the anatomical axes.

#### *Acceleration*

Usual calibration of the sensor outputs against a known reference will be expected in terms of validating the acceleration values.

### **7.1.5 Other ideas and notions**

#### *Wired logger assumed*

Based on the FIA Institute solution it was assumed that a similar approach would be followed with the sensor wired to the microprocessor and memory. However, with a wider remit, it may be advantageous to remove such a tether. Instead, a wireless link could remove the functional constraints imposed by this feature. To relay the information from the sensor to the logger, WiFi or Bluetooth could be used.

#### *Energy harvesting*

Assuming that there is no wire from the logger to the sensor, then there is the issue of how the energy consumption of the sensor can be met. One advanced option could be to incorporate some sort of energy harvesting, so that the motion of the wearer powers the device.

## **7.2 Accelerometers**

At the time of writing (in 2011) a number of tiny or 'miniature' accelerometers were available on the market. A summary of the products available then is given in Table 7-1. It should be noted that other similar products have become available since this research was undertaken. After contacting some of the companies responsible for the products shown, it became clear that not all of the sensors advertised at this time were available for sale; at least, not without a substantial order size.

Reviewing the products on offer it became clear that none of them matched the design requirements perfectly. However, the FIA Institute began negotiations with ST Electronics to see if a moderate amount of development could yield a useful accelerometer model.

This research then focussed both on validation of the system made available via the FIA Institute whilst also attempting to develop an equivalent low-cost solution which could be made available for wider use. This also helped to reduce the reliance on any one external third-party provider of technology.

**Table 7-1: Small accelerometer products advertised at time of writing**

<b>Label</b>	<b>Manufacturer</b>	<b>Description</b>	<b>Sensitive range</b>	<b>Survivable range</b>	<b>Data rate</b>	<b>Size (mm)</b>	<b>Comment</b>
<b>LIS302DL</b>	ST electronics ( <a href="http://www.st.com">www.st.com</a> )	MEMS motion sensor, 3-axis digital output accelerometer	$\pm 2 g / \pm 8 g$	10,000 <i>g</i>	100 to 400 Hz	3 x 5 x 0.9	Used in the iPhone
<b>FAR-S2AB series</b>	Fujitsu ( <a href="http://www.fme.fujitsu.com">www.fme.fujitsu.com</a> )	MEMS 3-axis small size & high sensitivity accelerometer	$\pm 5 g?$	< 5,000 <i>g</i>		5 x 5 x 2.3	Automotive applications
<b>KXPS5 series</b>	Kionix ( <a href="http://www.kionix.com">www.kionix.com</a> )	Tri-axis MEMS motion sensing accelerometers	$\pm 1.5 g$ to $\pm 6 g$			3 x 5 x 0.9	Digital or analogue output
<b>HAAM-346B</b>	Hokuriku	Piezoresistive 3-axis acceleration sensor	$\pm 2 g$	5,000 <i>g</i>	250 or 500 Hz	3 x 3 x 1	
<b>CXLTG-series</b>	MEMSIC ( <a href="http://www.memsic.com">www.memsic.com</a> )	High performance, three layer silicon MEMS, 3-axis accelerometers	$\pm 2 g / \pm 10 g$	1,000 <i>g</i>	> 200 Hz	5.7 x 3.7 x 2.8	

<b>AIS1200DS</b>	ST ( <a href="http://www.st.com">www.st.com</a> )	MEMS single-axis satellite acceleration sensor	200 g range	4,000 g	10.7 x 10.5 x 2.7	
<b>LIS331HH</b>	ST ( <a href="http://www.st.com">www.st.com</a> )	MEMS high full-scale 3-axes "nano" accelerometer	$\pm 6 g /$ $\pm 12 g /$ $\pm 24 g$	10,000 g	1,000 Hz	3 x 3 x 1
<b>Model 52</b>	Joint Sensor Instruments (Shenzhen) Ltd.	Piezo-resistive MEMS accelerometer	50, 200, 500 and 2,000 g		7 kHz	Pencil tip Looks big
<b>TRIO</b>	Tronics ( <a href="http://www.tronicsgroup.com">www.tronicsgroup.com</a> )	Miniature 3-axis Accelerometer	Low accelerati on range			1.05 x 1.65 x ?



### 7.3 Ears

In his discussion document on the use of artificial ears and challenges for headphone and telephone handset testing, Rasmussen (no date) comments that we cannot hope to find an average human who would be willing to offer their ears for test purposes. Instead there is a need for artificial test devices resembling ears.

Simulated ears are now available for this purpose in a range of sizes (small, medium, large) and hardness (35 or 55 00 Shore). Rasmussen describes the hardness of the ear in the following way,

*"If we measure the hardness of the ear lobe for a number of people it is 35 00 shore in average. If we test the pinna above the concha it is 55 to 70 00 shore. The tissue is backed by a cartilage of varying thickness and shape. If we measure the hardness above the concha a realistic average could be 55 "00 shore". (Rasmussen, no date)*

The evaluations of potential accelerometers for use in the ear, as reported in the following chapters, made use of a crash test dummy head to mount the ear-pieces. The adaptations made to the dummy in order to allow ear-pieces to be fitted made no attempt to simulate the hardness of the ear canal. In principle a soft rubber lining could have been added to the holes drilled in the skull of the dummy's head to make the interface more human-like. These weren't employed for the following reasons:

- As mentioned above, the hardness of the ear varies throughout the different regions. There does not seem to be clear guidance on the typical hardness of the human ear canal.
- Making such additional modifications to the dummy head would have substantially increased the cost and time involved in making those adaptations.
- The best solution to approximating a human ear would be to use a human and the full programme of evaluation testing for a new ear-piece was expected to include PMHS (post-mortem human subject) testing.

## **8 System performance**

This section describes:

- An initial evaluation of a low capacity prototype of a micro accelerometer. Both in vibration and drop tests;
- Follow-up testing of a similar prototype assessing the potential improvements offered with increased measurement capacity or sampling frequency.

In this testing, I defined the objectives and methods, wrote the study plan, coordinated the availability of the test objects and facility, installed and acquired data from the miniature accelerometers, analysed and interpreted the results and prepared this summary. I received assistance from staff members at TRL in safe operation of the facilities and in provision of data from the laboratory accelerometers and data acquisition system.

### **8.1 First prototype evaluation – vibration tests**

#### **8.1.1 Introduction**

A programme of testing was required to investigate the sensitivity of the new miniature accelerometers developed for the FIA Institute. These accelerometers have been commissioned with the intention of measuring head accelerations via ear pieces. To determine how suitable such accelerometers are for that purpose a programme of testing was needed.

The acceleration transducer had the characteristics shown in Table 8-1. The device was mounted on a small printed circuit board with an overall area of 3.6 x 5.1 mm.

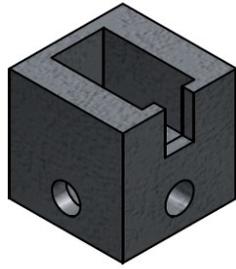
**Table 8-1: Tri-axial accelerometer device characteristics**

Parameter	Minimum	Typical	Maximum	Unit
Size		3 x 3 x 1		mm
Operating temperature range	-40		+85	°C
Product weight		20		mg
Voltage supply	2.16	2.5	3.6	V
Current consumption		300		µA
Output data rate		1,000		Hz
Low pass filter cut-off frequency		780		Hz

An initial phase of testing involved the fitting of one of the acceleration sensitive microchips into a small rigid block and shaking that block with a known acceleration magnitude and frequency of oscillation. This section of the report documents the comparisons made between the measurements coming from the miniature accelerometer and from a reference accelerometer used in the feedback loop to control the applied oscillations. These tests were carried out in the TRL accelerometer calibration facility during April and May 2012. The ultimate aim of this study was to investigate the potential for in ear accelerometers to provide valuable data from head impact events that could be used in the development of advanced head injury criteria.

#### *8.1.1.1 Mounting of the sensitive microchip component*

It was thought that to handle the miniature accelerometer in a conventional manner and to fit it to the shaker table in a robust fashion that it was necessary to place the microchip in a rigid block. The design of the block needed to accommodate the shape of the chip allowing the cable to exit without impingement. It also needed to keep the sensitive axes of the microchip aligned with the external faces of the block. To meet these requirements a small (8 mm per side) metal cube was shaped to accept the microchip (Figure 8-1). This cube was produced by rapid prototyping via metal laser sintering.



**Figure 8-1: Image of small metal cube produced to house the accelerometer chip and align the sensitive axes to the outer planes of the cube**

The microchip was held into the block with cyanoacrylate and covered with silicone sealant which was also used to provide strain relief for the cable as it exited the block.

### **8.1.2 Description of the testing**

As there was no convenient way of recording a Time-zero event for the miniature accelerometer data acquisition, data logging of the accelerometer was started before the shaking began. The oscillations were kept at stepped frequencies for at least 10 seconds to allow each frequency to be identified retrospectively in the data. As with industry standard calibrations of accelerometers used conventionally in crash testing, the oscillations applied to the miniature accelerometer were varied in the frequency range of 63 to 2,000 Hz. The highest frequency of the accelerometer sampling is known to be 1 kHz and it uses a low pass filter attenuating frequencies above 500 Hz, therefore the oscillatory frequency was made to exceed that, intentionally.

The amplitude of the oscillations was set to be around 20 *g*. This was chosen to be within the maximum sensitive range of the miniature accelerometer (24 *g*).

Complete frequency sweeps were obtained for the y and z-axis of the miniature accelerometer (excluding 125 Hz for the z-axis), whereas the x-axis result data were unfortunately truncated at 500 Hz due to an issue in the transfer process from the Event data recorder to the PC.

### 8.1.3 Results

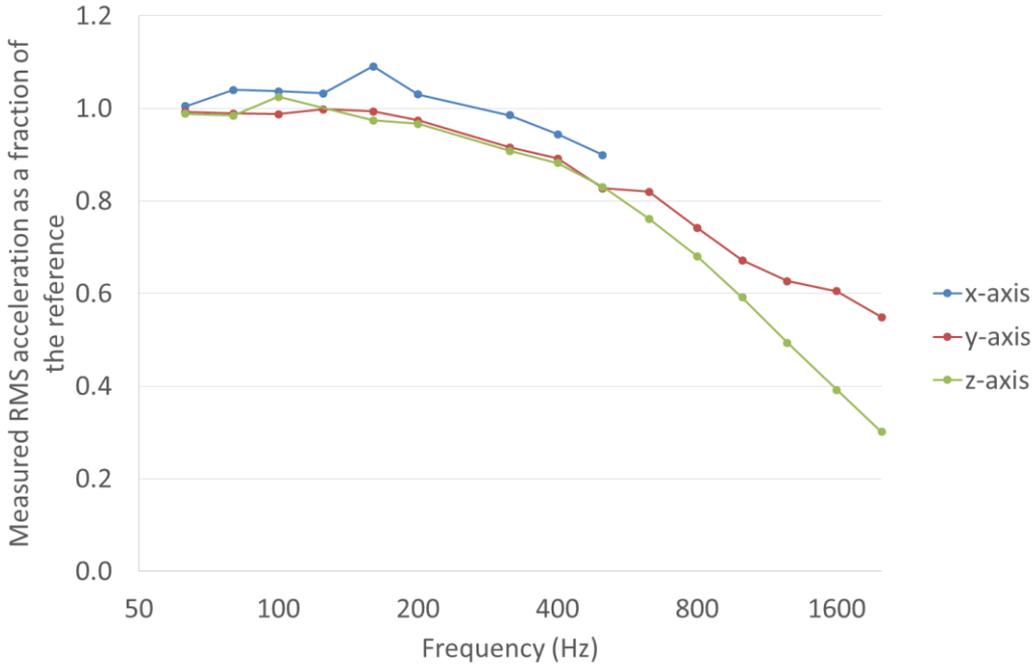
The measured accelerations obtained from the reference and miniature accelerometers and reported as RMS (root mean square) values are shown in Table 8-2. These data are plotted in Figure 8-2 with the miniature accelerometer RMS values shown as a fraction of the reference accelerometer values.

**Table 8-2: Reference and miniature accelerometer RMS values for different input frequencies of sine wave oscillations**

Frequency of oscillation (Hz)	x-axis		y-axis		z-axis	
	Reference (g)	Miniature accelerometer (g)	Reference (g)	Miniature accelerometer (g)	Reference (g)	Miniature accelerometer (g)
	<b>63</b>	14.45	14.52	14.59	14.49	14.36
<b>80</b>	14.38	14.96	14.71	14.55	14.64	14.42
<b>100</b>	14.28	14.81	14.57	14.39	14.50	14.86
<b>125</b>	14.24	14.70	14.89	14.86	†	†
<b>160</b>	14.26	15.56	16.02	15.91	14.47	14.10
<b>200</b>	14.24	14.68	14.55	14.17	14.41	13.92
<b>315</b>	14.26	14.06	14.32	13.12	14.42	13.10
<b>400</b>	14.28	13.48	14.48	12.92	14.42	12.72
<b>500</b>	14.22	12.79	14.44	11.95	14.24	11.83
<b>630</b>	†	†	14.90	12.21	14.67	11.17
<b>800</b>	†	†	14.94	11.08	14.39	9.80
<b>1000</b>	†	†	14.48	9.72	14.46	8.56
<b>1250</b>	†	†	14.64	9.18	14.69	7.25
<b>1600</b>	†	†	14.99	9.08	14.29	5.60
<b>2000</b>	†	†	14.79	8.12	14.56	4.39

† Data for this condition were not acquired.

From Figure 8-2 certain key features of the miniature accelerometer’s behaviour can be seen. For instance, the measured output decreases as the frequency increases. At 2 kHz, the miniature accelerometer was recording just 30 or 55 percent of the applied acceleration magnitude.



**Figure 8-2: Accelerometer RMS values as a fraction of the reference for different input frequencies of sine wave oscillations**

It should be noted that the sampling frequency of the data acquisition from the miniature accelerometer is limited at 1 kHz, with the 500 Hz low pass filter, therefore one may expect an underestimate of the measurements beyond this. At the applied oscillation frequency of 500 Hz the miniature accelerometer was recording about 80 percent of the applied magnitude; and at 1 kHz, the miniature accelerometer was down to recording 59 or 67 percent of the applied magnitude. With an applied frequency equal to that of the sampling rate it is expected that only one sample per oscillation will be measured. Therefore, this performance should be anywhere between 0 and 100 percent of the applied magnitude. In this instance, a Fast Fourier Transformation of the miniature accelerometer data showed that it was measuring a signal with a major frequency component of between 6 and 10 Hz. This indicates that

the nominal 1 kHz applied oscillation and sampling frequency were actually separated by just less than 10 Hz. At lower oscillation frequencies the miniature accelerometer was much better able to measure acceleration magnitudes closer to the applied magnitude. The decrease in response was progressive over the frequency range tested starting from about 160 to 200 Hz and reaching the levels at 1 kHz and 2 kHz as were described above. The first natural frequency of the human head is in the region of 100 to 150 Hz, with relative brain-skull movement occurring beyond those frequencies (Willinger et al., 2001). It is important that accurate measurements can be obtained up to and beyond this frequency range, where injuries are dependent upon relative movement of the brain within the skull.

At low frequencies, in a few cases the miniature accelerometer gives measurements in excess of the reference accelerometer, particularly with the x-axis data. This suggests that the gain/sensitivity may be set incorrectly for low frequency accelerations.

Up to 500 Hz oscillations, the transverse sensitivity (crosstalk – values measured in one axis when the oscillations are in a perpendicular direction) between axes was in the range of 1 to 11 percent. The largest values occurred in the z-axis measurements during y-axis excitation. The ISO (ISO 6487) performance requirements for a transducer state that the transverse sensitivity ratio shall be less than 5 percent in any direction. Therefore, the miniature accelerometer would not meet these requirements. It should be noted that small errors in the alignment of the accelerometer with the direction of oscillation would artificially produce an increase in the transverse sensitivity ratio. This would result in an increase in the acceleration measured in what should be a transverse direction and a decrease in what should be the aligned axis. Whilst there may have been slight inaccuracies with the way the accelerometer was mounted on the shaker table and in the rigid block, large deviations between the excited axis measurements and the reference magnitude were not seen at low oscillation frequencies. Based on these measurements it is considered that set-up inaccuracies may have accounted for one or two percent of the crosstalk rather than the five or six percent which would be required if the accelerometer was to meet the ISO requirement.

#### **8.1.4 Summary and conclusions**

Accelerometer calibration-type shaker table tests were carried out with the miniature accelerometer purchased by the FIA Institute for consideration as in-ear instrumentation. This required mounting the accelerometer in a bespoke rapid prototyped rigid block.

Oscillation frequencies were swept through a range from 63 Hz to above the sampling rate of the accelerometer data acquisition system (up to 2 kHz).

A substantial decrease in the measured accelerations from the miniature accelerometer was observed when compared with the reference accelerations. This decrease became larger progressively as the frequency of oscillations was increased from about 200 Hz to 2 kHz. This frequency sensitivity is likely to affect the acceleration measurements of short duration impact events.

Transverse sensitivity was also noted with the miniature accelerometer, which exceeded the ISO 6487 transducer requirement.

## **8.2 First prototype evaluation – drop tests**

### **8.2.1 Introduction**

As mentioned in the previous section, a programme of testing was required to investigate the sensitivity of new miniature accelerometers developed for the FIA Institute. These accelerometers were commissioned with the intention of measuring head accelerations via ear pieces. To determine how suitable such accelerometers are for that purpose a programme of testing was devised.

This phase of the testing involved fitting the ear-pieces, instrumented with FIA Institute demonstration accelerometers, into special holes I made in the side of a Hybrid III dummy head and dropping that head. This section of the report documents comparisons made between the measurements coming from the ear-piece accelerometers in simulated dummy-ears and the measured accelerations from the

conventional accelerometers mounted at the centre of gravity of the dummy's head. These tests were carried out in the TRL helmet drop test facility during March 2012.

As mentioned in the previous section, it should be remembered that the ultimate aim of this study was to investigate the potential for in-ear accelerometers to provide valuable data from head impact events that could be used in the development of advanced head injury criteria.

#### *8.2.1.1 Description of the testing*

The severity of the testing was limited by the measurement range of the in-ear accelerometers. They are only capable of measuring up to  $\pm 24 g$  in each axis. For this reason the drop height and impact surface were carefully matched so as to test close to this acceleration level without exceeding it.

In each test the dummy's head was dropped from a specific height to impact a flat surface, either padded or in initial tests, a rigid surface. The head was guided during the free-fall phase and released at impact.

In the tests with a padded impact, the impact surface was covered with a single (25 mm) thickness of CONFOR™ foam (CF-45, blue). This padding provided an impact with a duration of around 25 ms. These tests complimented initial drops onto the rigid surface where the impact duration was around only 10 ms.

#### *8.2.1.2 Modification to the Hybrid III head*

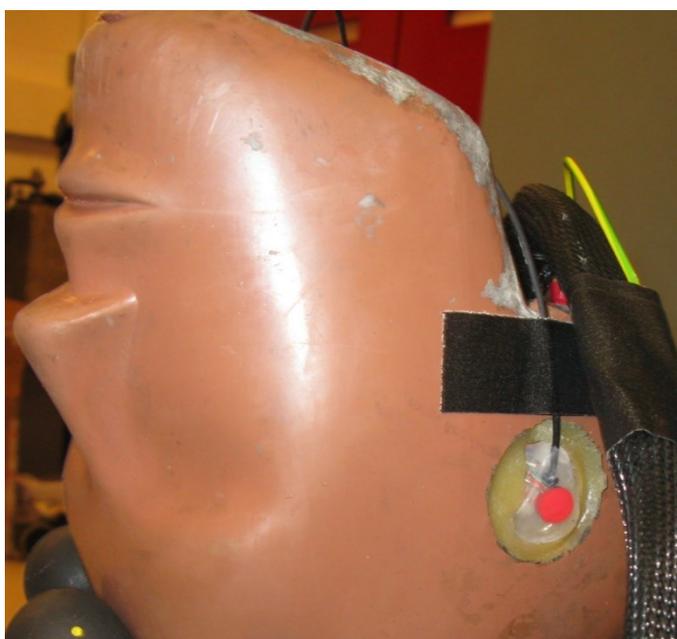
To allow the ear-piece accelerometers to be fitted to the Hybrid III dummy head it was necessary to fashion simulated ears in the dummy. Two features comprised these ears; firstly a hole was drilled into the side of the metal dummy skull, and secondly epoxy ears were moulded around this hole.

In humans, the ears (the auditory meatus) tend to be slightly posterior and inferior to the centre of gravity of the head (though this measurement depends on the particular details of the study [Yoganandan et al., 2009]). If the results from dummy testing need to be compared with those expected from a human subject it would make sense

to try and match these positions in the dummy. However, in this testing, it was deemed important to provide an ear measurement position as close as possible to the reference measurement. For this reason the holes drilled into the dummy skull were made directly over the moulded indentations in the dummy skull that indicate the reference centre of gravity prior to addition of the extra head ballast.

Once the holes were drilled, the ear-pieces were placed into them and then an epoxy outer ear was cast. The ear was moulded around the ear-piece to give a very good fit.

The accelerometers in the Hybrid III head were fitted in the standard way so as to be aligned with the head 'anatomical' axes. The placing of the miniature acceleration sensing component within the ear-piece cannot ensure any specific orientation. Therefore, before testing, it was not known how the sensitive axes of the ear-piece accelerometers were aligned with either the anatomical or laboratory coordinate systems.



**Figure 8-3: Moulded epoxy ear to surround ear-piece**

### **8.2.2 Method**

The climatic conditions required when using a Hybrid III headform in certification-type testing are that the temperature should be controlled to be from 20.6 to 22.2 °C with a relative humidity of 10 to 70 %. These requirements were not observed as a requirement for this test work as the objective of the testing was to compare measurements from the new accelerometers directly with reference accelerometers. However, the temperature was maintained within the range of 20.4 to 24.4 °C and humidity from 31 to 37 %.

The Hybrid III User's manual also stipulates leaving at least 2 hours between head certification tests. Due to the low severity nature of this testing, such a requirement was not considered as being necessary.

To investigate the directional sensitivity of the acceleration measurements four impact configurations were tested. Each of these was obtained by carefully adjusting the orientation of the dummy head prior to release. The four orientations resulted in the initial contact being between the surface and the following points in the dummy's head:

1. Forehead
2. Top of head
3. Between forehead and top of head aligned to act through head centre of gravity
4. Side of the forehead (oblique to mid-sagittal plane)

The matrix of tests including repeats is shown in Table 8-3. The column labelled 'Angle of head' shows the angle between the vertical axis of the dummy head, rotated about the y-axis, and the vertical laboratory axis.

For the oblique tests, the head was positioned as for the forehead tests but then rotated so the initial impact point was on a plane about 45 degrees between the mid-sagittal and mid-coronal planes.

In the last test, an additional layer of padding was added, this was provided by some soft packaging foam, the exact properties of which are not known.

In the ear-piece accelerometers, the maximum output data rate is 1 kHz. This was 100 times less than the sampling frequency used with the reference accelerometers. The data rate from the ear-pieces is matched with a CAN connection interface which reads the analogue to digital converter from the accelerometer at 1 kHz with 12 bits data precision. This output gave each channel a measurement precision of about 0.01 *g* ( $\pm 0.006$  *g*). The reference accelerations were recorded with a 16 bit data acquisition system and therefore had a similar level of precision despite being set to record  $\pm 400$  *g*.

The internal dummy accelerations were filtered using Channel Class 1000 phaseless filters. No post-processing was applied to the data from the ear-piece accelerometers. This is because the data sampling rate would not support the same filtering as the reference accelerometers, which is specified for head accelerations within SAE J211. Also the accelerometer output is reported to already have included a 500 Hz low pass single pole filter.

**Table 8-3: Test matrix**

<b>Test number</b>	<b>Orientation</b>	<b>Surface</b>	<b>Angle of head</b>	<b>Drop height (mm)</b>
<b>B262D05</b>	Top of head	Rigid	0°	25
<b>B262D06</b>	Top of head	Rigid	0°	25
<b>B262D07</b>	Top of head	25mm Confor	0°	97
<b>B262D08</b>	Top of head	25mm Confor	0°	206
<b>B262D09</b>	Top of head	25mm Confor	0°	206
<b>B262D10</b>	Top of head	25mm Confor	0°	206
<b>B262D11</b>	C-of-G	25mm Confor	50°	206
<b>B262D12</b>	C-of-G	25mm Confor	50°	206
<b>B262D13</b>	C-of-G	25mm Confor	50°	206
<b>B262D14</b>	Forehead	25mm Confor	57.5°	216
<b>B262D15</b>	Forehead	25mm Confor	57.5°	216
<b>B262D16</b>	Forehead	25mm Confor	57.5°	211
<b>B262D17</b>	Forehead	25mm Confor	57.5°	206
<b>B262D18</b>	Oblique (neck Y axis at ~50°)	25mm Confor	Angled in two planes	206
<b>B262D19</b>	Oblique (neck Y axis at ~50°)	25mm Confor	Angled in two planes	206
<b>B262D20</b>	Oblique (neck Y axis at ~50°)	25mm Confor	Angled in two planes	206
<b>B262D21</b>	Forehead	25mm Confor + 49mm other packaging foam	57.5°	206

### **8.2.3 Results**

The peak resultant head acceleration from each test as measured by the left or right ear-piece or at the dummy head centre of gravity is shown in Table 8-4.

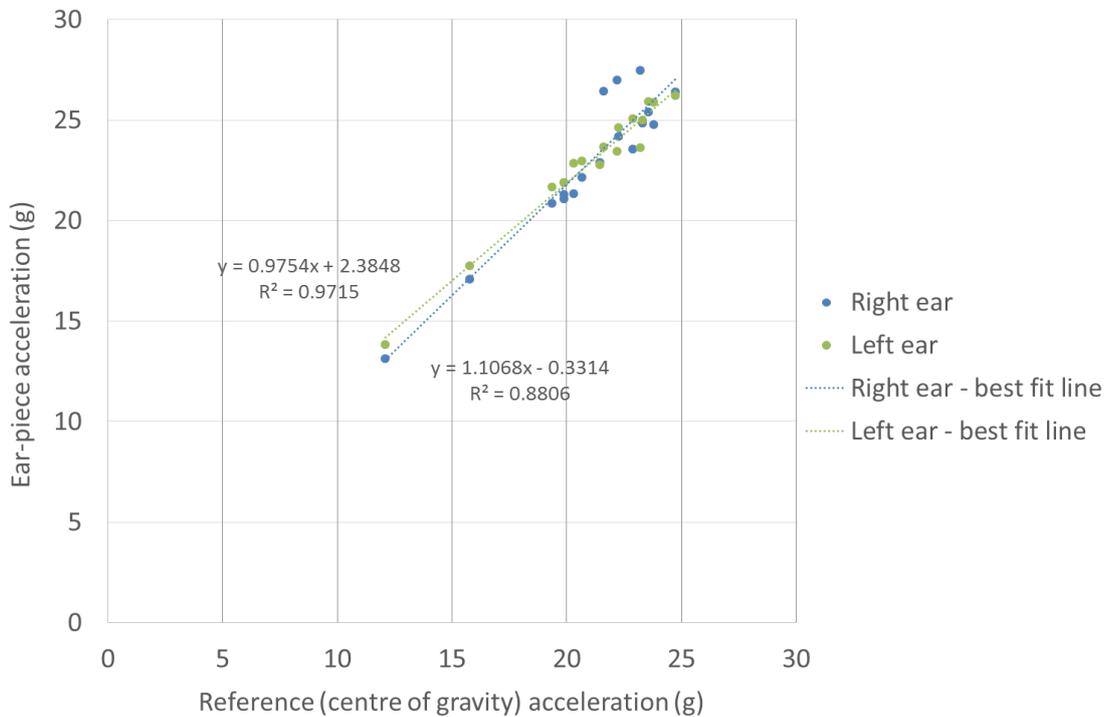
In every test the ear-piece accelerometers gave a higher peak resultant acceleration value than that from the reference accelerometers at the centre of gravity of the dummy's head. In some tests the left ear-piece yielded a higher value than the right and in other tests the right gave a higher value than the left. However, these differences were not statistically significant and may just be an indication of slight variation in test set-up.

The magnitude of the over-estimation in peak acceleration coming from the ear-piece accelerometers varied from 103 to 122 % of the reference acceleration value. There appears to be no clear pattern as to a particular configuration that produces substantially higher or lower over-estimates. The exception to this is the oblique tests where the right ear-piece gave a peak resultant acceleration of 26.9 *g* (as the mean of three tests), 121 % of the reference; compared with the left ear-piece which gave a lower acceleration of 23.6 *g*, 106 % of the reference. Excluding these tests the ear-piece accelerometers always gave a peak acceleration value within 114 % of the reference accelerometers at the head centre of gravity.

**Table 8-4: Peak resultant acceleration values**

<b>Test number</b>	<b>Left ear</b>	<b>Right ear</b>	<b>Centre of gravity (laboratory instrumentation)</b>
<b>B262D05</b>	25.1	23.5	22.9
<b>B262D06</b>	22.8	22.9	21.5
<b>B262D07</b>	13.8	13.1	12.1
<b>B262D08</b>	22.9	22.1	20.7
<b>B262D09</b>	21.9	21.3	19.9
<b>B262D10</b>	21.9	21.1	19.9
<b>B262D11</b>	21.7	20.8	19.4
<b>B262D12</b>	22.8	21.3	20.3
<b>B262D13</b>	25.0	24.8	23.3
<b>B262D14</b>	26.2	26.4	24.7
<b>B262D15</b>	25.9	25.4	23.6
<b>B262D16</b>	25.9	24.8	23.8
<b>B262D17</b>	24.6	24.2	22.3
<b>B262D18</b>	23.6	27.4	23.2
<b>B262D19</b>	23.7	26.4	21.6
<b>B262D20</b>	23.4	27.0	22.2
<b>B262D21</b>	17.7	17.1	15.8

The data from Table 8-4 are shown graphically in Figure 8-4. Best fit lines were used to derive an 'R<sup>2</sup>' coefficient, which was between 0.88 for the right ear and 0.97 for the left.

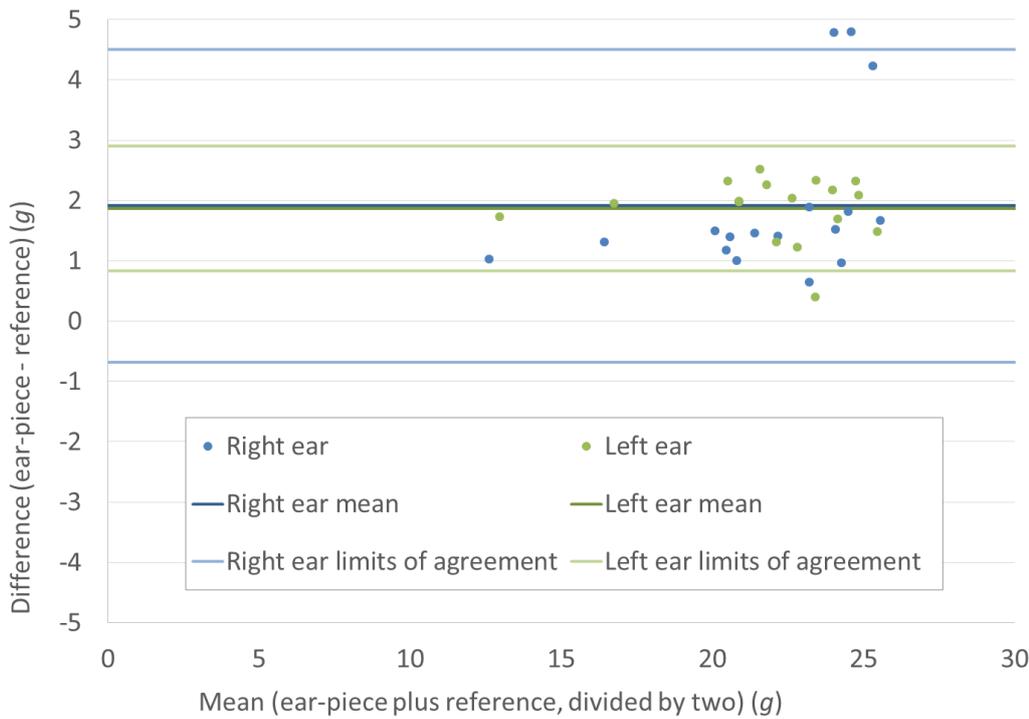


**Figure 8-4: Peak resultant acceleration values**

The same data from Table 8-4 and Figure 8-4 are plotted again in the Bland-Altman plot shown in Figure 8-5. This type of figure shows the difference between the new measurement system (the ear-pieces) and the existing system (laboratory accelerometers at the head centre-of-gravity) as the y-axis and the mean value of the two systems as the x-axis. A perfectly equivalent system will have data points dotted along the  $y = 0$  line; indicating that there is no difference between the two measurements. If the mean difference is different to 0, then there will be an offset with respect to  $y = 0$ . Limits of agreement are added to the figure about the mean difference to show confidence limits based on the mean plus or minus the standard deviation (precisely 1.96 times the standard deviation to generate confidence at the 95<sup>th</sup> percentile level, two-tailed). These limits of agreement show what can be expected, statistically, from the ear-pieces based on these tests. That is to say, that based on this sample we can expect 95 % of the measurements to fall within these limits. The new measurement system is considered to agree with the existing system if those limits are smaller than the practical constraints on accuracy set for an application (for instance, compared with the levels introduced in Section 6.1.3).

Both Figure 8-4 and Figure 8-5 show that the ear-pieces consistently provided peak resultant acceleration values greater than those from the centre-of-gravity sensors. This is not an issue in general as the ear-pieces could be used to generate sensor-specific injury risk functions and criteria. Alternatively, if a constant offset is determined, then a simple sum can be used to relate the ear-piece measurements to another system. For instance, with these data this could be accomplished by subtracting 2 *g*. However, it is illustrative of a systematic inaccuracy with the system and, whilst not large with these data, care needs to be taken to ensure that this offset is not omitted when drawing conclusions from the ear-piece measurements and, particularly, if used in deriving an injury risk function.

The limits of agreement are widest with the data from the right ear piece. From these tests it can be observed that, at the 95<sup>th</sup> percentile confidence level, the ear-pieces can over-estimate the peak resultant acceleration by almost 5 *g*. Referring back to the comments made in Section 6.1.2, then a  $\pm 5$  *g* precision was identified as being 'useful' in respect to measurements of peak resultant acceleration. These measurements from the first evaluation of the first prototype ear-piece accelerometers have 95<sup>th</sup> percentile limits of agreement which are just smaller than  $\pm 5$  *g*. As such they can be considered as being useful; but they are close to the limit, particularly with the 2 *g* offset.



**Figure 8-5: Bland-Altman plot of peak resultant acceleration values from first prototype accelerometer ear-pieces**

**(please note that right and left ear means lie close to one another)**

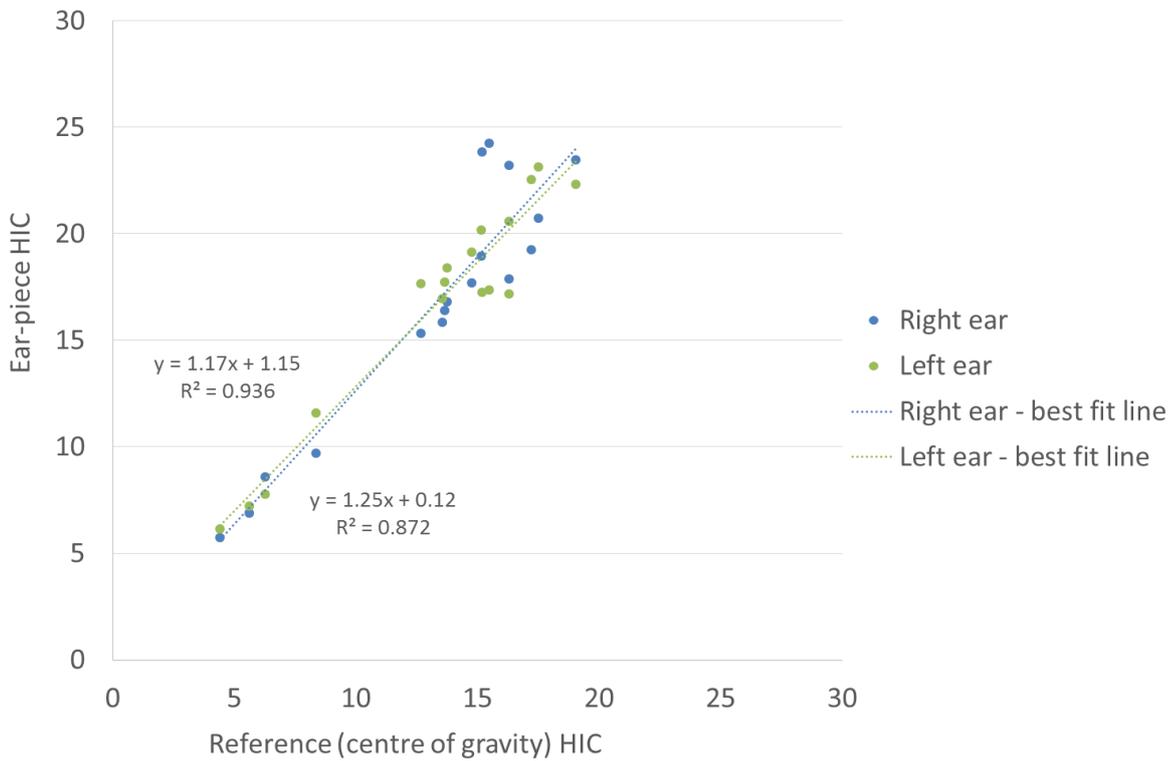
Tests B262D08, D09 and D10 were all top of the head impacts with the head dropped 206 mm onto a 25 mm thick, flat sheet of Confor foam. Tests B262D11, D12 and D13 were also conducted with the same impact conditions but with the head aligned so that the impact acted through the centre of gravity. Also, B262D18, D19 and D20 had the same oblique impact conditions as one another. Based on the peak acceleration values, the coefficients of variation for these three sets of three tests are shown in Table 8-5. With coefficients from 0.5 to almost 10 %, then the repeatability can be assessed as being good to marginal, according to the coefficient of variation assessment levels set in Section 6.1.2. The marginal level is beyond that identified earlier as allowing recommendation of a sensor system. However, the variation from the ear-piece accelerometers was not substantially greater than from the centre of gravity reference measurements. The ear-pieces were not systematically more variable with regard to these peak acceleration values. Instead the variation seems similar between all three of the accelerometer groups and is likely to reflect the test-

to-test variability in drop height precision and hence impact speed, behaviour of the Confor foam and changes due to temperature and humidity, etc. Given this understanding, further comparisons were not made on the basis of repeated testing at the intended same impact conditions. Instead, for every test, the centre-of-gravity measurements were taken to represent the 'gold standard' or 'industry norm' for assessing head impact severity. These define the input for the ear-piece accelerometers. The research question relates to the accuracy of the miniature accelerometer measurements for each and every impact compared with the conventional hardware and data acquisition system and not to aggregated mean measurements.

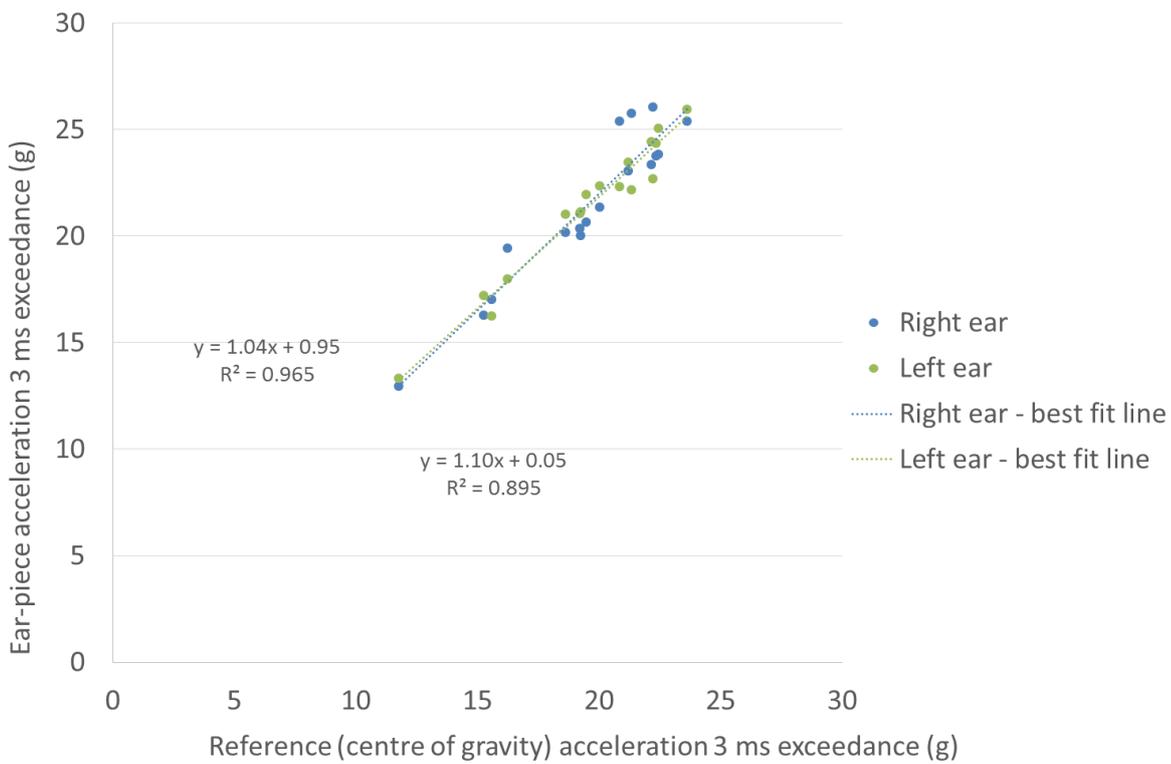
**Table 8-5: Coefficients of variation from peak resultant acceleration values in repeated tests**

Test group	Coefficient of variation		
	Centre of gravity	Left ear	Right ear
<b>B262D08 to B262D10</b>	0.023	0.028	0.026
<b>B262D11 to B262D13</b>	0.098	0.097	0.073
<b>B262D18 to B262D20</b>	0.036	0.005	0.019

As for Figure 8-4, equivalent results for HIC<sub>15</sub> and for the maximum acceleration exceeded for 3 ms are shown in figures 8-6 and 8-7, respectively. Similar trends are evident with similar correlation for these measures, as for peak acceleration. One feature that is evident with the HIC<sub>15</sub> values is that the ear-piece accelerometers overestimate HIC by more than peak acceleration or the 3ms exceedance value. The ear-piece HIC values are still always in excess of those from the centre of gravity reference accelerometers and are between 105 and 157 percent of the centre of gravity values.

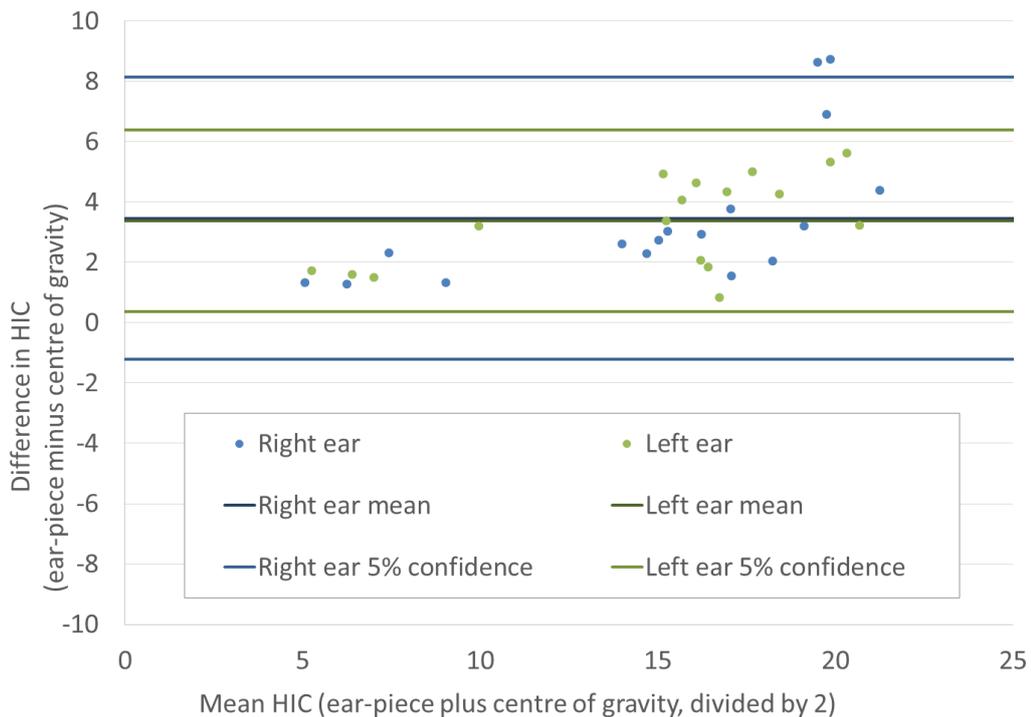


**Figure 8-6: HIC<sub>15</sub> values**

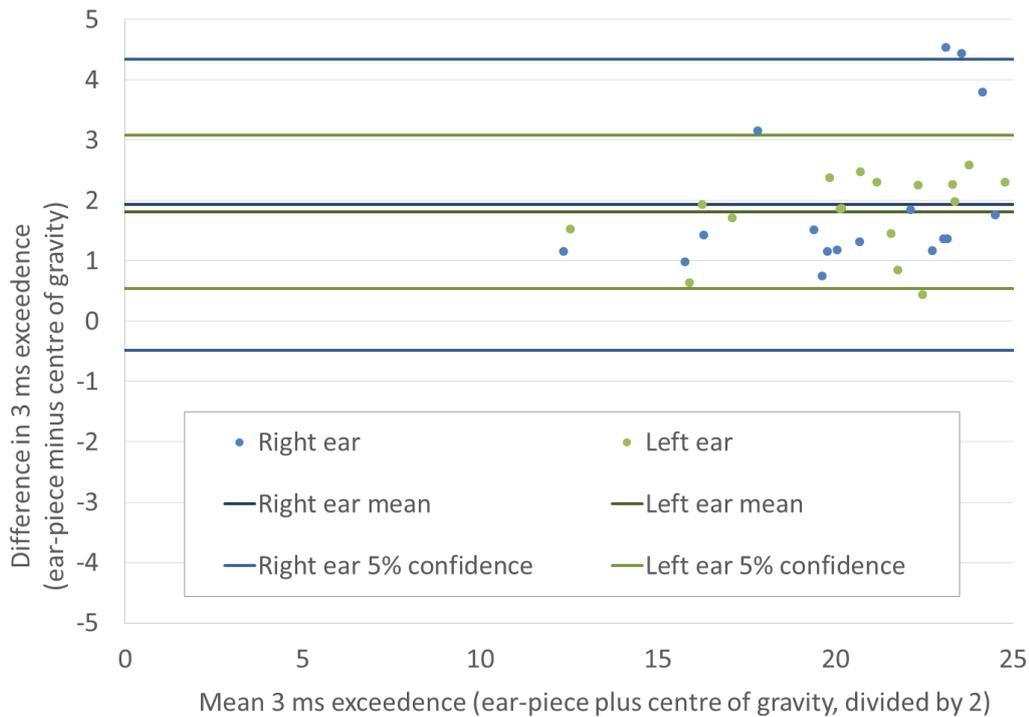


**Figure 8-7: Acceleration 3 ms exceedance values**

The overestimation of the HIC value from the ear-pieces compared with the centre-of-gravity reference accelerometers can be seen again in the Bland-Altman plots of these data. This plot together with the equivalent figure for 3 ms exceedance are shown in Figures 8-8 and 8-9, respectively. In both cases, as with the peak acceleration, the mean difference is above 0 confirming the overestimation. The 95<sup>th</sup> percentile limits of agreement with the HIC data are widest with the right ear and extend from -1.2 to 8.1. These limits of agreement are well within the confidence limits generated with existing HIC injury risk functions or the difference in HIC required to bring about a substantially different prediction of injury (skull fracture or severe brain injury) risk. The 95<sup>th</sup> percentile limits of agreement with the 3 millisecond exceedance values extend from -0.5 to 4.3 g. These are also within the useful range of accuracy set out in Section 6.1.2 ( $\pm 5$  g).



**Figure 8-8: Bland-Altman plot of HIC<sub>15</sub> values from first prototype accelerometer ear-pieces**



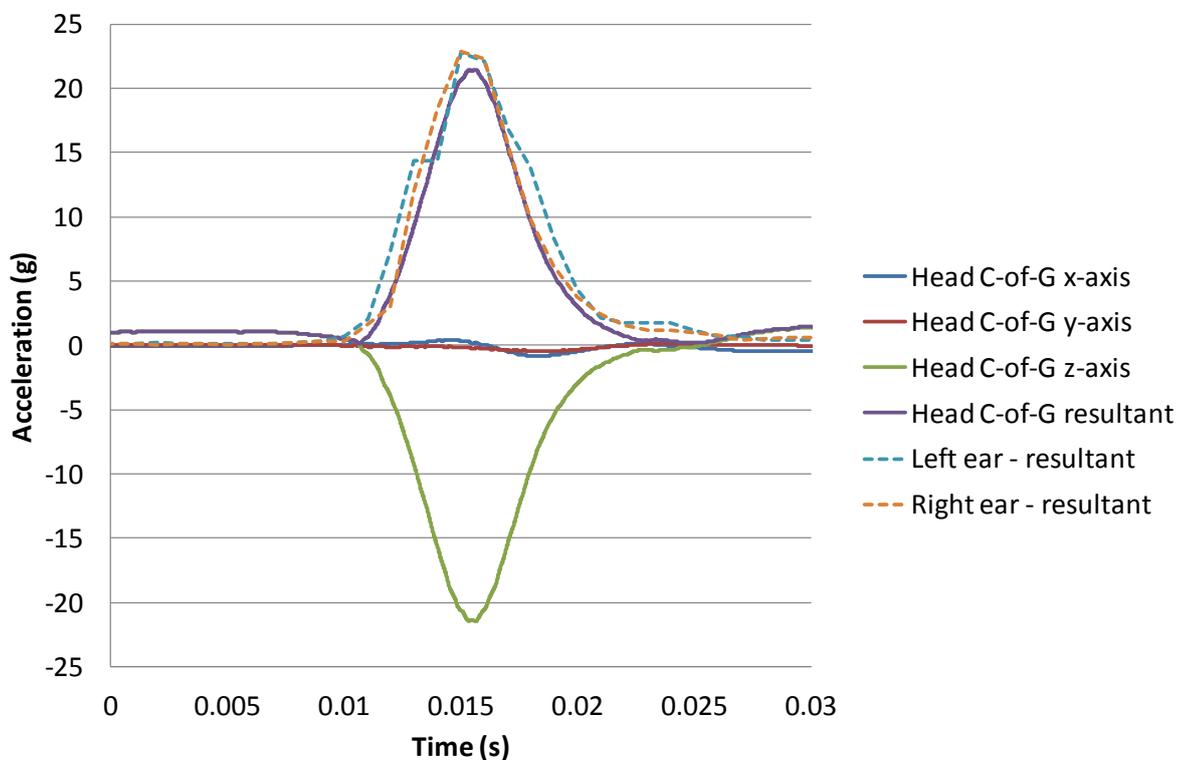
**Figure 8-9: Bland-Altman plot of 3 ms exceedance values from first prototype accelerometer ear-pieces**

Acceleration results from the second rigid impact surface, top of the head test (B262D06) are shown in Figure 8-10. The large negative peak is the z-axis acceleration from the centre of gravity reference accelerometer. It has a peak value of  $-21.5\text{ g}$  and duration of about 12 ms.

In this impact configuration the x and y-axis measurements are close to zero at the moment of peak loading. Therefore, the largest component of the resultant acceleration comes from the z-axis.

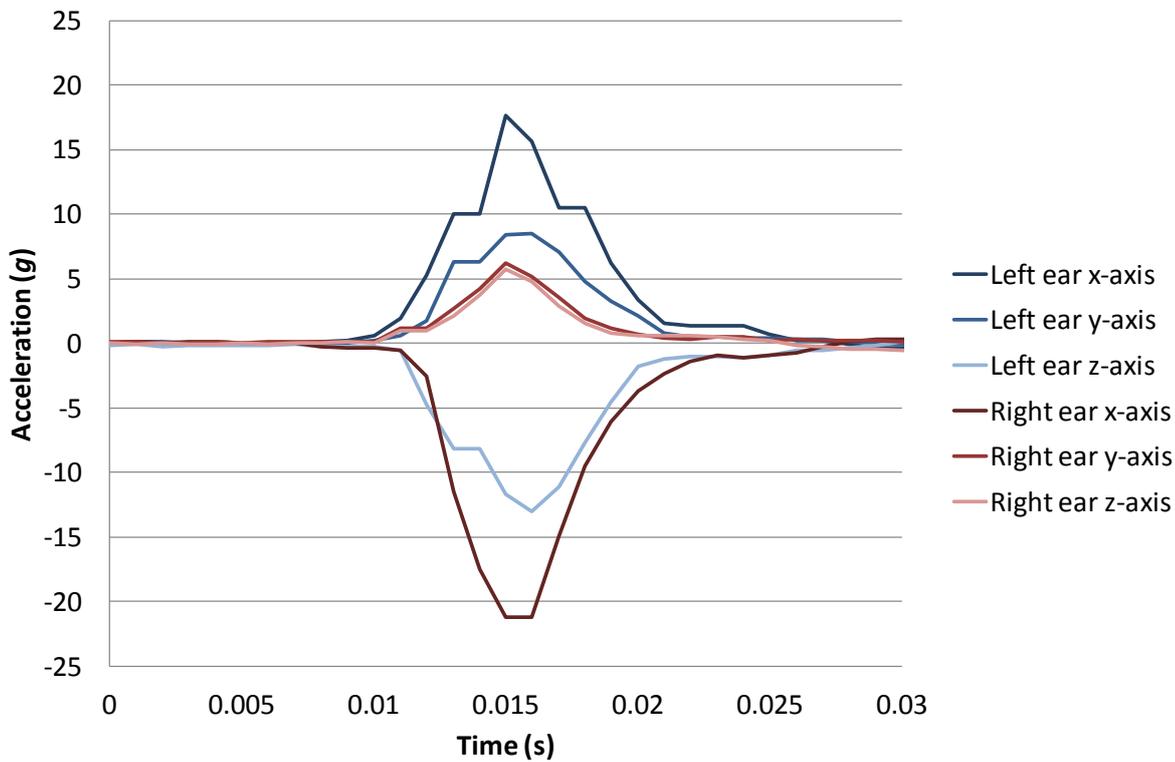
The resultant accelerations from the two ear-pieces are also shown in Figure 8-10. It is clear from this figure that the ear-piece accelerations seem to match quite closely that from the reference accelerometers. The duration of loading is similar to that shown by the reference accelerometers, perhaps being slightly longer for the left-ear. In this test, both the right and left ear show a very similar peak resultant value, of just less than  $23\text{ g}$ . This is 106 % of the reference peak value.

The more stepped response from the ear-piece accelerometers reflects the lower sampling frequency available with those sensors compared with the reference accelerometers.



**Figure 8-10: Head acceleration results from top of head rigid impact surface test**

The tri-axial components of the ear-piece resultant accelerations are shown in Figure 8-11. The resultant 23 g for the left ear comes from peak x, y and z accelerations of about 18, 9 and -13 g, respectively. Whereas the 23 g right ear peak resultant acceleration comes from peak x, y and z axis accelerations of -21, 6 and 6 g, respectively. This result shows how the relative orientation of the tri-axial accelerometers in the ear-pieces is different with respect to the head or laboratory frame of reference. None of the ear-piece accelerometer axes are aligned with an axis of the head.

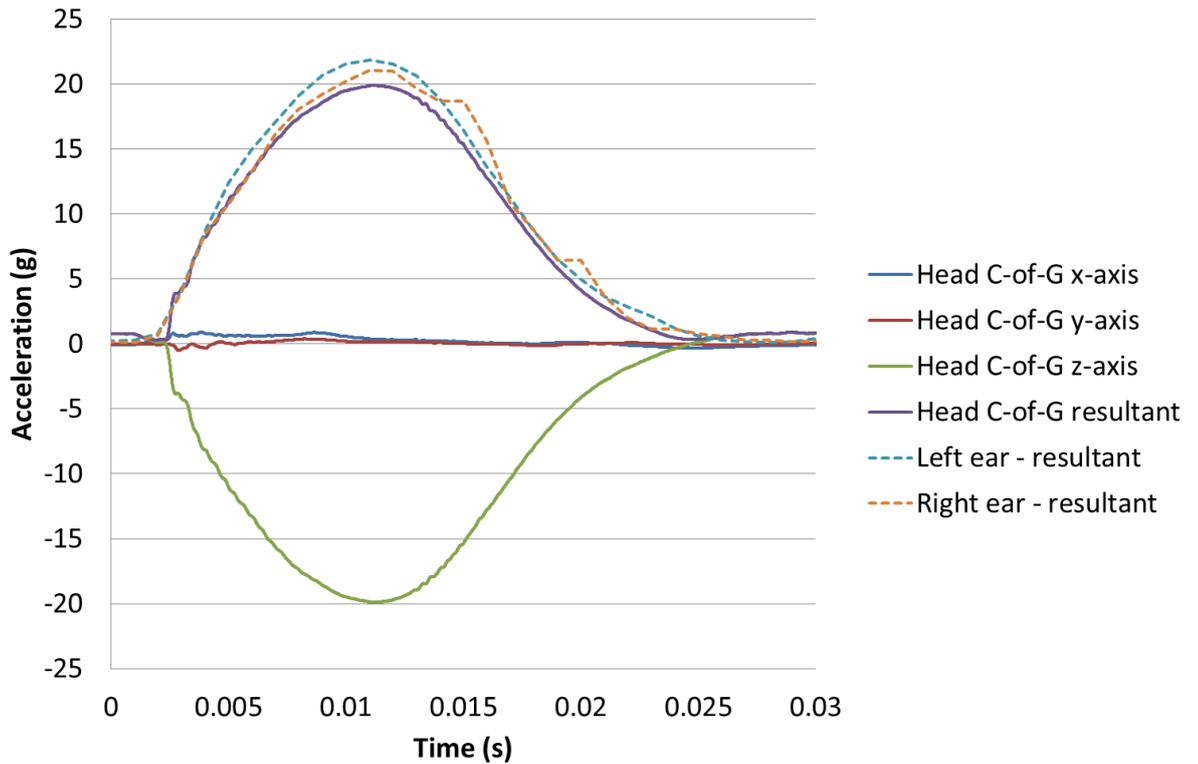


**Figure 8-11: Component head accelerations from top of head impact surface test**

The acceleration results from the fourth padded impact surface, top of the head test (B262D10) are shown in Figure 8-12. Comparing this padded impact to the rigid impacts as shown in Figure 8-10 then it can be seen how the impact duration has been extended through the addition of the Confor foam. The peak acceleration has not been reduced as the drop height was increased with the addition of the foam.

Subjectively it can be said that the ear-piece acceleration time-histories match quite closely those from the centre of gravity reference accelerometers. There is a difference in the magnitude of the peak resultant acceleration, with the ear-piece accelerometers producing higher peak values, HIC and 3 ms exceedance values than the reference accelerometers.

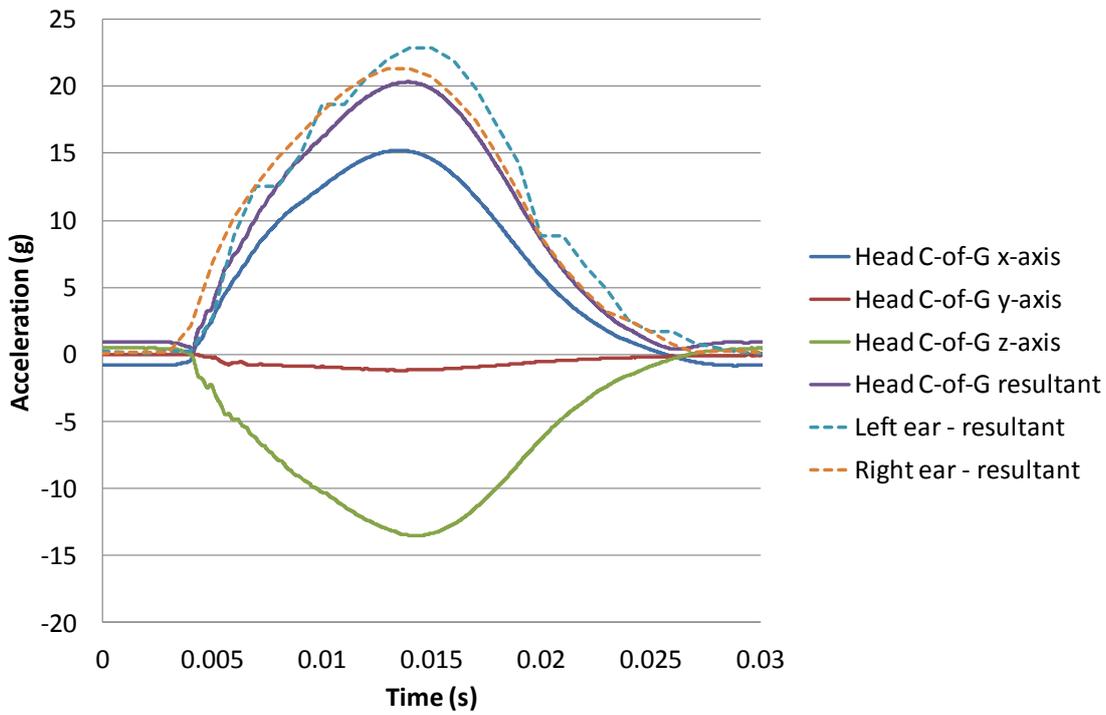
Also the top of head impact orientation can again be seen to generate almost zero acceleration in the transverse x and y-axes, based on the reference accelerometers mounted at the head centre of gravity.



**Figure 8-12: Head acceleration results from top of head padded impact surface test**

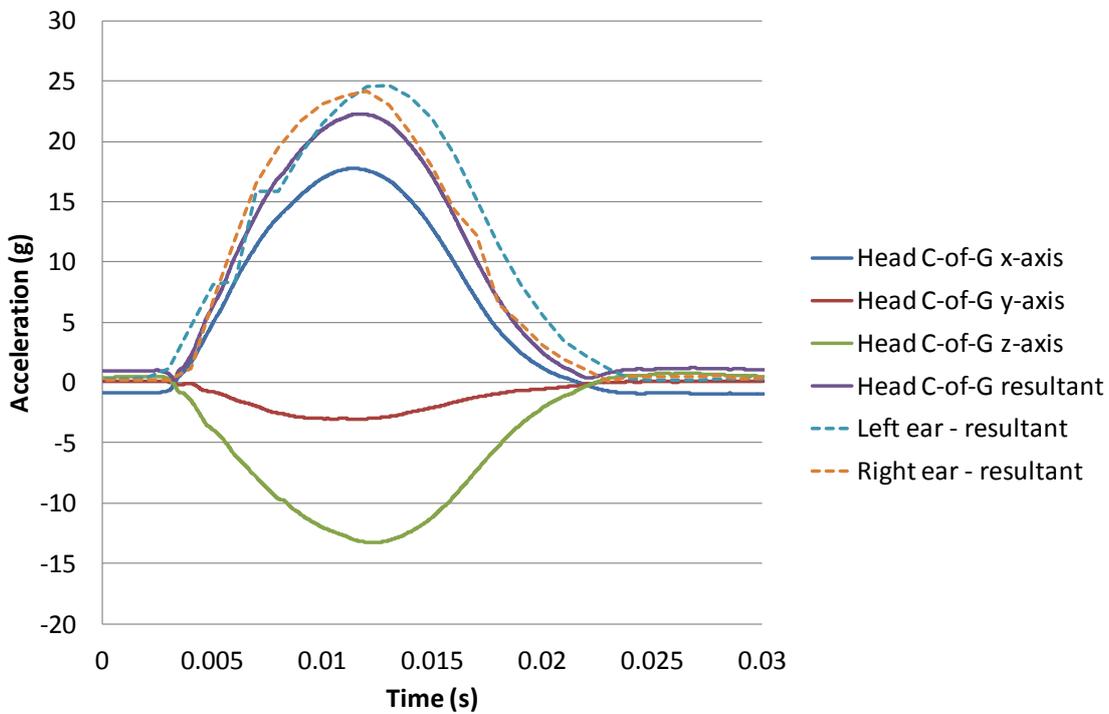
With tests B262D11 to 13, an effort was made to make sure that the point of initial contact was directly under the centre of gravity of the head. Acceleration time-histories from the second of these tests (B262D12) are shown in Figure 8-13.

The rotation of the head impact point forward from the top of the head leads to a different balance of x and z-axis components to the acceleration. Now the x-axis provides the greatest acceleration with a peak value of about 15 g. With a peak resultant acceleration of just over 20 g, this seems to be a slightly more severe impact than the top of the head configuration. That is likely to be a result from there being slightly less rotation of the head following impact with the centre of gravity alignment. Despite efforts to make the tests with the "centre of gravity" alignment the impact type with least rotation upon impact, the results suggests that the forehead alignment was even better (with the highest resultant acceleration values). This may be because of the subjective nature of the centre of gravity alignment and perhaps is also influenced by the profile of the surface of the head (with the irregular curvature causing rotation even with alignment through the C of G).



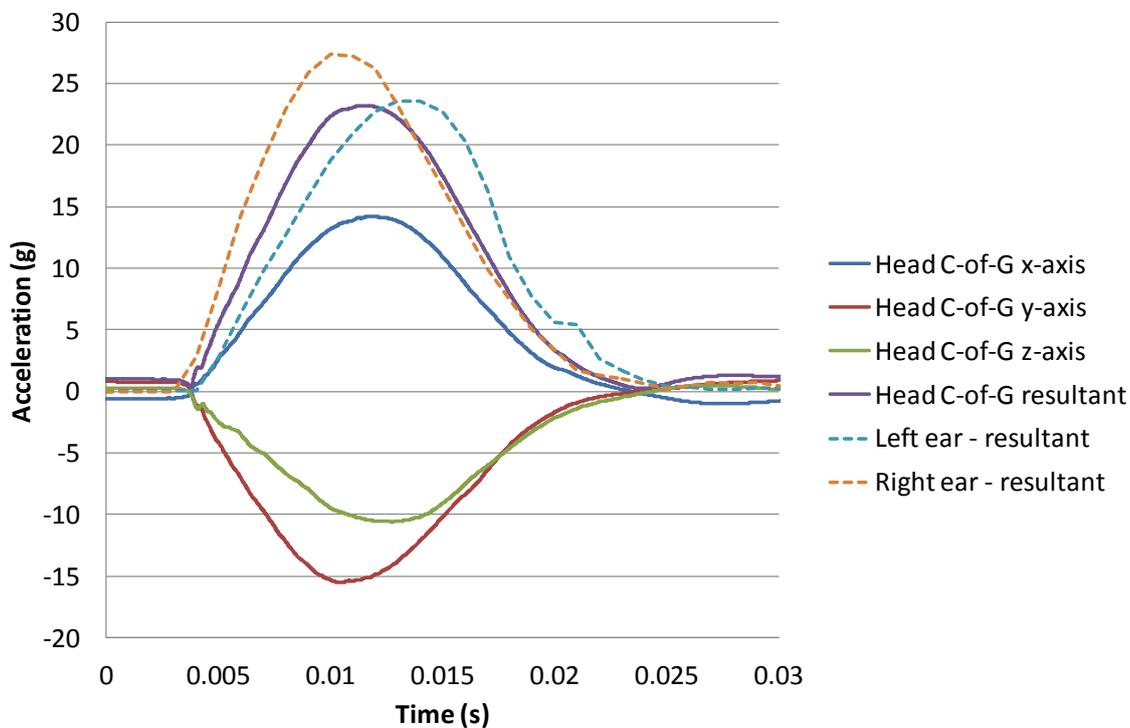
**Figure 8-13: Head acceleration results from centre of gravity padded impact surface test**

Equivalent head acceleration results from a forehead (B262D17) and oblique test (B262D18) are shown in Figures 8-14 and 8-15, respectively.



**Figure 8-14: Head acceleration results from forehead padded impact surface test**

In the oblique head drop test results, as shown in Figure 8-15 for the first of the three tests with this set-up, there was a substantial y-axis component to the head acceleration. At over 15 g the y-axis peak value was greater than both the x and z axis peak values.



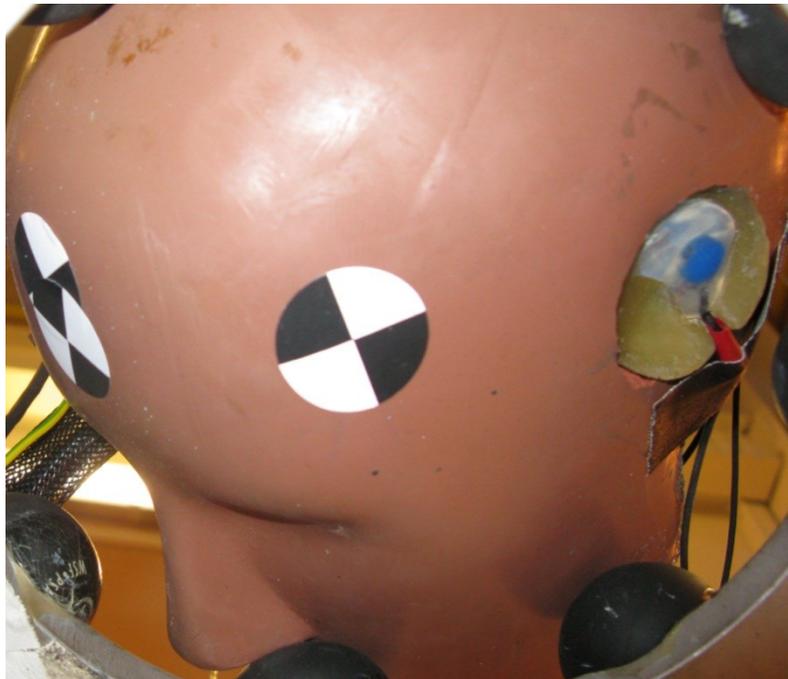
**Figure 8-15: Head acceleration results from oblique padded impact surface test**

The effect of introducing the oblique aspect to the impact was important to the magnitude and phasing of the accelerations measured at the simulated ears of the dummy head. With the impact directed to the left of the forehead (see Figure 8-16) the left ear response was of a lower peak resultant value and delayed in time relative to the right ear. The timing of the reference accelerometers' peak value is between the times at which the peak occurs from the two ear-pieces. This would be expected given that the reference accelerometers are approximately equidistant from the accelerometers mounted in the two ear-pieces. However, it is clear that the resultant acceleration response from either one of the ear-pieces is a less accurate representation of the acceleration at the centre of gravity than was the case in the impacts aligned with the mid-sagittal plane of the head.

Assuming that the measurements from the ear-piece accelerometers can be transformed into the head coordinate system it may be possible to derive the difference in x and z-axis measurements from one measurement site to another. Given that the distance between the ears is known, this would enable some estimate of head rotation to be generated for yaw and roll of the head. The feasibility of obtaining reasonable rotation estimates using such an approach is still to be investigated further.

However, it is also important to appreciate what head rotation can do to the fidelity of data obtained from measurement sites away from the centre of gravity. As mentioned, with measurements from two ears, equidistant from the centre of gravity, it is possible to give an approximate centre of gravity measurement. In this context obtaining measurements from two ears is not data redundancy, but vital for assessing the influence of rotation. With only a single ear-piece sensor, and no other assessment of head rotation, then there is no way to assess if rotation is causing an over or underestimate of the linear acceleration. As an example, in the oblique tests the right ear overestimated the peak resultant acceleration value by 18 to 22 percent; whereas the mean of the right and left ears gave a closer estimate, only 10 to 16 percent greater than the centre of gravity reference values.

Therefore, being able to derive head rotations as well as linear accelerations is not only important in terms of data for new injury criteria development, it is also important with respect to data fidelity applicable to existing linear criteria. As such, having more than one measurement site (e.g. two instrumented ears or an accelerometer array with offset sensors) or incorporating angular motion sensors is a requirement for robust head impact data generation.



**Figure 8-16: Impact point on the dummy head for the oblique tests**

#### **8.2.4 Summary and discussion**

A small series of dummy head drop tests has been carried out with the demonstration in-ear accelerometers provided by the FIA Institute. The ear pieces provided were attached to the dummy via the addition of simulated ear canal holes and epoxy outer ears.

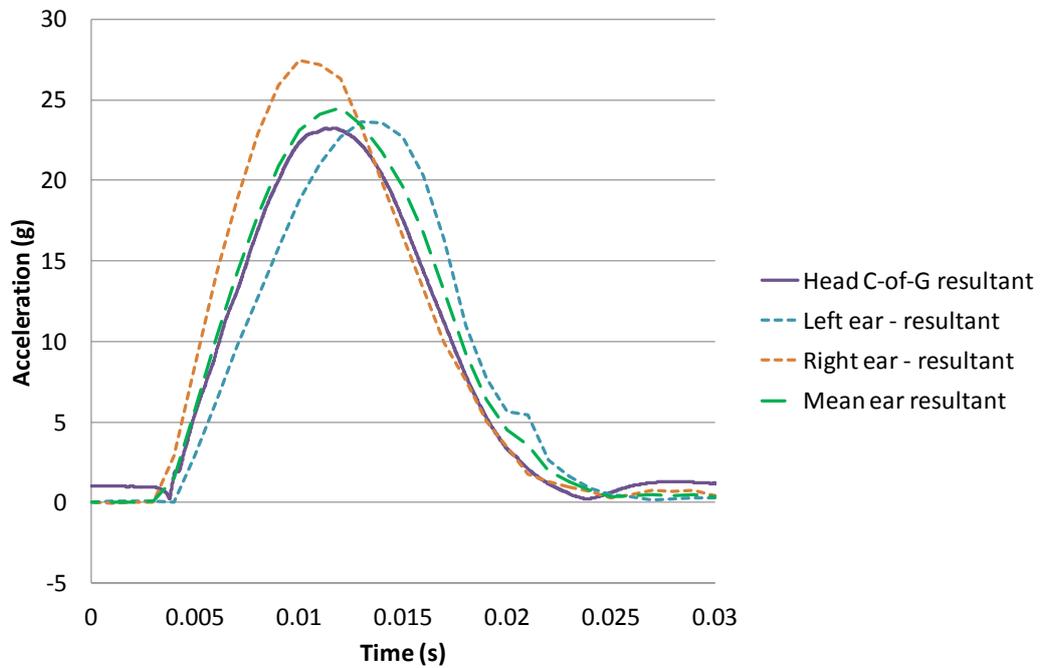
Measurements from the ear-piece mounted accelerometers were compared with reference acceleration measurements provided by conventional accelerometers mounted at the centre of gravity of the dummy head. Resultant acceleration was used to consider peak acceleration, the Head Injury Criterion (HIC) and the acceleration exceeded for 3 ms.

The severity of loading was tuned to give accelerations close to, but within, the maximum measurement range of the demonstration accelerometers of  $\pm 24 g$ . This peak acceleration was accompanied with a loading duration of 20 to 25 ms for the majority of the tests.

In all of the tests, the peak resultant acceleration, HIC and 3 ms values from the ear-piece accelerometers exceeded that from the reference accelerometers. This is slightly surprising as the reference accelerometers are mounted directly to the skull of the dummy head. The ear-piece accelerometers are connected to the dummy head via a hole in the skull or the epoxy outer ear. Whilst these components may be 'rigid' the in-ear accelerometers are still contained within the ear-piece itself (which is of a rubbery constitution). As the ear-piece is deformable it was expected that it would damp or attenuate the accelerations measured within it. This attenuation is not present in the results. One explanation for this behaviour is that the sensitivity, or gain, for the accelerometers is set so as to try and compensate for the position in the ear-pieces. If this is the case, it appears that in this low severity testing the sensitivity may need to be reviewed.

Apart from the over-estimation of the acceleration, the ear-piece results match those from the reference accelerometers well. The duration is similar for all tests. However, it should be noted that the timing of the responses was not synchronised, but was set arbitrarily as part of the processing. This means that if there was any phasing problem with the demonstration accelerometers, then it would have been masked by the process used to generate the results. This is unavoidable in tests of this type where the ear-piece accelerometer 'T0' event can only be set by a level trigger in the crash event data recorder being used to acquire the ear-piece accelerations. There is currently no means of synchronising the ear-piece and another data acquisition system; although the ear-piece data are synchronous from left to right ears.

As was the intention of the oblique impact configuration tests, rotation of the head was shown to generate differences in the accelerations measured on one side of the head or the other. Neither side matched the centre of gravity acceleration as well as in the tests aligned with the mid-sagittal plane of the head. It is suggested that where the loading to the head is away from the mid-sagittal plane, then the mean response from the two ear-piece accelerometers needs to be considered (as shown in Figure 8-17).



**Figure 8-17: Head resultant acceleration results from oblique padded impact surface test**

With loading away from the mid-sagittal plane, there is potential for differences in the measured accelerations to be used in quantifying the head rotation. This potential is recommended for evaluation as a future issue. Any such processing will require the ear-piece accelerations to have been transformed into the head coordinate frame of reference. It is interesting to consider how this could be done in the end application. For instance, perhaps the ear-piece wearer could hold their head in a specified position whilst a short tranche of calibration data is acquired. Without such calibration the possibility of any rotational measurements is severely limited. Furthermore, even resolution to x, y and z axis components will be somewhat arbitrary without calibration. Hence, robust analysis of the acquired data may be reduced to considering only resultant accelerations. On this basis the process of calibrating the orientation of the ear-pieces seems to be an important aspect to address in moving towards the end application. Long-time period averaging of steady-state signals may offer a solution in this regard; otherwise, corroborating information may be necessary.

### **8.2.5 Conclusions**

A small series of dummy head drop tests has been carried out with the demonstration in-ear accelerometers provided by the FIA Institute. The ear pieces provided were attached to the dummy via the addition of simulated ear canal holes and epoxy outer ears.

Measurements from the ear-piece mounted accelerometers were compared with reference acceleration measurements provided by conventional accelerometers mounted at the centre of gravity of the dummy head.

In all of the tests, the peak resultant acceleration,  $HIC_{15}$  and 3 ms exceedance values from the ear-piece accelerometers exceeded that from the reference accelerometers. However, in other aspects such as test-to-test variation, the ear-piece results match those from the reference accelerometers well. The duration and shape of the acceleration responses from the different accelerometers is similar for all tests.

It was noted that the low sampling frequency of the ear-piece accelerometers was evident in the acceleration time-histories. From these results it is not clear whether the sampling frequency will become an issue for the end use of these sensors. It is recommended that this issue is kept in mind during other phases of the in-ear accelerometer evaluation programme.

The results from this testing were analysed principally with respect to the resultant accelerations from each tri-axial sensor. To compare results from each individual axis then a method of coordinate system transformation needs to be developed. For the end use application this transformation will need to take place in a scenario where there is no centre of gravity reference accelerometer conveniently aligned with the head/anatomical coordinate system.

### **8.3 Updated sensor evaluation**

The preceding sections documented a series of dummy head drop tests performed with the demonstration in-ear accelerometers provided by the FIA Institute. Measurements from the ear-piece mounted accelerometers were compared with reference acceleration measurements provided by conventional accelerometers mounted at the centre of gravity of the dummy head.

In all of the tests, the peak resultant acceleration, HIC<sub>15</sub> and 3 ms exceedance values from the ear-piece accelerometers exceeded that from the reference accelerometers. However, in other aspects, the ear-piece results match those from the reference accelerometers well. The duration and shape of the acceleration responses from the different accelerometers is similar for all tests.

However, to compare the response from the reference and in-ear accelerometers it was necessary to shift the data. This time-shift was applied by the author and was made in a subjective manner.

Based on this feedback, the in-ear accelerometer data acquisition system was modified to allow an external trigger system to be recorded. Therefore a test was repeated to check whether the manually applied time-shift was reasonable.

#### **8.3.1 Description of the testing**

As before, the dummy's head was dropped from a specific height to impact either a flat surface, padded with a single (25 mm) thickness of CONFOR™ foam (CF-45, blue) or a rigid concave surface. The head was guided during the free-fall phase and released at impact.

The padded test was to the forehead whereas the impacts with the rigid concave surface were to the top of the head. The drop heights used in the rigid impacts were 100, 150 or 200 mm. A single test was carried out from each height.

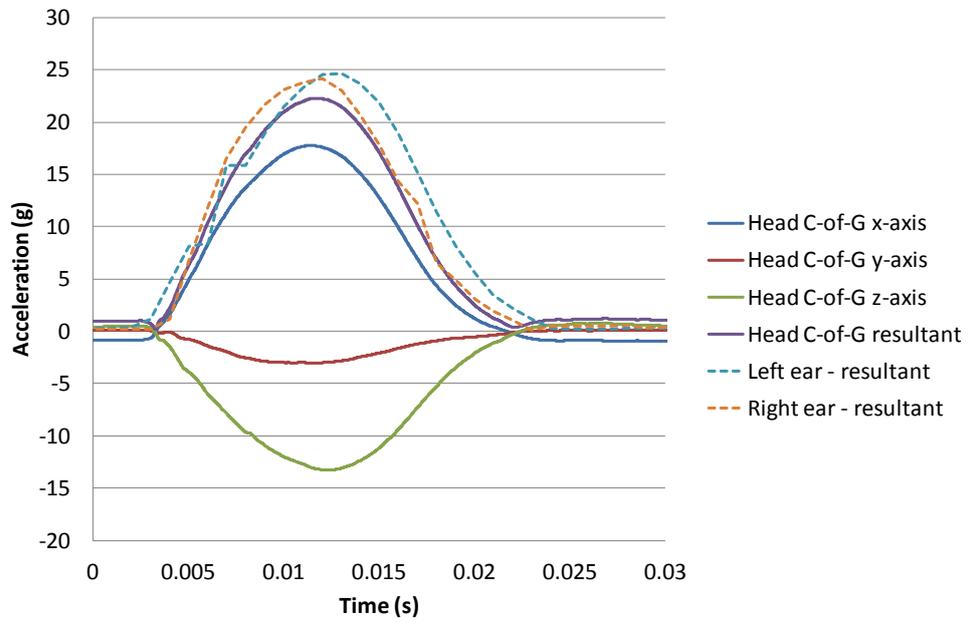
## **8.3.2 Results**

### *8.3.2.1 Padded impact*

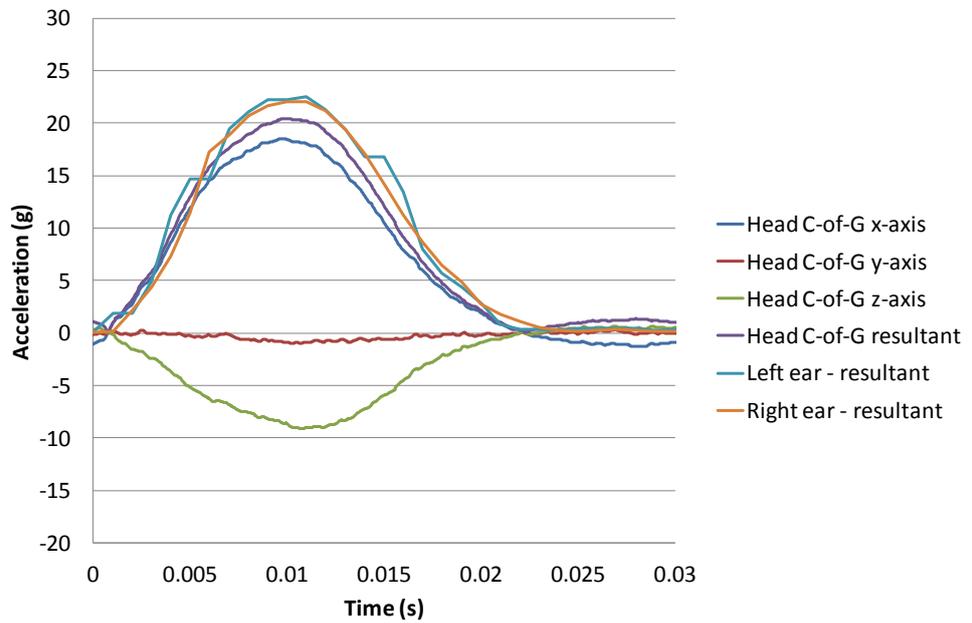
The results from a previous padded test to the forehead (B262D17) are shown in Figure 8-18. The equivalent result from the recent test (B262D27), without the need to time-shift the data, is shown in Figure 8-19. From these two figures, it can be seen that in the second test, the loading to the head was slightly less severe than in the first. The resultant accelerations in Figure 8-19 have a lower peak value than those from Figure 8-18 (and lower  $HIC_{15}$  and 3 ms exceedance values too). Also, there seems to be less y and z-axis acceleration in the second test, which conversely brings the x-axis acceleration closer to the resultant than in Figure 8-18.

Setting these minor differences aside, it can be seen that the timing of the ear-accelerometer responses, as compared with the centre-of-gravity reference, is similar in both figures. There is no temporal offset between the ear and reference accelerometer responses. This suggests that the time-shifting of response data reported previously does seem to be reasonable based on this result generated in a similar severity impact.

It should still be noted that the sampling frequency of the ear-accelerometer data acquisition system is 1 kHz. When considering the signal from the external trigger it is important to remember that the zero-time point as shown at the leftmost edge of Figure 8-19 may have occurred up to 1 ms earlier for the ear-accelerometers.



**Figure 8-18: Head acceleration results from forehead padded impact surface test**

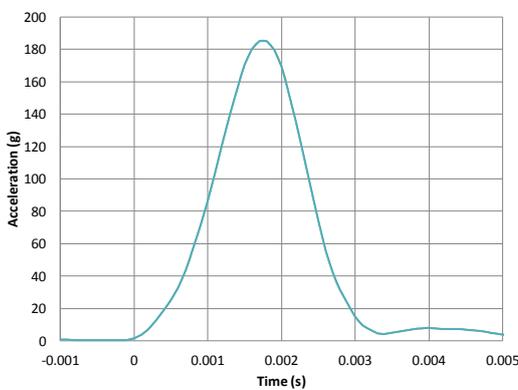


**Figure 8-19: Head acceleration results from forehead padded impact surface test – without time-shifting**

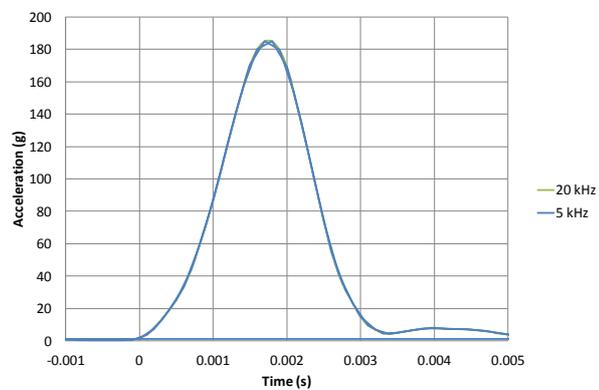
### 8.3.2.2 Theoretical measurements

As the results generated with the ear-piece accelerometers were limited by the 24 g measurement limit it seemed interesting to consider what could be measured with a theoretical accelerometer of an appropriate range but with a reduced sampling frequency. For this purpose the reference acceleration response from a 200 mm drop test onto a rigid concave surface was taken (originally sampled at 20 kHz). This was plotted against alternative curves where the data had been artificially sampled at lower frequencies of 10, 5, 2 and 1 kHz. The resulting graphs are shown in Figures 8-15 to 8-23.

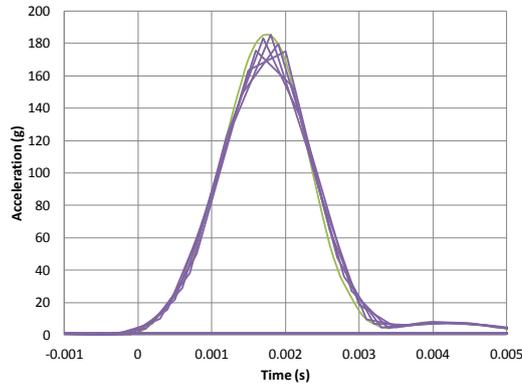
Where the sampling frequency was reduced, it presented options as to exactly which time point to take the measurement. Each figure below shows the different options for each sampling frequency on the same graph. This is why it seems as though there are several lines crossing each other. In effect two lines are plotted for the 5 kHz response, five lines for the 2 kHz and ten lines for the 1 kHz response.



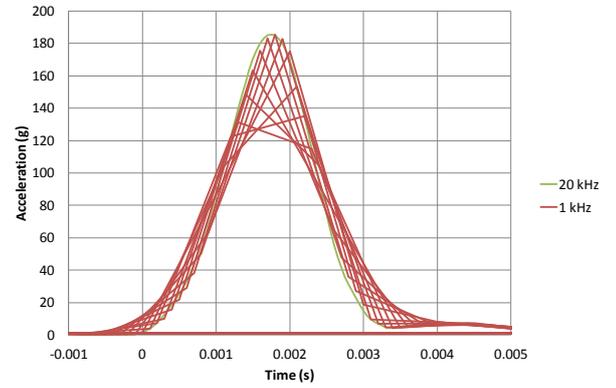
**Figure 8-20: Head acceleration results from top of head rigid concave impact surface test – 200 mm drop height – 20 kHz and 10 kHz sampling**



**Figure 8-21: Head acceleration results from top of head rigid concave impact surface test – 200 mm drop height– 20 kHz and 5 kHz sampling**



**Figure 8-22: Head acceleration results from top of head rigid concave impact surface test – 200 mm drop height– 20 kHz and 2 kHz sampling**



**Figure 8-23: Head acceleration results from top of head rigid concave impact surface test – 200 mm drop height– 20 kHz and 1 kHz sampling**

This sequence of figures shows that some deviation to the duration of the loading event can be introduced on the basis of a reduced sampling frequency. Also, the peak value measurement can be severely underestimated depending on the sampling rate and exactly when the measurement is sampled.

### **8.3.3 Conclusions**

Drop tests have been performed with a Hybrid III head fitted with demonstration FIA Institute ear-piece accelerometers which are similar to those carried out in Section 8.2.

In a low severity flat padded surface impact a new trigger mechanism was tried for the ear-piece data acquisition system. The result of using this trigger suggests that the previous results reported with manual time-shifting of the response are still appropriate. There is no substantial offset between the applied acceleration and the output from the ear-pieces.

Theoretical reduction in the sampling rate suggests that 1 kHz sampling will inevitably lead to a reduction in the measured peak acceleration and some deviations in

response around the onset and end of the acceleration event. Application of a low-pass filter (as with the 780 Hz cut-off implemented with the ear-piece accelerometers) is also likely to influence the ability of a sensor system to capture short duration, high frequency events.

## **8.4 Revised sensor evaluation**

Since issuing the results from the initial prototype accelerometers to the FIA Institute, they have been able to make two alternative revised sensors available for evaluation.

1. The first of these was an equivalent sensor with a maximum measurement range of 24 *g*. However, rather than generating a digital output signal, the sensor output to the Event Data Recorder was analogue. The advantage of going with an analogue system would be that a faster sampling frequency was available; measuring at 5 kHz instead of the 1 kHz available with the digital sensors. The disadvantage would be in the noise that could be picked up between the sensor and the analogue to digital convertor.
2. The second was a similar digital system as tested previously, but with a maximum sensitive range of up to 400 *g* instead of 24 *g*.

Both of these new prototypes were evaluated in a smaller set of experimental head drops. Again, the Hybrid III head was used to mount the accelerometer earpieces for all of the tests. Results are presented for the digital system. My observations regarding the analogue system are below.

### **8.4.1 24 g analogue accelerometer**

Based on the analogue system made available to TRL for evaluation, it was only possible to evaluate the data from either the right or left earpiece at any one time.

One fundamental thing that was noticed in the processing of the analogue system data was that the sampling frequency was not constant between all the data points. Due to the way the analogue output was coded into the CAN data packets for the EDR, the sampling frequency flipped between either 2.5 or 5 kHz.

Based on the initial results with the analogue and digital systems, it is already evident that there are differences between sensor types.

With the analogue accelerometers, they offer a higher rate of sampling. This may give a better initial (rise-time) synchronisation with the reference accelerometer response. The higher frequency of sampling will allow closer following of reference response shape. However, in drop tests the exact peak acceleration value still shows limited resolution. Also, presumably due to noise in the signal, there appear to be step jumps away from the reference value. There are also, often different durations for left to right earpieces, which could be a demonstration of the typical event to event repeatability that is possible with those sensors.

Furthermore, concerns arose over the viability of the analogue system in general. It was already showing symptoms of noise on the signal even in the laboratory environment. This could be substantially noisier when used in an open cockpit racer.

Any sampling system needs to log both ears at the same time. It is also suggested that the final sampling system needs to record at a constant frequency, rather than having to switch between 2.5 and 5 kHz. At the moment the system takes 3 out of 5 measurements from 5 kHz to create a 3 kHz sample.

Finally, the recreation of the analogue signal requires post-processing, which is not needed with the digital output, which is therefore handled more easily.

These issues would need to be resolved before an analogue system could be implemented. In fact, I proposed that this should be addressed before further validation work was undertaken with the analogue system.

#### *8.4.1.1 400 g digital accelerometer*

With the updated digital accelerometer system, the maximum sensitive range of the sensors was increased to  $\pm 400 g$ . However, unfortunately, other aspects of the prototype earpieces had not been updated. Therefore, the cable lengths provided with the earpieces limited the extreme drop heights that could be used.

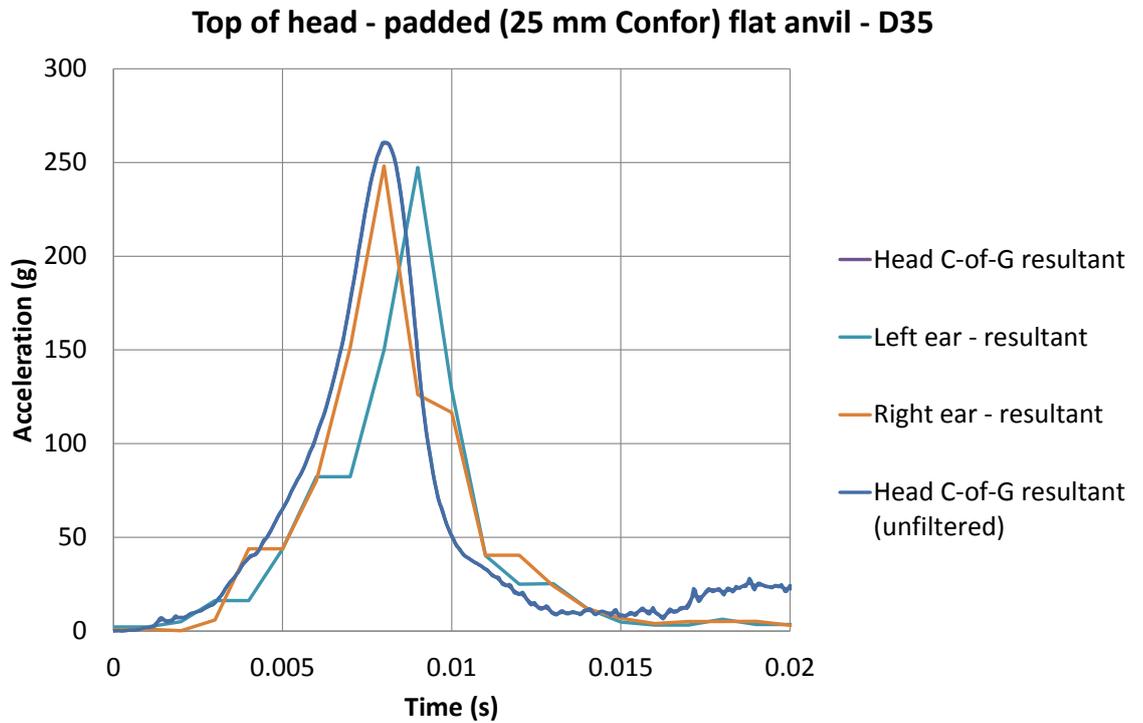
Even with such limitations, it was possible for me to achieve a drop height of 2.5 m, this produced an impact speed of 6.6 m/s. For the digital accelerometers with the increased drop height, the 130 mm diameter flat impact surface was padded with 25 mm Confor (CF45 blue) foam padding. The severity of loading was chosen to reflect typical responses for a helmeted head in standard drop test conditions (reaching a peak acceleration of 100 to 200 *g* in about 3 ms).

The impacts were aligned with the top of the head. During the first test, some rotation of the dummy head after the initial contact was noticed. A second test was carried out which gave a better alignment of the centre of gravity of the head with the impact point, and hence less rotation.

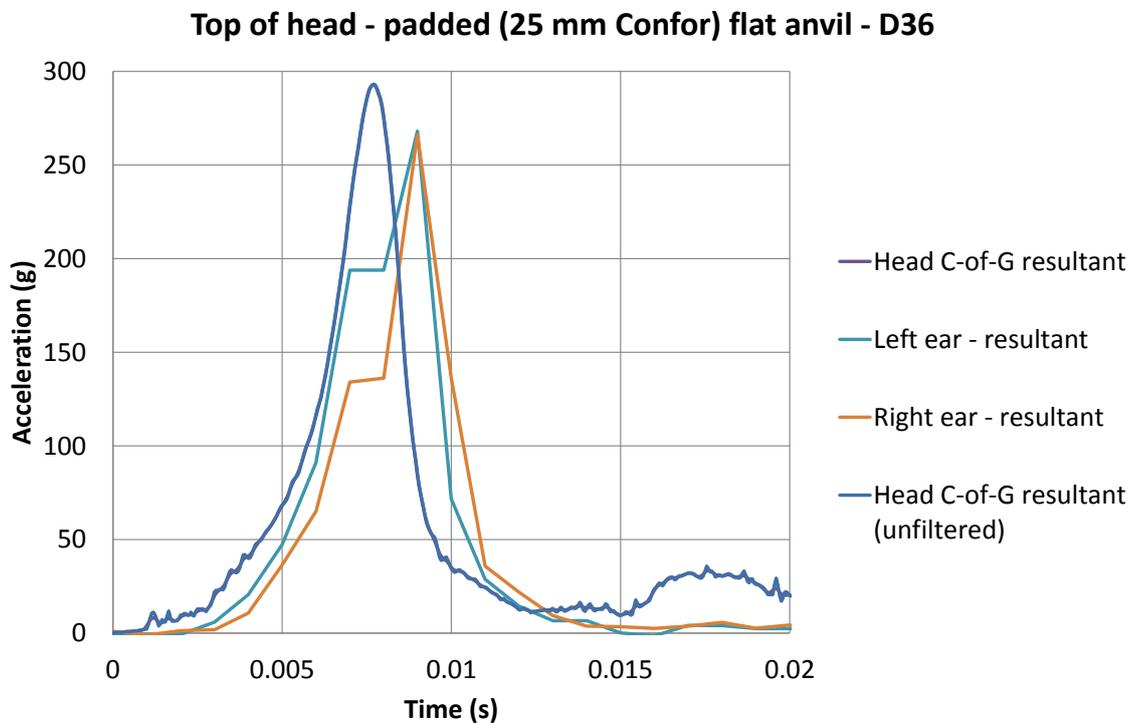
The peak resultant acceleration, HIC<sub>15</sub> and 3 ms exceedance values from these two tests are shown in Table 8-6. Due to an issue with the processed data, HIC and 3 ms exceedance values for the ear-pieces were only generated from the left ear data.

**Table 8-6: Top of head – padded (25 mm Confor) flat anvil tests; peak acceleration, HIC and 3 ms exceedance values**

Test	Peak resultant acceleration ( <i>g</i> )			Head Injury Criterion (HIC)		Peak 3 ms exceedance ( <i>g</i> )	
	Centre of gravity	Left ear	Right ear	Centre of gravity	Left ear	Centre of gravity	Left ear
<b>B262D35</b>	261	249	250	1611	1091	114	102
<b>B262D36</b>	293	266	266	1879	1804	106	150



**Figure 8-24: Impact to the top of the dummy head, dropped from 2.5 m onto flat surface padded with 25 mm depth of blue Confor, Test 1**



**Figure 8-25: Impact to the top of the dummy head, dropped from 2.5 m onto flat surface padded with 25 mm depth of blue Confor, Test 2**

The results from these two tests with the higher acceleration limit accelerometers demonstrate both the positive and negative aspects of this prototype earpiece design.

In the first test the alignment of the timing of the response was good for the right earpiece. The peak value was also close to that measurement with the reference accelerometers (with less than a five percent difference in peak values).

From the second test the timing of the peak from the earpiece accelerometers occurs late with respect to the reference accelerometers (9 ms versus 7.7 ms). Also the magnitude of the peak is underestimated by more in the second test than in the first (~9 %). Although, this magnitude still represents a value in excess of the reference accelerometer half a millisecond before or after the peak value (i.e. closer than the worst case sampling at 500 Hz or 1,000 Hz; the frequencies of the low pass filter and sampling rate of the earpiece accelerometer, respectively).

In either case, the shape of pulse is clearly limited by the resolution in acceleration and time. This affects the assessment of the Head Injury Criterion (HIC) and 3 ms exceedance, which differ between the ear-piece and the centre of gravity reference by more than the peak value alone. In the first of the two tests, the ear piece HIC is 68 % of the centre of gravity value and in the second test, the 3 ms value is overestimated by 41 %. Finally, the differences between the two tests, and in particular how well the earpiece response matches that from the centre of gravity of the head, illustrates the potential variability due to the sampling rate. As mentioned in Section 8.3.2.2, with a sampling rate of just 1 kHz for impact events leading to head loading with a duration of less than 10 ms, then there is an inevitable chance that the absolute peak value is not measured.

## **8.4.2 Summary**

### *8.4.2.1 Findings regarding existing prototypes*

Accelerometer calibration-type shaker table tests were carried out with the miniature accelerometer purchased by the FIA Institute for consideration as in-ear instrumentation. This required mounting the accelerometer in a bespoke rapid prototyped rigid block.

A substantial decrease in the measured accelerations from the miniature accelerometer was observed when compared with reference accelerations. This decrease became larger progressively as the frequency of oscillations was increased from about 200 Hz to 2 kHz. This frequency sensitivity is likely to affect the acceleration measurements of short duration impact events.

Transverse sensitivity was also noted with the miniature accelerometer, which exceeded the ISO 6487 transducer requirement.

A small series of dummy head drop tests has been carried out with the demonstration in-ear accelerometers provided by the FIA Institute. The ear pieces provided were attached to the dummy via the addition of simulated ear canal holes and epoxy outer ears.

In all of the initial tests, the peak resultant acceleration,  $HIC_{15}$  and 3 ms exceedance values from the ear-piece accelerometers exceeded that from the reference accelerometers. However, the duration and shape of the acceleration responses from the different accelerometers is similar for all tests. In later tests with an updated 400 g digital sensor, the earpiece accelerometers then underestimated the peak acceleration value and were less accurate in HIC and 3 ms exceedance values.

In high severity, short duration acceleration events, differences can now be seen between the ear-piece responses and those from the conventional reference accelerometers.

Theoretical reduction in the sampling rate suggests that 1 kHz sampling and application of a low-pass filter will inevitably lead to a reduction in the measured peak acceleration and some deviations in response around the onset and end of the acceleration event. The influence of the frequency response of the underlying sensor was not estimated or established.

The extent of the differences caused by the low sampling rate will depend on the frequency, the duration and magnitude of the loading applied to the head and also how well coupled the earpieces are to the skull.

The results from this testing were analysed principally with respect to the resultant accelerations from each tri-axial sensor. To compare results from each individual axis then a method of coordinate system transformation needs to be developed.

#### *8.4.2.2 Requirements for additional research*

So far this evaluation of the FIA Institute accelerometer prototypes has focused on either rigid mounting of the accelerometers in a solid block, or having the earpieces mounted in a rigid dummy skull surrounded with a stiff epoxy 'ear'. In this way the coupling of the sensitive element to the loading device should be very good compared with the softer human ear. Therefore, what is missing from a complete evaluation of the sensors is some idea as to the performance of the system in a human ear. For this reason, it is hoped that PMHS testing will be carried out in the future to address this missing information.

#### *8.4.2.3 PMHS testing*

To make in-ear accelerometers acceptable for use in high-end motorsport divisions, it is considered necessary to have the accelerometers mounted within the communications ear-pieces already being worn by, for example, racing car drivers.

Evaluation of accelerometer ear-pieces in a dummy headform constitutes a best case for the ear-piece. To make a receiving orifice in the dummy head, the skull was drilled and the surrounding flesh cut-away. Also an ear representation was made out of a stiff epoxy. Therefore when the earplug is put into the drilled hole and epoxy ear it has the most rigid connection with the skull that could be possible. This minimises the relative motion of the accelerometers and the skull hence producing the best correlation between the measured accelerations in the ear and with the reference accelerometers rigidly attached to the skull and mounted at the centre-of-gravity of the head. In a human the interface between the ear-piece and the ear canal will be less rigid. There will always be some compressible tissue lining the ear canal that limits the coupling of the ear piece to the skull.

Knox et al. (2008) tested an earlier prototype of an in-ear accelerometer with the ear-piece placed in a simulated rubber ear. They evaluated the system in vibration and impact tests. In their conditions they found that the rubber ear caused a phase shift between the ear and reference accelerations and also raised the measured amplitude above the reference peak. Knox et al. describe a method for trying to correct for the error caused by the phase lag.

The correction required for a human ear is likely to be different from both a rubber ear of the type used by Knox et al. and a rigid dummy ear. Therefore the need for and magnitude of phase-lag corrections can only really be described via human subject testing.

With living humans, they are unlikely to volunteer for higher severity impact tests and it would be ethically challenging to administer such tests. Also it is difficult to provide reliable reference acceleration data. Instead to generate the necessary comparison data between the human ear and the reference skull accelerations Post-Mortem Human Subject (PMHS) tests are required.

Begeman et al. (2006) have already evaluated the frequency response and coupling of earpiece accelerometers to the human skull using PMHS and a previous design of accelerometer. Data from vibration and drop tests were analysed to determine the effective frequency response of the earpiece package when mounted in the ear canal. It was the intention of this testing to define the impact conditions where the earpiece accelerometers could be expected to give accurate data.

The testing by Begeman et al. used three PMHS heads. Two types of earpiece material were evaluated a softer material and that provided by EarEverything and Sensaphonics as would be used by racing car drivers.

The exact vibration tests varied between the heads but together consisted of 10, 30, 50, 70 or 100 Hz oscillations in each of the three axes: x, y and z. Drop tests were conducted either without a helmet leading to a 40 *g* peak acceleration or with a helmet in tests with a peak acceleration of 100 *g*.

The sinusoidal testing showed that the earpieces worked well at low acceleration levels and particularly in the y-axis direction. Significant overshoot was observed in the earpiece responses at 30 Hz in the x and z-directions. At higher frequencies this overshoot increased relative to accelerometers mounted rigidly to the skull. The earpieces made from a softer material had a greater tendency to overshoot than the Sensaphonics earpieces.

In the drop test, impact simulations, the ear accelerations were reported to correlate well with the reference accelerometers with some overshoot or undershoot. However, Begeman et al. noted that off-axis accelerations could give large spurious accelerations. They suggested the use of a stiffer and less massive earpiece material to improve the coupling with the ear.

It is hoped that the prototypes now being developed for use by the FIA Institute will address some of these former issues.

### **8.4.3 Conclusions**

Shaker table frequency response tests and drop tests with new prototype miniature accelerometers have been undertaken.

The results from these tests have been sent on to the FIA Institute who commissioned the work and the prototypes.

Based on the findings and other considerations the following course of action has been proposed:

- The 1000 Hz digital system was now commercially available.
- The 5000 Hz analogue system would require substantial additional investment and development time before being fit for use as an in-ear sensor for motorsports applications.
- The FIA Institute Group agreed to proceed with the 400 *g*, 1000 Hz digital version.
- It was proposed to run some human subject validation tests at Wayne State University, so far this work has not been commissioned.

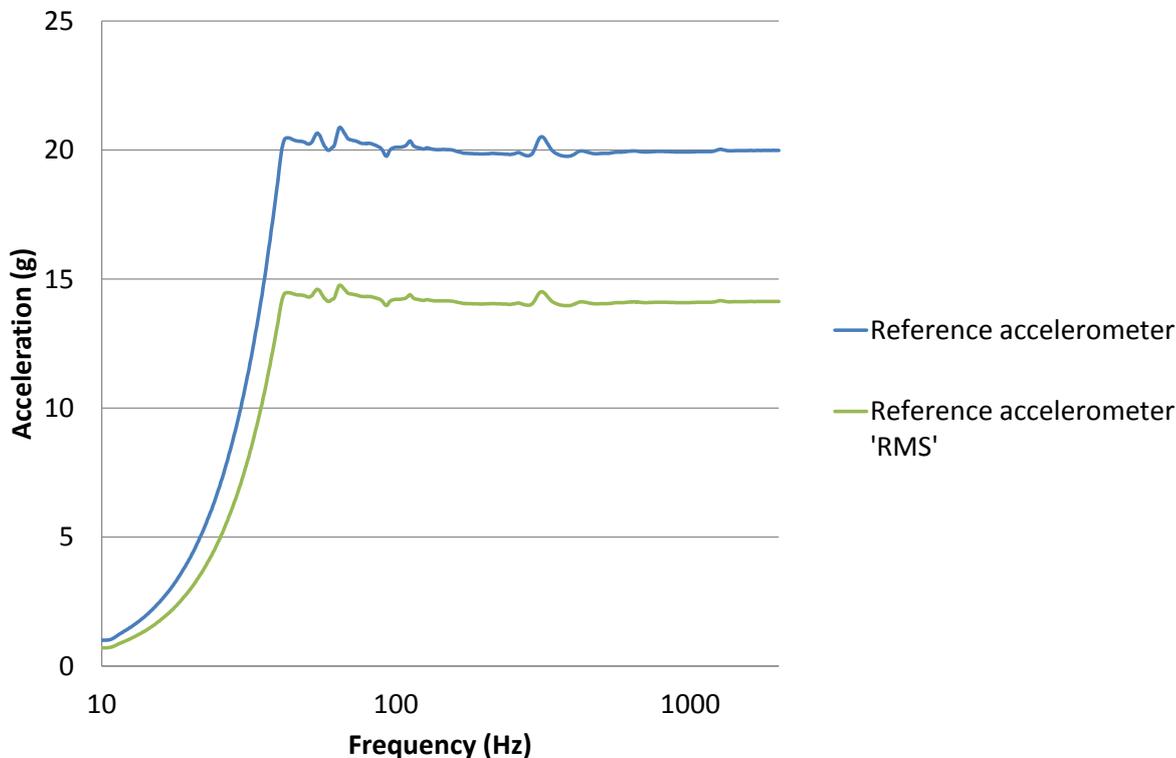
## **8.5 Second phase of shaker table trials**

### **8.5.1 Introduction**

With an update to the miniature accelerometers to be used in the ear-pieces in Formula 1, the opportunity was taken to repeat the frequency response tests performed with earlier prototypes.

An example of an unmoulded accelerometer pair was obtained. One accelerometer was rigidly bonded into a mounting block so that it could be glued to the shaker table. For each test, a sinusoidal input was applied; the frequency was swept from 10 Hz to 2,000 Hz, with the amplitude rising to 20 *g* at 40 Hz and being maintained at that level until the end of the frequency sweep. The root, mean, square (RMS) acceleration level for a peak amplitude of 20 *g* is 14.14 *g*. The applied pulse for one of the tests is shown in Figure 8-26. The total sweep took 30 seconds with the frequency increasing

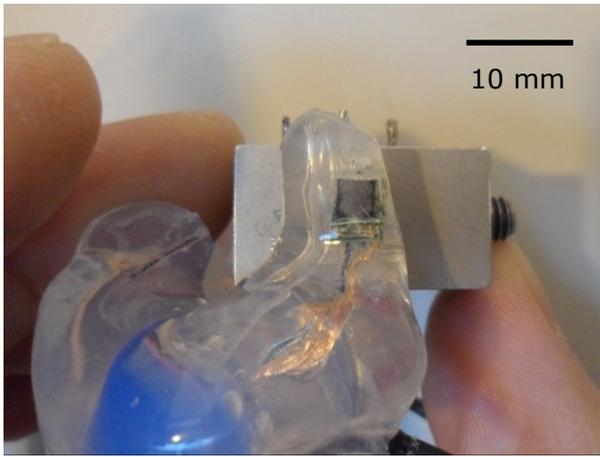
exponentially throughout that time. It should be noted that the RMS line was calculated as a proportion of the peak amplitude rather than being a smoothed function of the temporal acceleration.



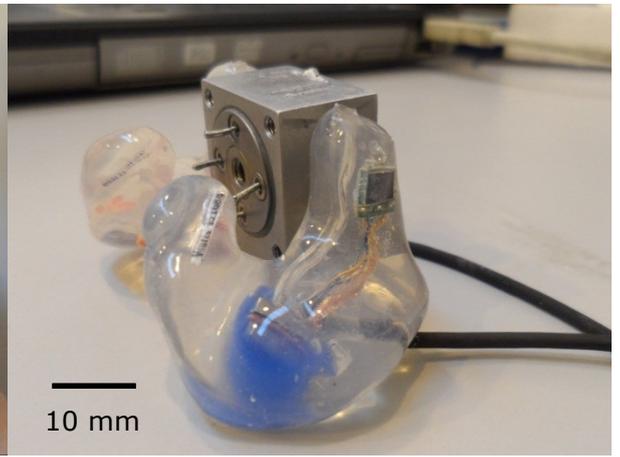
**Figure 8-26: Nominal applied acceleration for frequency response tests**

The accelerometer was tested with each of its sensitive axes aligned with the direction of oscillation of the shaker table, producing results for the x, y and z-axes.

After testing the rigidly mounted accelerometer, demonstration ear-pieces were glued to the shaker table via a rigid block and also tested following attempts to align the three orthogonal axes with the direction of oscillations. Figures 8-27 and 8-28 show the ear-pieces glued onto the rigid block for mounting on the shaker table, prior to testing.

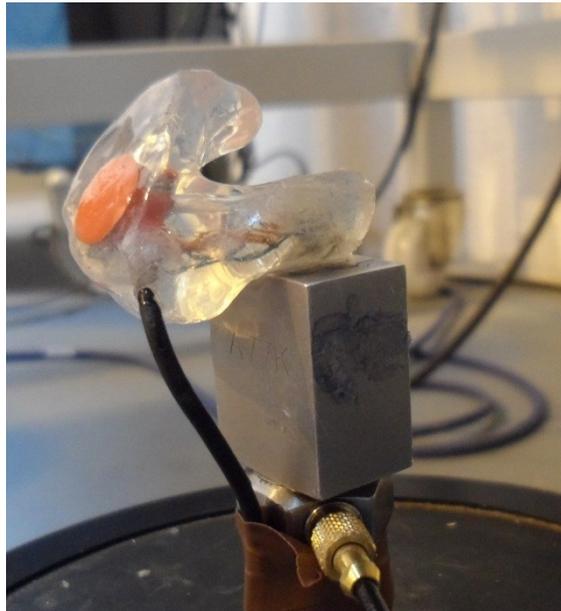


**Figure 8-27: Ear-pieces glued to block for x-axis frequency tests**



**Figure 8-28: Ear-pieces glued to block for y-axis frequency tests**

Due to the constraints of fitting an ear-piece on top of the shaker unit, for the z-axis test, only the right ear-piece was used, not both the right and left (Figure 8-29).



**Figure 8-29: Right ear-piece glued to block for z-axis frequency tests**

For a final investigation, the other unmoulded accelerometer was set into a wider rigid cube with a gap of about 1 mm around it. This gap was then filled with a silicone rubber compound with a Shore A hardness of around 10 (Figure 8-30). The silicone surround was applied with two objectives in mind:

1. Firstly, it was considered that a millimetre of rubber might approximate the skin around the ear-canal and offer some understanding of the difference between mounting an accelerometer directly to the skull and placing it in the ear-canal.
2. Secondly, it has been suggested that adding a small amount of silicone around the accelerometer chip in an event data recorder may isolate it from high frequency noise which could otherwise mask underlying acceleration data.

It is accepted that a cube of material surrounding the accelerometer would allow different vibrational modes to that of an approximately cylindrical ear canal moulding. However, it was thought that this would still provide a useful contrast against the rigid mounting to block and the full ear-piece.



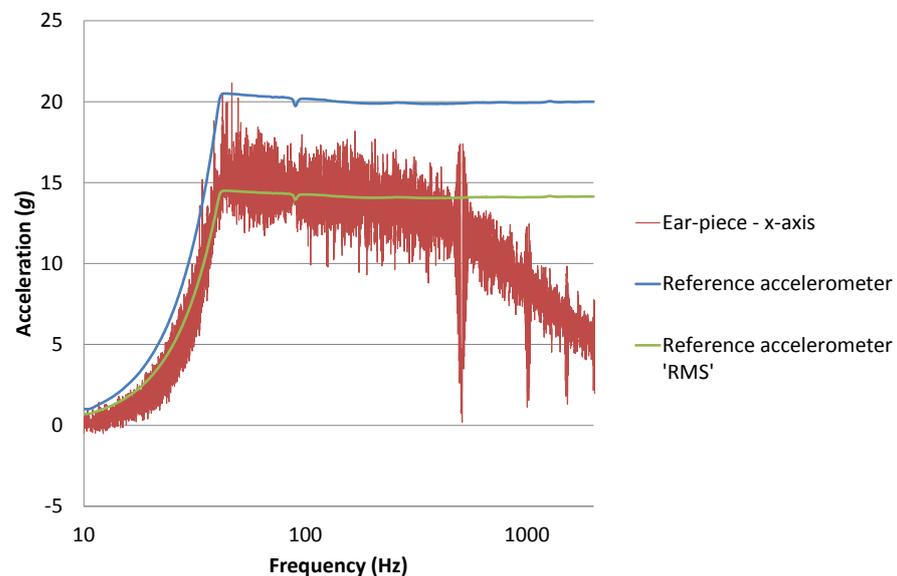
**Figure 8-30: Accelerometer mounted within rigid block (8 mm sides) but surrounded by approximately 1 mm of rubber**

## 8.5.2 Results

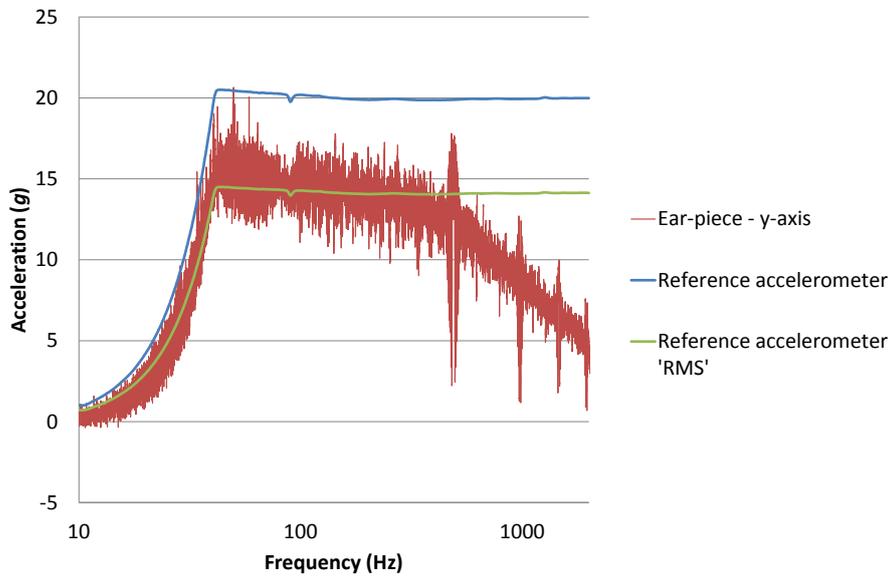
### 8.5.2.1 Rigid mounting

The results from the frequency sweeps with the accelerometer mounted rigidly are shown in Figures 8-31 to 8-33, for the x, y and z-axes, respectively.

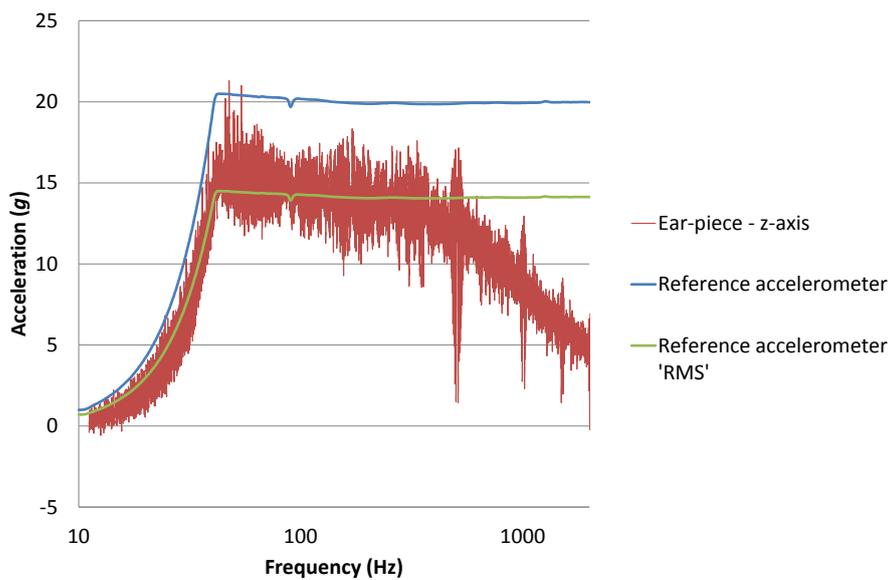
With the accelerometer mounted rigidly, without any soft rubbery compounds around it, this configuration represents the condition where the closest possible match can be expected between measured and input signals. This is likely to be the optimum condition for the miniature accelerometer frequency response.



**Figure 8-31: Frequency response for x-axis of accelerometer when mounted rigidly to the shaker table**



**Figure 8-32: Frequency response for y-axis of accelerometer when mounted rigidly to the shaker table**



**Figure 8-33: Frequency response for z-axis of accelerometer when mounted rigidly to the shaker table**

These figures are extremely similar from one alignment to the next. This suggests that there is no fundamental change in frequency sensitivity from one axis to another for the accelerometer device when mounted rigidly. This is slightly unexpected as the microelectromechanical system (MEMS) accelerometer is, in essence, a two dimensional array of strain sensitive elements.

Throughout all frequencies and the three sweeps, the response from the miniature accelerometer is variable. It appears to be a noisy response when compared with the applied acceleration. For these curves, only twelve samples were used to generate the 'root, mean, square' value. If more values had been used then a smoother behaviour could have been generated. However, resolution would have been lost when considering the frequencies at which resonance is observed.

Resonant, or counter-resonant frequencies can be seen at multiples of 500 Hz. This is to be expected as the sampling frequency for the accelerometer unit is 1,000 Hz. Therefore, each sample is taken at a similar or opposite position in the sinusoidal impulse.

From 10 to about 200 Hz the miniature accelerometer values match closely the reference acceleration. In this frequency range the accelerometer unit seems accurate.

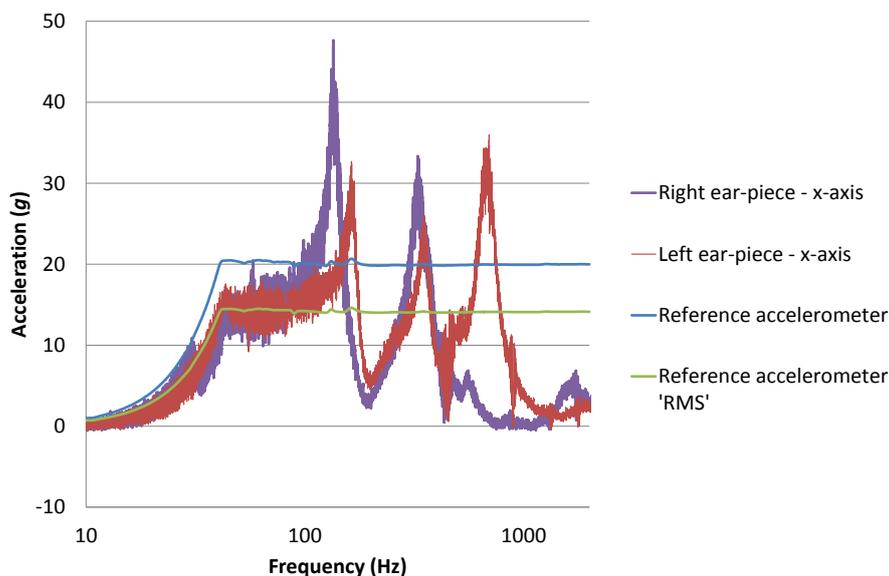
After about 200 or 300 Hz there is a clear reduction in the miniature accelerometer values, when compared with the reference acceleration level. By 1,000 Hz the miniature accelerometer values are about half of the reference acceleration and this ratio continues to decrease until the end of the swept frequencies.

#### 8.5.2.2 *Ear-pieces*

The results from the frequency sweeps with the ear-pieces attached to the shaker table are shown in Figures 8-34 to 8-36, for the x, y and z-axes, respectively.

Unlike the rigidly mounted accelerometers, the three axes for the ear-pieces are not very similar in response. This is not surprising as the shape of the ear-pieces is not equal in all three dimensions, with varying amounts of the rubber moulding between the accelerometer and the mounting block in the different arrangements and also all around the accelerometer unit. The particular shape and depth of material will allow different vibrational oscillations and bending modes to be excited throughout the frequency range.

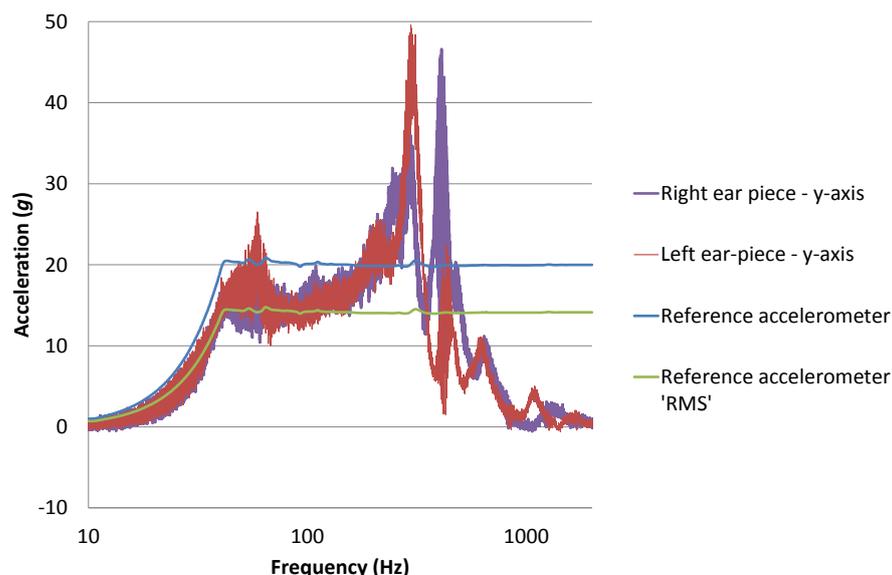
The x-axis results (Figure 8-34) show three primary resonant frequencies at 130-155, 330-350 and 680 Hz. Interestingly, the last resonance is seen only in the left ear-piece and not in the right.



**Figure 8-34: Frequency response for x-axis of accelerometer with ear-pieces mounted to the shaker table**

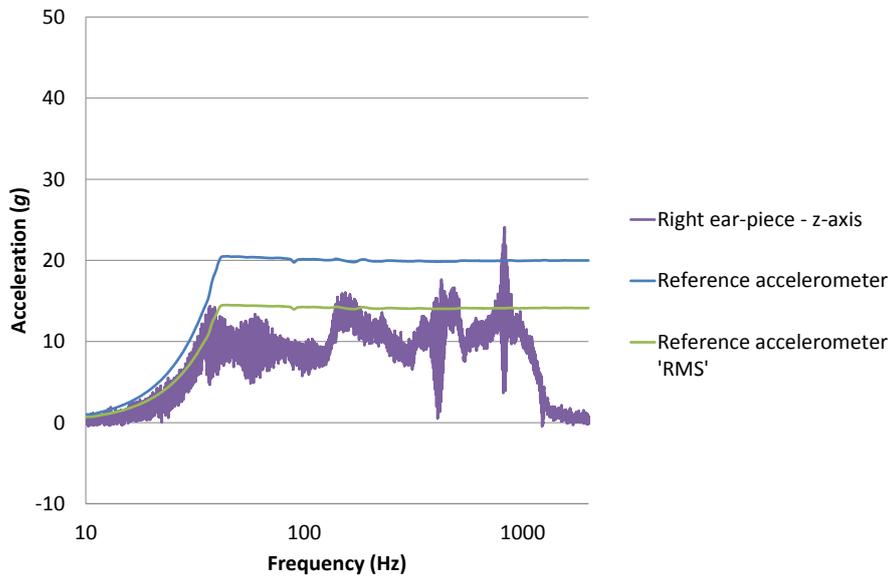
A similar trend as with the x-axis ear-piece results is shown also in the y-axis (Figure 8-35). In this case, the resonant frequencies are at 285 and 400 Hz. These results therefore indicate that the measured acceleration may not be dependent only on the orientation of the ear-piece around the accelerometer, but also on other variations from one side (ear) to the other.

It should be noted that the ear-pieces are moulded to fit specifically the ear of the 'driver' or wearer. As ears are not perfectly symmetrical then there would be no reason to expect perfectly symmetrical results. Instead, these data indicate the sensitivity of the results to small changes in the shape and depth of the surrounding material. Over about 100 Hz there can be substantial deviation of the ear-piece accelerations from the reference. Hence the response of the ear-pieces could be somewhat misleading in events lasting less than 10 ms.



**Figure 8-35: Frequency response for y-axis of accelerometer with ear-pieces mounted to the shaker table**

For the z-axis response, shown in Figure 8-36, the ear-piece measurement was almost always below that of the reference accelerometer. The exceptions are due to a few resonant frequencies, notably with a peak just below 1 kHz. Based on the rigidly mounted accelerometer results, we know that there is no systematic difference in the three sensitive axes when the ear-piece is excluded. Therefore the data shown in Figure 8-36 seem to reflect the different mounting of the ear-piece to the shaker table. For the z-axis test, the ear-piece was glued to the top of the mounting block, whereas for the x and y-axes the ear-pieces were glued to the side of the block. It seems that the different connection orientation may have provided a different link with the excitation oscillations; i.e. the mounting on top providing some mechanical damping (through the compressive loading) not seen when mounted on the sides.



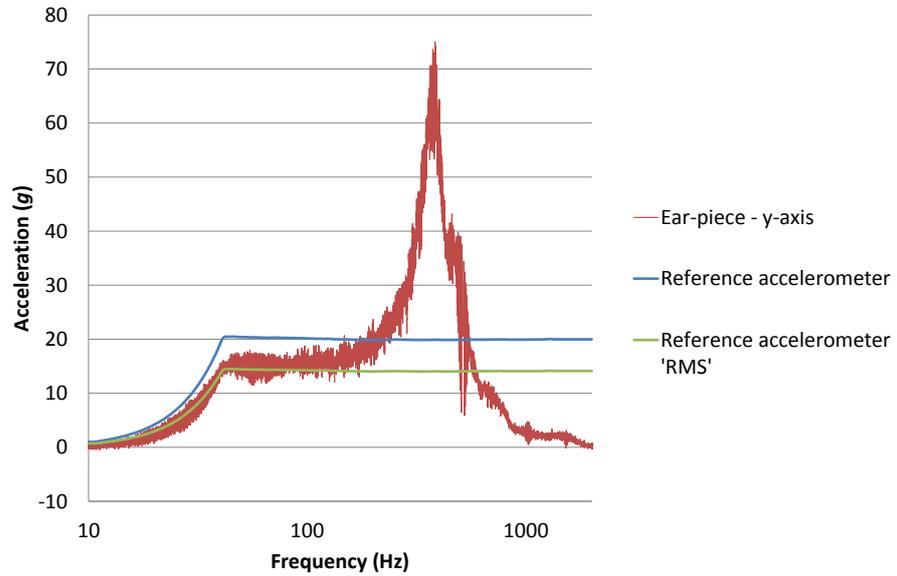
**Figure 8-36: Frequency response for z-axis of accelerometer with right ear-piece mounted to the shaker table**

#### 8.5.2.3 *Silicone spacing*

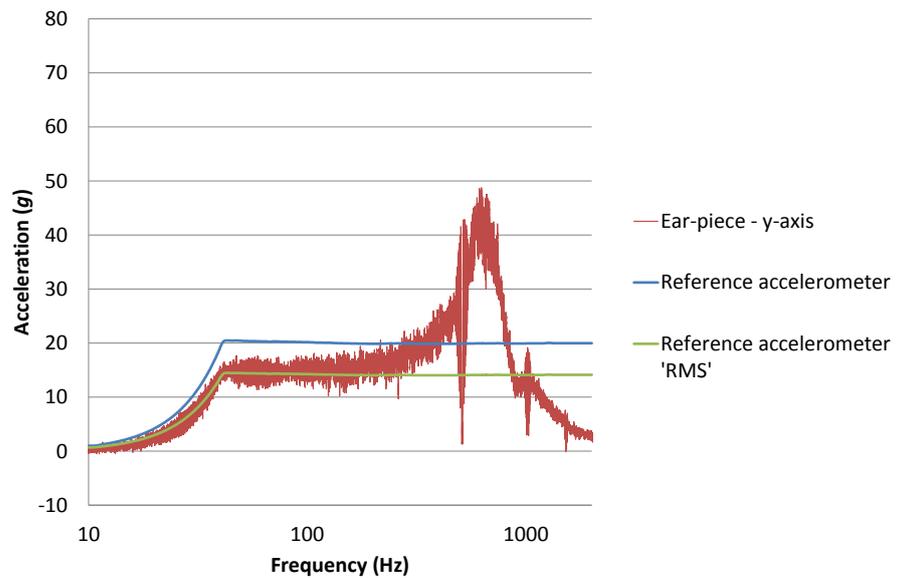
The results from the frequency sweeps with an accelerometer mounted in about 1 mm of silicone rubber within a small block which was rigidly attached to the shaker table are shown in Figures 8-37 and 8-38 for the y and z-axes, respectively.

Again in this installation configuration, as with the rigidly mounted accelerometer, the measured acceleration matches accurately that of the reference accelerometer up until about 200 Hz. From that frequency onwards, up until 550 to 800 Hz depending on the axis, the measured acceleration is greater than the input. This is due to the vibration of the silicone.

With a duration of about 5 ms, the head drop tests onto a single layer of Confor™ foam would have a frequency of about 200 Hz. The drops onto the rigid concave surface produced shorter duration events, which would be within the range of frequencies noted in the previous paragraph, where the measured acceleration exceeded the input.



**Figure 8-37: Frequency response for y-axis of accelerometer when mounted in silicone rubber**



**Figure 8-38: Frequency response for z-axis of accelerometer when mounted in silicone rubber**

In terms of the intended use of the silicone as a damper, or mechanical filter, in event data recorders, it can be seen that there is no damping of the signal below about 900 Hz. Even then it is not clear that the silicone has offered much (if any) damping compared with the rigidly mounted accelerometer units.

Regarding comparisons with the small layer of skin in the ear canal; it is not expected that the skin would provide such resonance as the silicone. Instead, some isolation from oscillatory modes might be expected with skin. Therefore a different material, other than a silicone rubber would have to be used to mimic the lining of the ear canal.

### **8.5.3 Summary of results from shaker table testing**

Three configurations of mounting the miniature accelerometers to be used in ear-pieces were tried during shaker table frequency sweeps in the range of 10 to 2,000 Hz.

1. With the accelerometers rigidly mounted to the shaker table, no substantial difference was observed between the three measurement axes. All showed resonance around the harmonics of the sampling frequency. The miniature accelerometers began to under-read the input acceleration from about 200 to 300 Hz and above.
2. A softer mounting with 1 mm of low hardness silicone rubber surrounding the accelerometer was tried. This arrangement showed large over-reading with regard to the reference acceleration in the range from 200 to about 600 Hz. It is recommended that the option for mounting the accelerometers in such a soft rubbery compound is avoided when trying to obtain accurate acceleration measurements. It is assumed that this set-up is not a good approximation of the soft skin covering the hard bony structures within the ear canal.
3. The accelerometers were also evaluated when mounted in ear-pieces as will be used in the real-world application for generating head accelerations from motorsports participants. It was evident that the rubbery material used in the ear-piece and the shape of the ear-piece allowed various resonant frequencies to be excited. The exact frequencies causing over-estimation with regard the reference signal depended on the orientation of the ear-piece and also the ear. This indicates that small changes from ear-piece to ear-piece could cause differences in measurement above about 135 Hz.

In general, the miniature accelerometers seem to provide a robust and accurate measurement of the reference signal in the frequency range from 10 to at least 100 Hz. Above this frequency there was an intrinsic tendency for the sampling to cause under-reading with respect to the input signal. However, this under-reading can be complicated by resonance due to harmonics with the sampling frequency, excitation modes within the ear-piece and resonance with rubbery mounting materials. For this reason, higher frequency signals (above 100 Hz) should be considered with extreme care, so as not to underestimate the inaccuracies associated with the accelerometer hardware and its mounting in the ear canal.



## 9 Development of a low-cost alternative system

The results generated with the FIA Institute system were generally encouraging. However, it was felt, by the author that the scientific community could benefit from an equivalent system which was more widely available for data generation. For this purpose a low-cost ear-piece solution was developed.

### 9.1 Production of a system

The objective of this component of the research was to develop an initial prototype of a low-cost miniature accelerometer system. This would demonstrate how the necessary components for such a system can be purchased and integrated into an operational system without huge investment.

The components of the prototype system were:

- Olimex Pinguino PIC development board
  - Programming of the board to read data from the sensors was carried out by a colleague at TRL based on the information I provided, and as taken from the data sheets for the sensors.
- Micro SD card
- USB to mini USB cable
- Rechargeable Lithium ion battery
- Assembled accelerometer printed circuit board
  - This featured the STMicroelectronics H3LIS331DL MEMS motion sensor: a low power high  $g$  3-axis digital accelerometer ([http://www.st.com/web/en/catalog/sense\\_power/FM89/FM89/SC444/P\\_F253712](http://www.st.com/web/en/catalog/sense_power/FM89/FM89/SC444/P_F253712)) with a maximum sampling rate of 1 kHz.
  - The accelerometer transducer had the characteristics shown in Table 9-1. The device was mounted on a small printed circuit board with an overall area of 6 x 9 mm.

- The layout of, printing and mounting of components on the circuit board was provided by colleagues at TRL and a third party supplier.

**Table 9-1: Tri-axial accelerometer device characteristics**

Parameter	Minimum	Typical	Maximum	Unit
Size		3 x 3 x 1		mm
Operating temperature range	-40		+85	°C
Voltage supply	2.16	2.5	3.6	V
Current consumption		300		µA
Output data rate		1,000		Hz
Low pass single pole filter		780		Hz

### 9.1.1 Demonstration of function on table-top

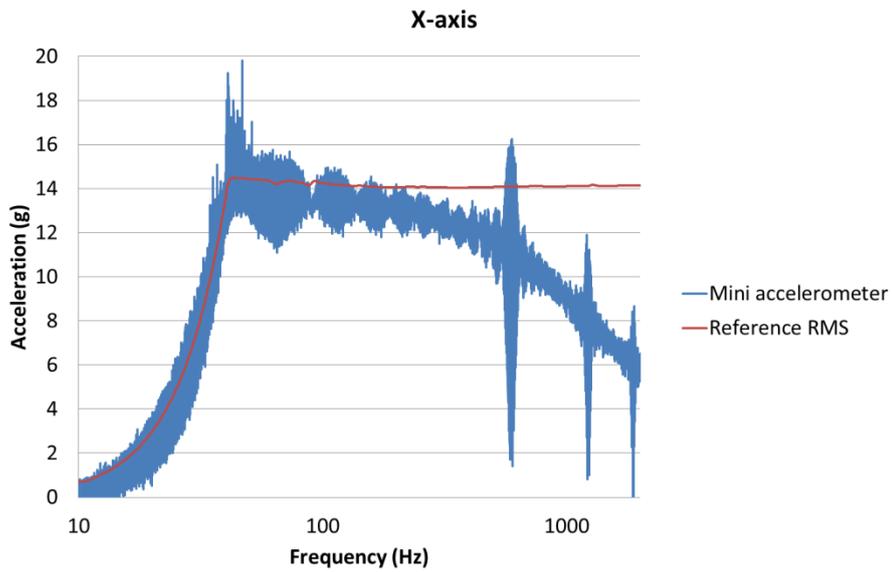
The ability of the sensor to detect accelerations was shown using a laptop and USB to serial connector. Either a live display could be obtained attempting to show the acceleration measurements from the three axes being sampled at 1 kHz. Alternatively, a data file could be saved and viewed later. During this process, it was determined that individual measurements running at 1 kHz were particularly difficult to discern via a live display, whilst watching values scroll across a computer screen. Hence, a step down sampling function was built into the software to reduce the speed to 200 Hz, which is slightly more reasonable to see in a live display.

When viewing the data under small accelerations due to shaking the sensors by hand an important observation was made. The signal to noise ratio at these levels is very low. The oscillation of the measurement, through noise, was  $\pm 2 g$ . Depending on the required sensitivity in the final application, this level of uncertainty could be an issue. As an example, having an inaccuracy of  $\pm 2 g$  on a peak acceleration of 20 g is ten percent and may represent a concern if the 18 to 22 g value was being used in a

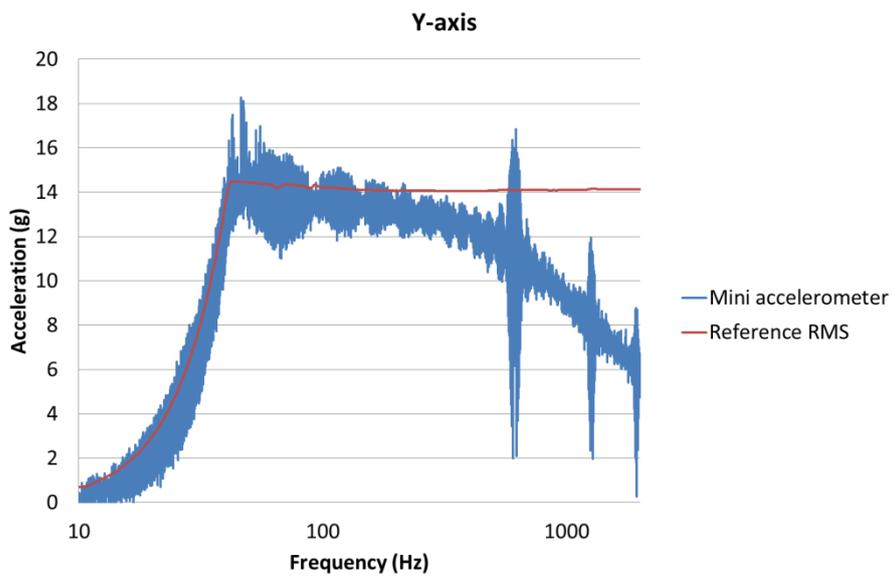
single assessment of injury risk. Although in severe impact events it may be a minor contributory factor in the accuracy of the system; with regards to injury metrics such as peak acceleration and HIC, for instance.

### **9.1.2      *Demonstration of function during shaker tests***

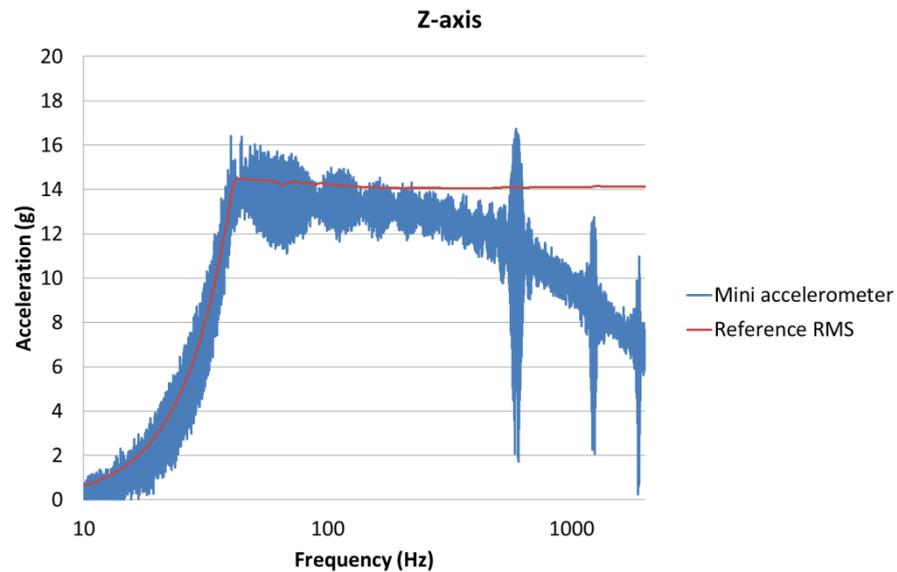
The evaluation of the accelerometer system followed a set routine for evaluating and calibrating accelerometers. A sinusoidal acceleration was applied with a reference measurement being taken. In these tests, the applied signal builds from 0 to 20 *g* (the root-mean-square of which is about 14). At the same time the frequency of the oscillation increases from 10 to 2,000 Hz. The whole cycle takes about 30 seconds. The prototype sensor performed evenly in all three axes and produced results as shown in the following three figures (Figures 9-1 to 9-3). It should be noted that there was no synchronisation of the start time possible in the setup. Therefore the two response curves were shifted 'by-eye' retrospectively so that consistently high acceleration measurements coincided, occurring first at around 40 Hz. As such it is not certain that the sensor system under-reads in comparison with the reference during the ramping-up phase. An alternative alignment could show a system response either closer or further from the reference.



**Figure 9-1: Shaker table test result for the x-axis; mini accelerometer output plotted with the root-mean-square (RMS) of the reference acceleration**



**Figure 9-2: Shaker table test result for the y-axis**



**Figure 9-3: Shaker table test result for the z-axis**

As mentioned in the previous section, the mini-accelerometer signal is noisy. This could be smoothed as 'post-processing', but to preserve the frequency response information generated in this sweep, no filter was applied. The frequency response does show some large features. The accelerometer output appears to be accurate up to about 300 Hz, at which point the mini accelerometer drops in relation to the reference. This can be expected since the mini system samples at only 1,000 Hz. Therefore it is no longer guaranteed to be able to sense the peak oscillatory acceleration in each sinusoidal wave. There are also harmonic effects at around approximately, 500, 1000 and 2000 Hz where the sampling point could coincide with the peak or middle of the applied sinusoidal function. These are to be expected based on the Nyquist frequency and folding or aliasing artefacts around multiples of half of the sampling frequency.

### 9.1.3 Demonstration of function during drop tests

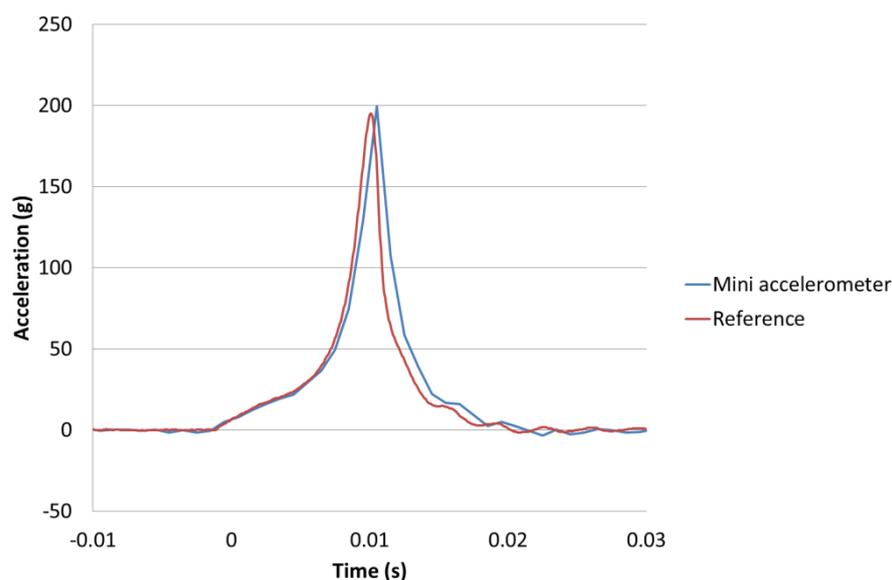
This evaluation was carried in the Helmet Drop Test Facility at TRL. The setup involved sticking (using double-sided tape) the mini accelerometer and a normal accelerometer to a linearly guided rigid impactor and dropping that onto some foam. Figure 9-4 shows the impactor in the top of the image with the double layer (50 mm) of blue CONFOR® foam (CF45) covered in a thin sheet of rubber at the bottom of the image. The rubber was used to give a little bit of additional stiffness compared with a purely CONFOR-based surface.



**Figure 9-4: Setup for drop tests with mini-accelerometer used in parallel with reference accelerometer.**

Twelve tests were completed. The only problem encountered in the use of the prototype system was that the power supply for the development board failed. This was repaired easily and testing could resume. Once the system makes use of the battery instead of a wired mains power supply, then this will no longer be an issue; the faulty connection will be redundant.

The severity of the pulse applied to the accelerometers was varied by changing the drop height from 1.5 m to 1 m (changing the impact speed from a nominal 5.4 to 4.4 m/s) and by changing the foam that the impactor was dropped onto. Example results are shown in Figure 9-5. From this example it is evident that the mini-accelerometer response followed that of the reference accelerometer very closely. There seems to be a slight deviation in timing, which may occur if the nominal 1 kHz, is slightly inaccurate. Otherwise the shape of the curve and the peak values agree well. This is a short duration event, with the whole response being over in around 20 ms. The fact that the mini-accelerometer can reproduce the shape of the curve with only 20 points demonstrates a useful level of precision in capturing the severity of the event.



**Figure 9-5: Back-to-back outputs from the mini and reference accelerometer when dropped 1.5 m onto 50 mm of CONFOR® CF-45 foam**

The table of tests undertaken in this prove-out phase is shown in Table 9-2. A range of impact conditions were selected to show the accelerometer performance in a variety of scenarios. These varied from a large depth of soft energy absorbing foam through to a smaller depth where the headform penetrated to a point where the foam was substantially compressed and the impact became much harder. This condition was

intended to represent a 'worst case' condition, where an energy absorbing component has been fully compressed or compromised in some way. This gives a short duration, high severity loading which was anticipated to be difficult to detect accurately with a low sampling rate sensor system. Finally, some lower speed impacts were also added to the matrix. Head Injury Criterion (HIC) values were in the range of 75 to 1150. It was intended that these variations span a range of conditions relevant to head loading scenarios in sporting events which could be associated with minor to moderate head injuries occurring.

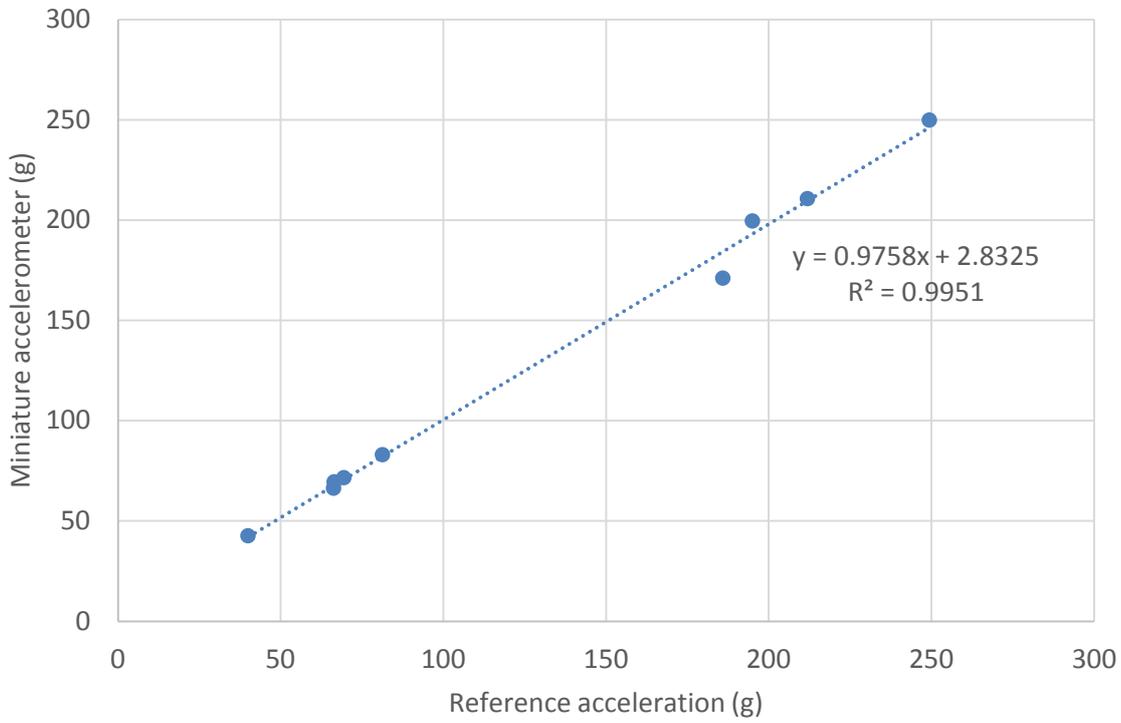
The data from Table 9-2 are shown graphically in Figure 9-6 where values exist for both the reference and miniature accelerometer systems. A best fit line is shown together with the 'R<sup>2</sup>' coefficient value of 0.995. This linear regression supports the assertion that there is very close agreement over the peak resultant acceleration value between the two accelerometer systems.

The data are also shown in Figure 9-7 with the Bland-Altman plot. The 95<sup>th</sup> percentile confidence limits of agreement for the miniature accelerometer, based on these data are about  $\pm 11 g$ . This is on the threshold of accuracy which can be considered useful based on the rationale introduced in Section 6.1.2, but is still within  $\pm 12 g$ .

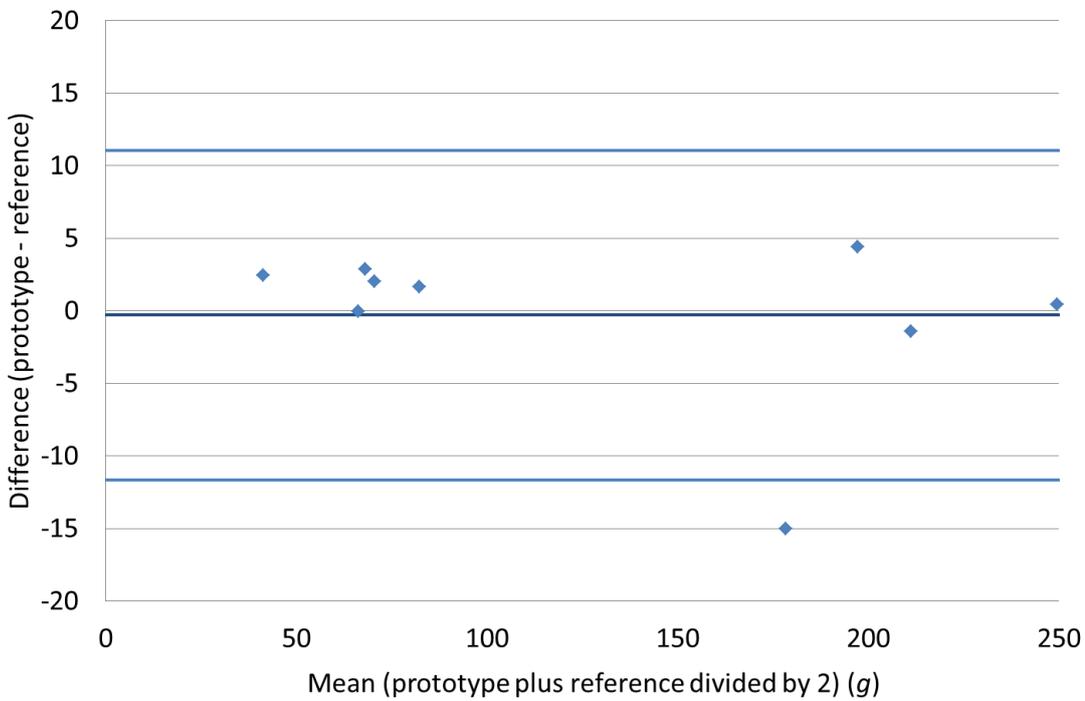
**Table 9-2: Test matrix and results from drop test comparisons**

<b>Test number</b>	<b>Foam</b>	<b>Recorded impact velocity (m/s)</b>	<b>Peak acceleration from reference (g)</b>	<b>Peak acceleration from prototype (g)</b>	<b>Difference (%)</b>
<b>1</b>	100mm CONFOR CF-45	5.42	40.0	42.5	6.1
<b>2</b>	75mm CONFOR CF-45	5.38	62.1	†	
<b>3</b>	50mm CONFOR CF-45	5.38	195.0	199.4	2.3
<b>4</b>	50mm CONFOR CF-45	5.39	185.9	170.9	-8.1
<b>5</b>	50mm CONFOR CF-45	5.40	227.0	†	
<b>6</b>	50mm CONFOR CF-45	5.39	310.7	†	
<b>7</b>	50mm CONFOR CF-45	5.37	212.0	210.5	-0.7
<b>8</b>	50mm CONFOR CF-45	5.44	249.4	249.8	0.2
<b>9</b>	75mm CONFOR CF-45	5.42	81.4	83.0	2.0
<b>10</b>	50mm STYROFOAM	4.36	66.3	66.3	-0.1
<b>11</b>	50mm STYROFOAM	4.36	69.5	71.5	2.9
<b>12</b>	50mm STYROFOAM	4.41	66.6	69.4	4.3

† In these tests, no mini-accelerometer data was obtained due to a failure in the power supply.

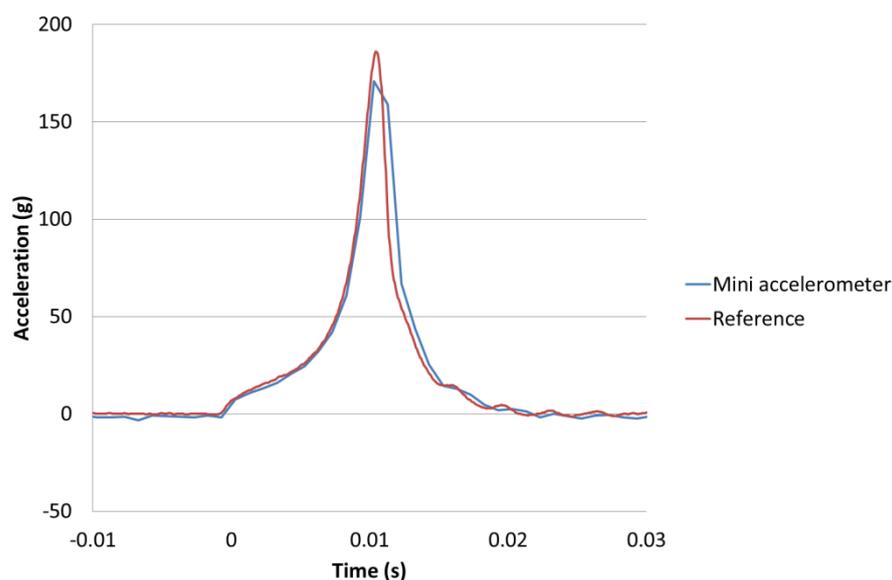


**Figure 9-6: Peak resultant acceleration values**



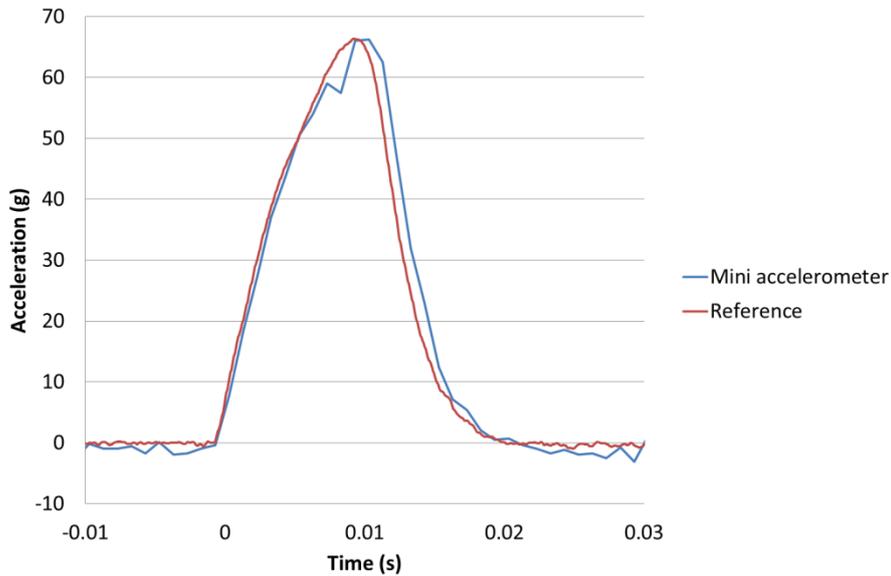
**Figure 9-7: Bland-Altman plot of peak resultant acceleration values**

The greatest difference between the mini-accelerometer and the reference, of 8.1 % of the peak value, occurred in a test onto 50 mm of CONFOR® foam. The responses from the accelerometers for this test are shown in Figure 9-8. From this figure, it can be seen what happens when the nearest samples from the mini-accelerometer bridge the peak of the event – there is a substantial underestimate of the peak value.



**Figure 9-8: Back-to-back outputs from the mini and reference accelerometer when dropped 1.5 m onto 50 mm of CONFOR® CF-45**

The closest match between the mini-accelerometer and the reference, of 0.1 % of the peak value, occurred in one of the drop tests onto the STYROFOAM (as used in some helmet linings). The responses from that test are shown in Figure 9-9. Despite a small error in the measurement of the mini-accelerometer just prior to the peak value, the comparison of the two pulses is very similar.



**Figure 9-9: Back-to-back outputs from the mini and reference accelerometer when dropped 1 m onto STYROFOAM**

Tests 4, 7 and 8 were set-up with a consistent impact surface dropping onto 50 mm of CONFOR® foam covered with a slim sheet of vinyl rubber. Based on the peak acceleration values, the coefficients of variation for these three tests are shown in Table 9-3. Due to the underestimated peak value from the miniature accelerometer unit in the first of these tests, the coefficient of variation is larger for the miniature accelerometer peak resultant values than for the reference sensors. However, without that result, the variation is very consistent between the two accelerometer systems.

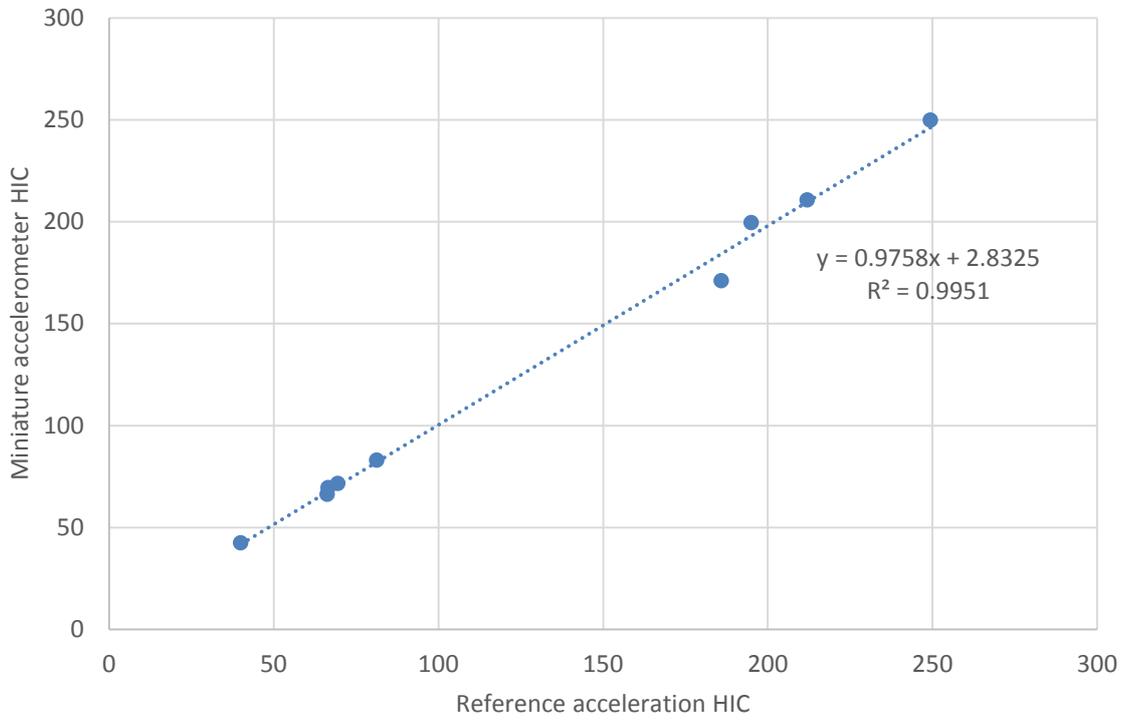
It can be noted that the peak resultant acceleration values from this test condition were more sensitive to test-to-test variations than the head drop tests considered in Section 8.2. With a coefficient of variation greater than 14 %, neither accelerometer system would meet the acceptance criteria for variation described in Section 6.1.2. This illustrates the complexity of repeating accurately the conditions of these drop tests from one instance to the next.

**Table 9-3: Coefficients of variation from peak resultant acceleration values in repeated tests**

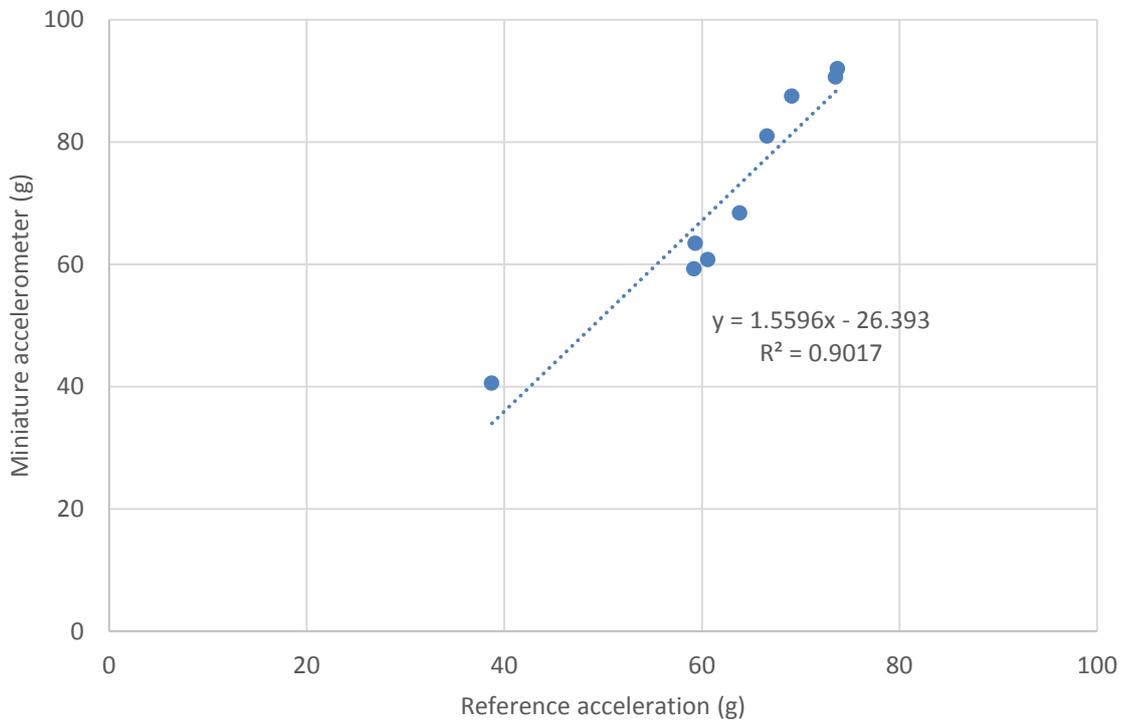
Coefficient of variation	
Reference	Miniature accelerometer
0.148	0.188

Head Injury Criterion (HIC<sub>15</sub>) and 3 ms exceedance resultant acceleration values from this series of tests are shown in figures 9-10 and 9-11. The values from the reference accelerometer comprise the x-axis and the values from the prototype system the y-axis. The HIC values show a similar 'R<sup>2</sup>' coefficient to that from the peak resultant acceleration values, as shown in Figure 9-6. However, the 3 ms exceedance values show a lower coefficient value. Also there is a substantial intercept with the line of best fit through the points, whereas reason suggests there should be no prototype acceleration with no reference acceleration. The implication of this result is that the duration of loading is having an adverse effect on the accuracy of the prototype measurements when considered over 3 ms. In particular, the prototype sensor system, overestimates the 3 ms exceedance value for those tests where the reference accelerometer value was over 60 g.

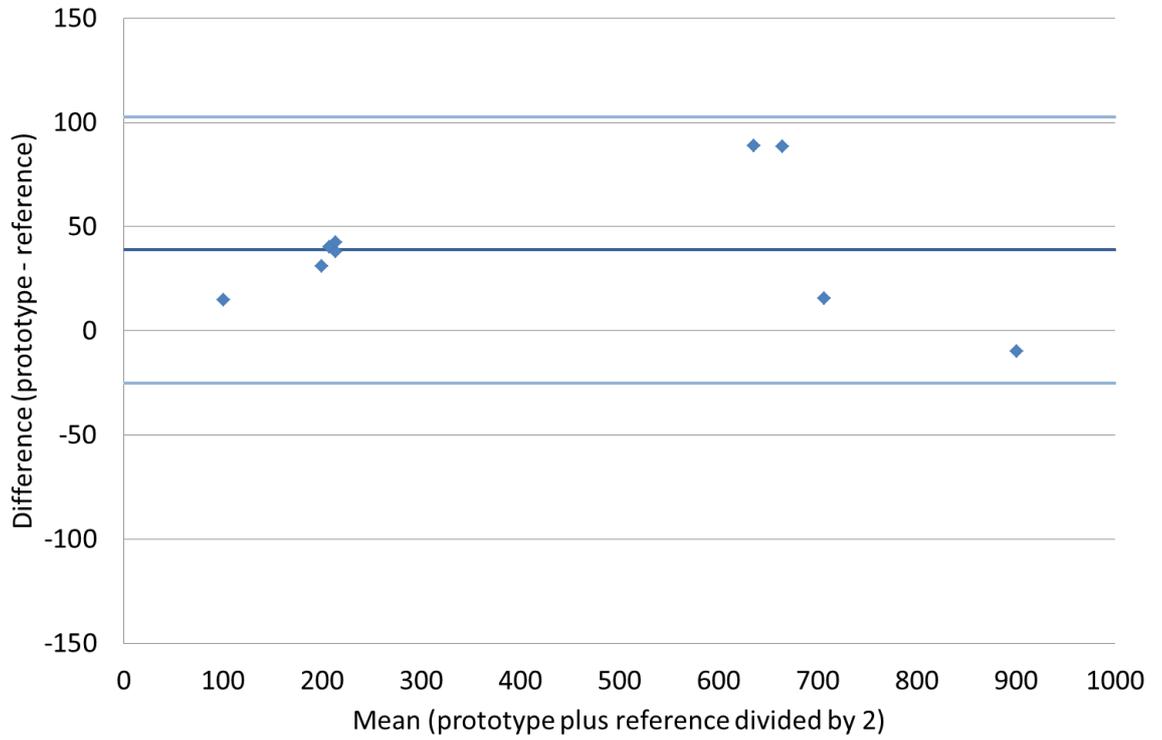
The Bland-Altman plots of these data (figures 9-12 and 9-13) also show an offset from zero mean difference between the two measurement systems; with the miniature accelerometer having a higher than zero mean offset in both HIC<sub>15</sub> and 3 ms exceedance. Such an offset would need to be removed or accounted for within the predictive function when using these measurements to estimate a risk of injury. The 95<sup>th</sup> percentile limits of agreement for HIC<sub>15</sub> are + 103 / - 25 and for 3 ms exceedance are + 25 / - 7 g. This remains a useful accuracy for HIC risk of injury predictions but is marginal for 3 ms exceedance according to the framework introduced in Section 6.1.2.



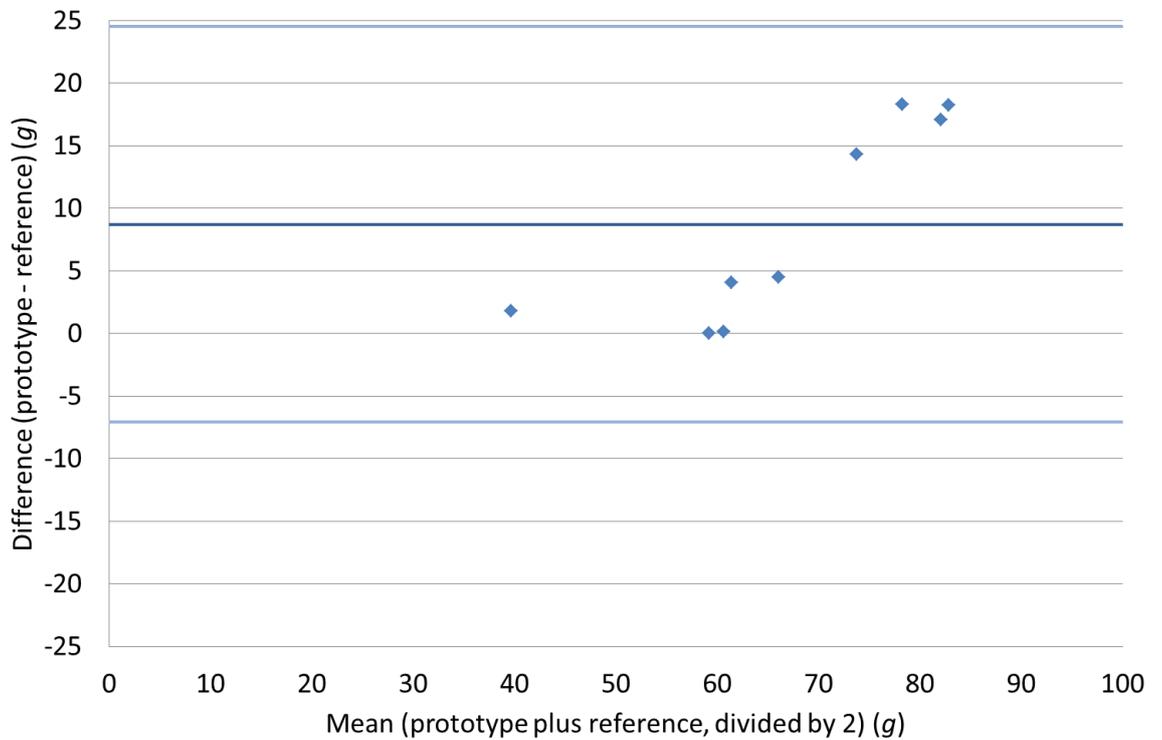
**Figure 9-10: HIC<sub>15</sub> values**



**Figure 9-11: 3 ms exceedance values**



**Figure 9-12: Bland-Altman plot of HIC<sub>15</sub> values from drop tests with miniature and reference accelerometer systems**



**Figure 9-13: Bland-Altman plot of 3ms exceedance values from drop tests with miniature and reference accelerometer systems**

#### **9.1.4 Additional development**

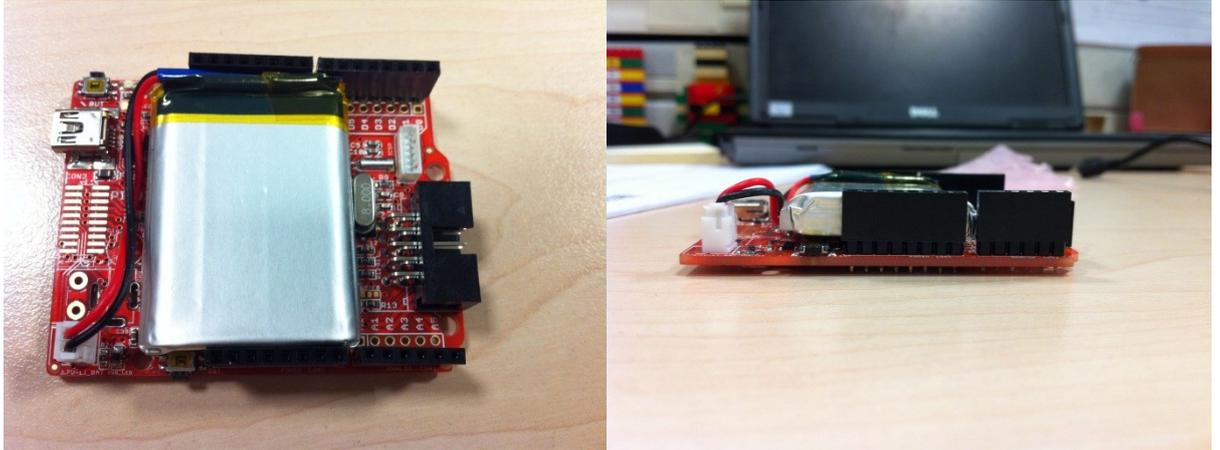
The test results generated above indicate that the prototype system presents a very good basis for further development. The frequency response is as could be expected, given the sampling frequency limitation of the principal accelerometer components. The matching of the acceleration time-histories and particularly the peak resultant acceleration and HIC values are closer than expected, given such a limitation.

After promoting the strengths of the prototype system, it remains to complete the additional steps required to take it from a prototype desk-based system to something that could be used by a sports participant in acquiring head loading information.

To be of general use, the system still needs the following developments, or demonstration of function:

- Data logging via the acquisition board
  - So far, the data acquisition has been via a live link with a laptop, where software (third party) allows collation of values and recording to a file, per event. However, to be a standalone unit, without a live, wired link to additional hardware, it must be possible for the system to record the same data to the micro SD card. This additional feature was programmed by a colleague at TRL.
- In a user friendly package, rather than exposed circuit boards, etc. there would be a closed unit which can be readily handled without fear of damage.
  - So far there has been no effort to mount the raw components into a box or package. For the system to be worn during a sporting event then consideration must be given to the overall packaging and how comfortable it is to be worn and how safe. Basic prototype boxes have been 3D printed, but will need refinement before use.

- By removing some unnecessary components from the development data logging board, it has reduced the height to be packaged. It has also given space to place the battery on top of the board without protruding above other components. This is shown in Figure 9-14.



**Figure 9-14: Data logger, with unnecessary components removed and battery placed on top**

- A version of the 3-D printed prototype boxes used to house the data logger, as used in the subsequent trials, is shown in Figure 9-15.



**Figure 9-15: Box housing the data logger, or acquisition unit**

- Potentially the acceleration sensitive components may need to be incorporated into a helmet or head guard
  - As mentioned, the evaluation prototype has not packaged the printed circuit boards at all. This is potentially useful for the accelerometer unit where the flat underside of the board can be pressed against the object of interest yielding a precise orientation of the sensitive axes, with regard to the object. However, as mentioned in the design specification section (Section 7.1) it is unlikely that it will be acceptable ultimately to implement the adhesion of the accelerometers to the skin of sports participants directly. Instead, they may have to be mounted in an item of sporting wear (for instance some form of head gear). Completion of this step would allow characterisation of the damping brought about by the head gear, in comparative tests between the exposed accelerometer tests and the imbedded accelerometers.
  
- Processing of the recorded data in a consistent manner
  - As a research tool, the accelerometers are ready to be used. However, the data from each test are handled by a researcher in an ad-hoc manner. It would be sensible to build into the logger basic processing functions. As an example, an anti-aliasing filter would be useful in avoiding sampling artefacts and additionally (and perhaps optionally), a filter could be applied to reduce noise extraneous to the impact event for quick post-impact interpretation. Furthermore, depending on how the accelerometer units are mounted, some calculations could be useful to transform the data outputs to a different coordinate system. These functions were intentionally left out of the prototype until the need for them was demonstrated, but the potential usefulness of post-processing in the logger remains.

Once data are transmitted from the logger to a device by which a user can look at the outputs, then there could be a need to develop a user interface. At the moment the

researcher is able to process the data to show peak acceleration etc. However, this isn't feasible for a non-technical user. Instead something that calculates and displays simple metrics such as peak acceleration and direction could be useful. In addition some interface whereby a user could input injury information (e.g. concussive symptoms) could be valuable for future analysis.

#### **9.1.5 Conclusions**

A prototype micro accelerometer system was developed which could potentially be used for the assessment of head impact loading severity during sporting events.

On the desk, noise was detected on the signal from the micro-accelerometers. This could provide a source of inaccuracy during relatively low severity impact events.

In shaker-table, calibration-type evaluations, the frequency response of the micro accelerometer was reasonable given that the maximum sampling frequency is only 1 kHz.

In drop tests with an impact duration of about 20 ms and a peak acceleration value of 60 to 250 g, then the response of the micro-accelerometer was excellent. It seemed able to capture the shape of the acceleration event and always had a peak value within 10 % of the reference accelerometer. The conditions used in the drop tests are expected to be relevant to head loading scenarios in sporting events which could be associated with minor to moderate head injuries.

In a market place with a constantly increasing number of alternative devices, it remains to be shown that this device fills a necessary gap between larger helmet mounted accelerometers and the bespoke in-ear devices used in Formula 1. Increased fidelity over the helmet mounted systems would be a necessary next step for demonstrating the value of such technology in this application and this is likely to vary depending on the precise method used to attach the accelerometer to the head of a sportsperson. Furthermore, use of the system or a suitable alternative is a prerequisite for benefit to be realised in the safety community.

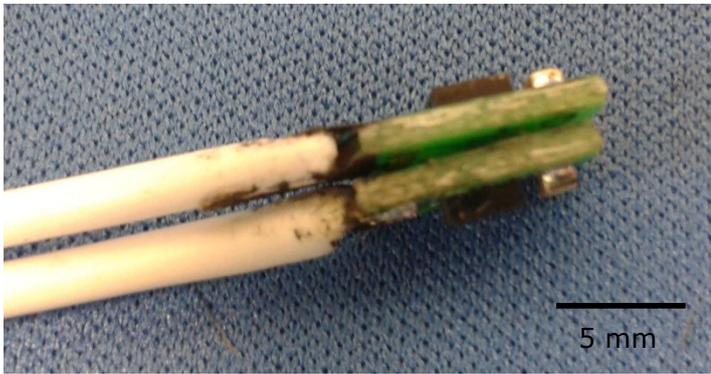
## 9.2 Integration into ear-pieces

To mount the accelerometer units in a person's ears, keeping them in position and in a unit that is comfortable, it is necessary to embed them in an ear-piece. The process used with the motorsports system incorporates the following aspects.

1. A (positive) mould is taken of the user's ears
2. A (negative) cast is taken of the mould
3. (positive) Ear-pieces are moulded from the cast and incorporating the accelerometers

As this is a professional service, Step 1 is carried out by an audiologist at a venue remote from the place responsible for incorporating the instrumentation into the ear-piece. Steps 2 and 3 are carried out by a service provider which usually sells ear-piece monitors for use by musicians. For example, ACS Custom (<http://acscustom.com/uk/>) provides ear-pieces for the X-Factor television music competition using the same process as for provision of their motorsport 'ACS Driver Communicators'. These communicators are an example of the technology on which the motorsports in-ear accelerometer is based.

Initially it was considered that this procedure could be by-passed if the accelerometers could be pressed into extremely cheap disposable earplugs (the kind available for 10 pence per pair, or less). As a demonstration of the approach, an accelerometer sensor board and an angular rate sensor board were placed back to back (as in Figure 9-17). Details of the angular rate sensor (digital gyroscope) unit are provided in Table 9-4. A small slit was made in the end of a disposable earplug using a scalpel and then the sensors were pushed into the earplug and the earplug pushed into the modified 'ear' of the Hybrid III dummy head.



**Figure 9-16: Image showing a miniature printed circuit board containing an accelerometer chip pressed back-to-back with a circuit board containing an angular rate sensitive chip**



**Figure 9-17: Cheap earplug containing sensor circuit boards fitted into the modified Hybrid III dummy 'ear'**

**Table 9-4: Tri-axial angular rate sensitive device characteristics**

Parameter	Minimum	Typical	Maximum	Unit
Size		3 x 3 x 1		mm
Operating temperature range	-40		+85	°C
Voltage supply	2.2	3.0	3.6	V
Current consumption		5.0		mA
Output data rate		757.6		Hz

This approach didn't provide a viable way of mounting the sensitive elements in an ear-piece in a way that could be worn comfortably during exercise. With the circuit boards in the earplug it was no longer able to conform to the shape of the ear. This meant it was poorly retained in the ear and when tried by the author, it was also mildly uncomfortable even for short periods. In response to this it was decided that a fitting procedure more like that used with the motorsports system would be necessary to make the low-cost alternative a reasonable proposition for sports participants.

To avoid the complication of requiring sports participants to have to visit an audiologist and the cost of having bespoke ear-pieces moulded commercially, other options for placing the sensors in the ear-pieces were investigated. As a simplification of the whole procedure it was attempted to fit the miniature accelerometers into the ear-piece during the time of taking the first mould of the ear. The initial trial of this was carried out on the modified Hybrid III crash test dummy head with ears.

The moulding process was based around the use of Proguard UK 'Customised Personal Earplugs' (<https://www.proguarduk.co.uk/>). These are a commercial 'mould your own earplugs' solution designed to give a custom fit and provide hearing protection for a range of applications, such as: at music events, motorcycling, sleeping, swimming, in industrial environments, shooting, travelling, etc. The moulding process is relatively straightforward and well described in a few simple steps documented on a sheet provided with each earplug box. An abbreviated summary of the steps is provided below for information. The moulding material is silicon based and provided in two little pots, with a pair of pots for each ear. The process begins with removal of the material from these pots:

- Remove the silicone from the pair of containers (see Figures 9-18 and 9-19)



**Figure 9-18: Earplug material as provided in two containers**



**Figure 9-19: Earplug material ready to be removed from containers and blended together**

- Blend the two halves of material together
- Knead vigorously (the advice here is to knead for 30 seconds; however, it was found that a shorter period of 15 seconds was more appropriate, leaving the silicone in a soft pliable state for easier pressing into the ear)

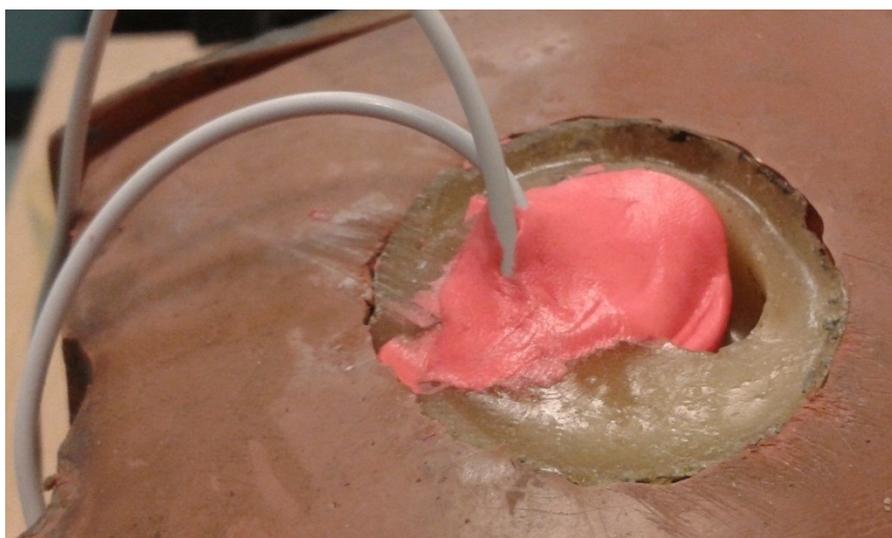
- Place the ball of material into the ear, push into the middle of the ball allowing the silicone to spread and take the shape of the ear (as in Figure 9-20)



**Figure 9-20: Earplug material pressed into the ear and setting**

- Wait for the silicone to set, before removing the earplugs
  - After 10 minutes the ear plugs can be removed, but they need 30 to 45 minutes to harden completely

The result of pressing back-to-back sensor units into the silicone whilst it was still soft and curing in the ear of the dummy can be seen in Figure 9-21.



**Figure 9-21: Image of the silicone earplug during the fitting process into the right ear of the modified Hybrid III dummy including sensor units**

One issue with this concept was that the precise location of the sensor units wasn't known. The silicone material for the earplugs is coloured and opaque and therefore it is difficult to judge how far the sensors should be pushed into the ear at this time. As a result it seemed possible to push the sensors too far. If attempted with living subjects, this could cause discomfort. Also, from a technical perspective it was difficult to guarantee that the silicone material completely encapsulated the sensors. The result of the trial with the dummy 'ear' demonstrates this, as shown in Figure 9-22.

Following this trial fitting it was decided that the removal of the mould casting and remoulding steps from the ear-piece fabrication procedure was not reasonable for trial participants. The convenience, comfort and safety for the participant was at risk and would be easier to ensure with these steps included. Therefore further experimentation was carried out to refine the process.



**Figure 9-22: Moulded silicone ear-piece for the modified Hybrid III dummy ear containing a miniature accelerometer and angular rate sensor, which can be seen exposed through the earplug material**

Advice was received on what material to use in creating a cast of the ear-pieces from different sources. A plaster cast was dismissed. It was anticipated that the complex structure of the cast and the solid nature of the plaster would make it difficult when trying to remove moulded ear-pieces carefully. Instead a silicone rubber cast was created.

The product used for this purpose is called OOMOO® ([http://www.smooth-on.com/Silicone-Rubber-an/c2\\_1113\\_1136/index.html](http://www.smooth-on.com/Silicone-Rubber-an/c2_1113_1136/index.html)). Unfortunately, the first cast showed that the earplugs float in OOMOO and therefore it was clear that for subsequent attempts the earplugs needed to be pinned in position at the bottom of the container used in the casting process. Once this was realised, it quickly became apparent that the flexibility of the OOMOO made it a good choice for this application. It was hard enough that a detailed and accurate mould could be created whilst also being flexible enough that the finished earplugs could be quickly and easily removed. It was also discovered that release agent was important to aid in the demoulding and in helping to prevent the silicone rubber cast from keying to the silicone rubber earplugs.

Figure 9-23 shows the rubber cast of an earplug. Figure 9-24 shows the same cast once the ear-plug has been removed.



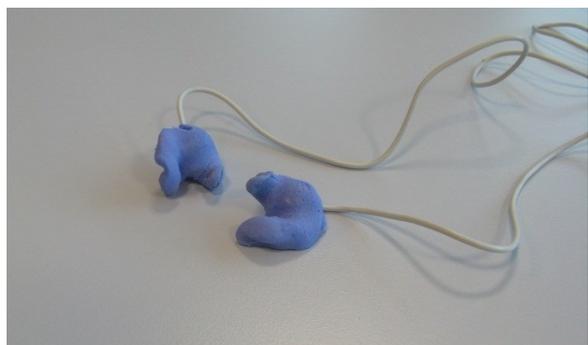
**Figure 9-23: A red Proguard UK mould-your-own 'Customised Personal Earplug' in blue OOMOO® cast**



**Figure 9-24: OOMOO® ear plug cast once the plug has been demoulded**

When the cast was ready for use in moulding new earpieces, then a fresh mould-your-own kit was used. A small piece of the material was taken out from each half (both of the two compounds). This was combined first and used purely to ensure that the sensors were completely encapsulated. After ten minutes, when hard to the touch, the remaining material was combined and pressed into the cast. The sensors were then pushed into the ear canal cavity within the cast until resistance could be felt, thereby ensuring the sensors were placed deep within the mould. This complete unit was then left to cure as instructed.

Examples of a set of ear-pieces and an individual ear-piece are shown in Figures 9-25 and 9-26, respectively. These were created using the procedure described above.



**Figure 9-25: A pair of instrumented ear-pieces ready to be used by a trial participant**

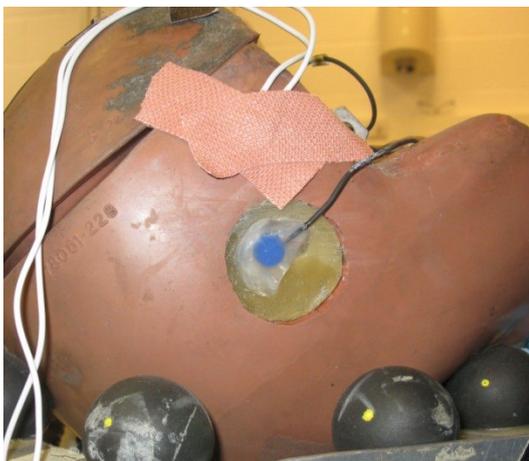


**Figure 9-26: The instrumented earplug component of the ear-pieces**

### 9.3 Validation of positioning

A small series of experiments was conducted to validate the fidelity of data that could be expected mounting sensors in the ears of participants. The experiments were formed of drop tests with a Hybrid III dummy head. This was the same head as was used in previous validation testing; the one with modified 'ears'. Three sensor systems were used for the purpose of checking the quality of data that could be generated in impacts of this kind:

- Laboratory grade accelerometers and angular rate sensors were positioned at the centre of gravity of the head. The data were sampled at 20 kHz. This data is considered to provide the baseline against which the other two systems can be evaluated
- The FIA Institute in-ear accelerometer system was used. One ear, (the left-hand-side) was populated with an FIA ear-piece (as shown in Figure 9-27), whilst the other ear-piece was stuck to the side of the head or to the chin of the dummy to keep it out of the way during the testing.



**Figure 9-27: FIA Institute in-ear accelerometer ear-piece fitted to the modified Hybrid III dummy head**

- The prototype system described in the previous section was used as well. In this case, an instrumented ear-piece for the right ear of the dummy was provided using the same moulding process as was described. This included both a tri-axial accelerometer circuit board and a tri-axial angular rate sensor (Figure 9-28). In addition equivalent circuit boards were used but not moulded into an earplug. Instead the free sensors were either pressed into the lining of a fabric headband (Figure 9-29) or taped to the side of the head using a fabric plaster (Figure 9-30). This was to give some data as to the accuracy of measurements that could be expected if other, crude but simpler, methods of attachment were used.



**Figure 9-28: Prototype sensor ear-piece fitted to the modified Hybrid III dummy head**



**Figure 9-29: Prototype sensor system fitted to the dummy head via a fabric headband**



**Figure 9-30: Prototype sensor system fitted to the dummy head via a fabric plaster**

For this testing, as well as the usual definition of the study parameters and analysis and reporting, I was also solely involved in the experimental work operating the drop test rig and acquiring all data from the three data acquisition and instrumentation systems.

In each test the head was dropped onto a flat surface padded with 50 mm of blue Confor™ (CF-45). The drop heights were varied within the range of 0.5 to 1.5 m.

Twelve tests were completed in this series. Prototype data were acquired in only eight of these twelve. Therefore the first conclusion from this testing is that the data acquisition unit is not fit for purpose.

For these impacts the unit was mounted onto the carriage being used to support the head during pre-drop preparations. The carriage was arrested at the bottom of the drop phase (after the head had hit the impact surface) by striking an arm with a rubber tip. The height of the arm relative to the height at which the head hit the impact surface was controlled with compressed air. The precise pressure supporting the arm and hence the deceleration created as the carriage was brought to a stop is unknown. Despite this uncertainty over the tolerance of the data acquisition unit, it is still clear that for further evaluation of the sensor system a more robust data acquisition system is required. It should be crash hardened to some extent. Experience with other circuits used in the telematics industry suggests that tolerance to 100 *g* can be achieved through circuit design without substitution of components. In this instance, the use of an SD card for recording the data is not ideal. It would be better to have this memory as a non-removable unit, embedded within the circuit itself.

In addition the cables connecting the ear-pieces with the data acquisition box kept pulling free of the box under their own inertia. For most cases where this occurred, the data acquisition unit tripped into a mode where no further data were sampled and the existing data from the experiment could not be retrieved after the event. This behaviour would be helped if it was only necessary to have one cable running to each ear, rather than one running to each sensor board; then the mass of the cables close

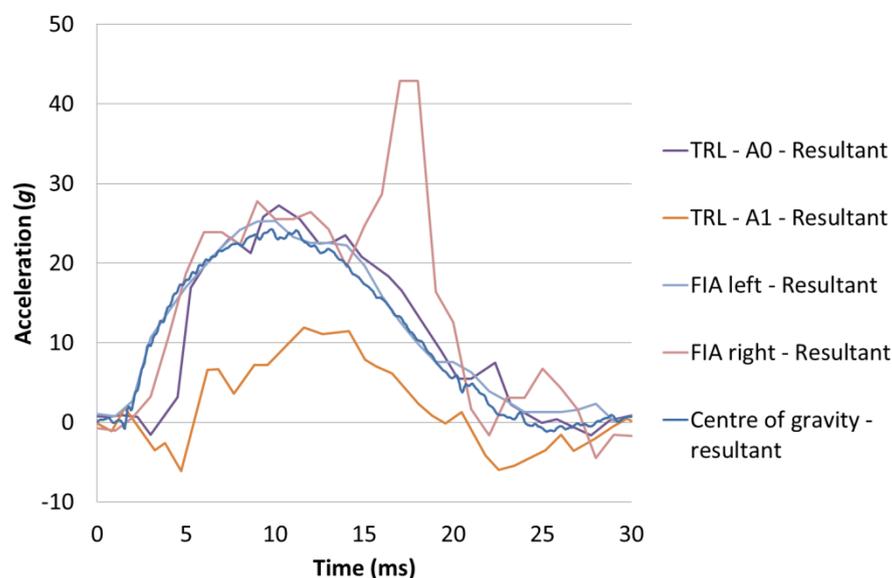
to the box could be halved. This is a design that would be possible and sensible knowing that the sensitive circuit boards will always be used together (e.g. one accelerometer and one angular rate sensor). In that case they can share power and communications as the system uses serial sampling of the sensitive chips.

For the eight tests where data were collected from all three sensor systems, it was possible to compare the outcomes. It should be noted that the angular rate sensor circuit board mounted in the ear of the dummy failed and the FIA Institute ear-pieces do not include an angular sensor. Therefore no comparisons can be made with the angular velocity recorded via the laboratory standard instrumentation at the centre of the head can be made. Also in a further two tests, when mounted in the headband, the prototype accelerometer 'A1' gave anomalous data, which was unreliable and could not be used for comparison with the other sensors.

With regard to the acceleration measurements, Figure 9-31 shows an example of a comparison chart following a drop test from 0.5 m onto a 50 mm thickness of Confor™ (CF-45, blue). To provide related measurements, only the resultant acceleration of the head is plotted. This negates the issues surrounding the resolution and alignment of each of the three axes from tri-axial sensors, each with a different and unknown orientation to the head and the centre of gravity instrumentation. The resultant accelerations are able to show negative values as the resultant was calculated before any offset was removed.

The accelerations from the centre of gravity of the head were filtered with a Channel Filtering Class of CFC\_1000 (SAE, 2003). The other channels were not processed. The prototype accelerometer 'A0' was stuck to the side of the head in this test whereas, the accelerometer 'A1' was in the moulded ear-piece. Also, the FIA Institute left ear-piece was in the ear of the dummy, whilst the 'right' sensor was loosely taped to the face.

It can be seen in Figure 9-31 that three of the sensors follow the general trend of the resultant acceleration at the centre of gravity: the two FIA Institute ear-pieces and the prototype sensor mounted to the face of the dummy head. The right FIA Institute ear-piece shows a large spike in acceleration following the peak from the other channels. This is uncharacteristic of the event; however, as the ear-piece was only loosely taped to the face of the dummy, it can be imagined that movement of the ear-piece with respect to the head could account for this. If this second peak is discounted, then this FIA Institute ear-piece still provides a closer representation of the head accelerations than the prototype sensor mounted in the ear of the dummy.



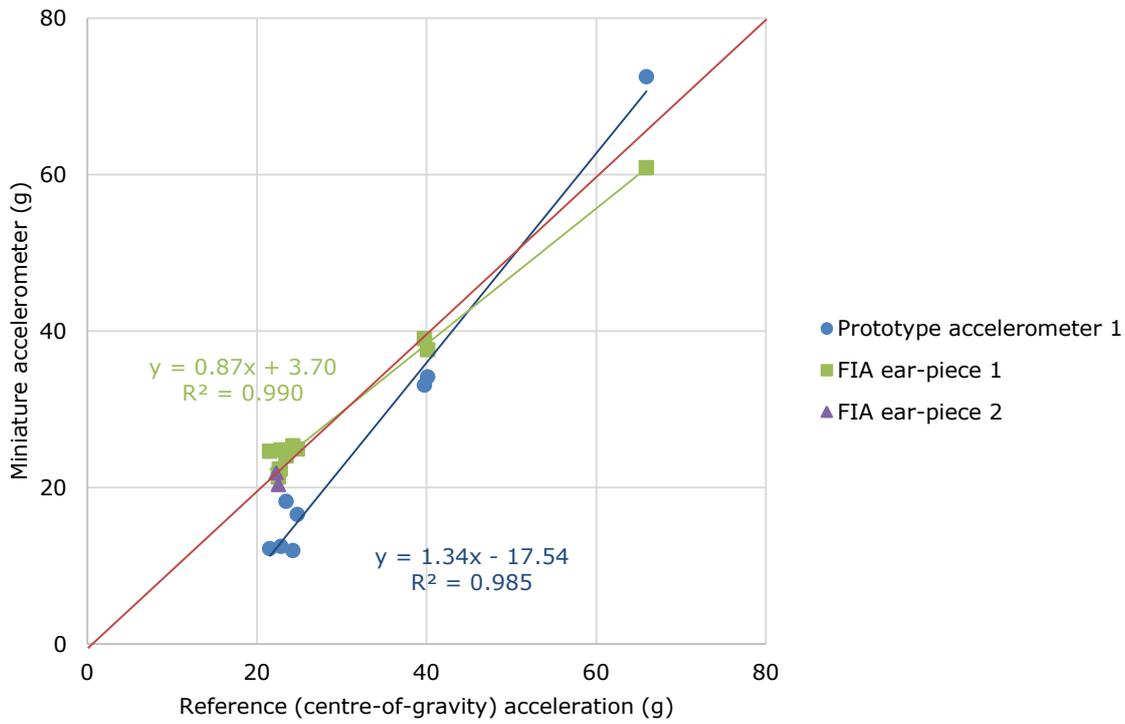
**Figure 9-31: Measured accelerations from a 0.5 m drop test onto 50 mm Confor™ foam**

Peak resultant acceleration results from the full test series are shown in Table 9-5. For each test, this table shows the location of each sensor and the peak acceleration value. For the first four tests, the in-ear accelerometers were only mounted in the cheap ear plugs and not the moulded OOMOO® ear-pieces.

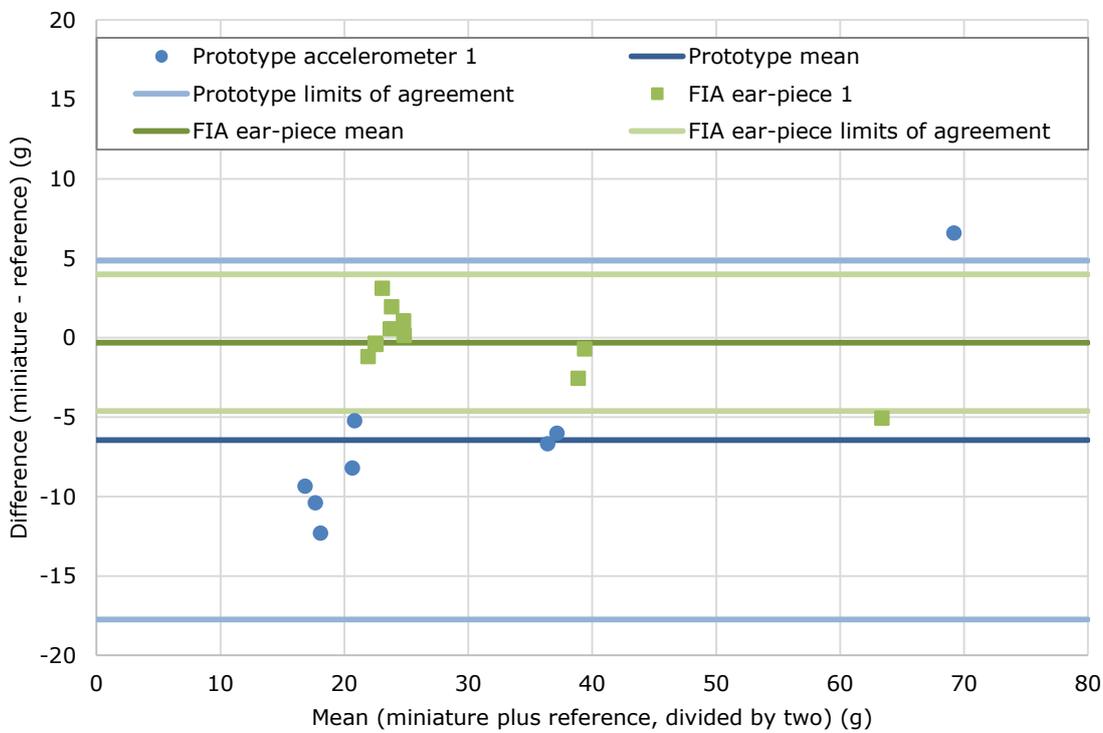
**Table 9-5: Peak resultant acceleration values**

Test number	Drop height (m)	Impact location on head	Centre of gravity (g)	Proto accel. 1 (g)	Proto accel. 1 position	Proto accel. 2 (g)	Proto accel. 2 position	FIA 1 (g)	FIA 1 position	FIA 2 (g)	FIA 2 position
G077D016	0.5	Forehead	22.3	38.3	Left ear	35.1	Neck, RHS			21.9	Right ear
G077D017	0.5	Forehead	22.8	35.4	Right ear	44.6	Neck, LHS	22.3	Left ear		
G077D018	0.5	Forehead	22.7	40.2	Right ear	45.1	Headband	22.3	Left ear		
G077D019	0.5	Forehead	22.5	37.7	Neck, LHS	52.0	Headband	21.3	Left ear	20.4	Right ear
G077D-A	0.5	Forehead	24.2	12.0	Right ear	27.3	Neck, LHS	25.3	Left ear		
G077D-A 004	0.5	Forehead	24.8	16.6	Right ear	40.0	Headband	24.9	Left ear		
G077D-A 005	1	Forehead	39.8	33.1	Right ear			39.1	Left ear		
G077D-A 006	1	Forehead	40.2	34.2	Right ear	37.8	Neck, LHS	37.6	Left ear		
G077D-A 007	0.5	Crown	23.5	18.2	Right ear	44.3	Neck, LHS	24.0	Left ear		
G077D-B 002	0.5	Side	21.5	12.2	Right ear			24.6	Left ear		
G077D-B 005	0.5	Side	22.9	12.5	Right ear	20.9	Headband	24.8	Left ear		
G077D-C 002	1.5	Forehead	65.9	72.5	Right ear	78.1	Neck, LHS	60.9	Left ear		

The prototype sensor in the ear was expected to be coupled tightly to the head of the dummy. However, it read substantially lower accelerations than the centre of gravity sensors. In the five tests at this drop height, to the forehead, right side and crown of the head, this ear-mounted sensor provided a peak value which was between 49 and 78 percent of the peak value in resultant acceleration from the centre of gravity sensors. In the two tests with a drop height of 1 m, the peak acceleration was higher and the ear-mounted sensor closer to the centre of gravity measurements (being 83 or 85 % of the centre of gravity peak value). Peak resultant acceleration values from ear-piece mounted accelerometers are shown in Figure 9-32. The Bland-Altman plot of these data is shown in Figure 9-33. These figures shows that the prototype sensor was less accurate in matching the centre of gravity reference than the FIA Institute in-ear accelerometers. The mean of the differences in peak values compared with the centre of gravity reference peak values is further from zero with the prototype sensor and the limits of agreement are wider. With 95<sup>th</sup> percentile confidence limits of agreement of  $\pm 11 g$  around a mean difference offset of  $-6 g$ , then the limits with the prototype are of marginal use according to the framework set out in Section 6.1.2. However, the negative limit at  $-18 g$  would be unacceptable without accounting for the offset. The prototype accuracy is within  $\pm 5 g$  so would be 'useful' according to the same framework.

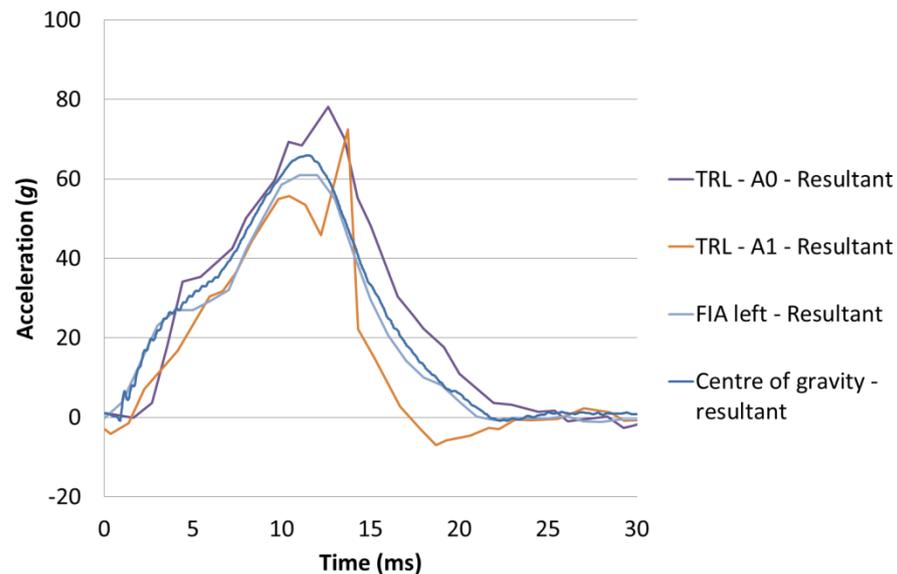


**Figure 9-32: Peak resultant acceleration values from ear-piece mounted accelerometers**



**Figure 9-33: Bland-Altman plot of peak resultant acceleration values from ear-piece mounted accelerometers**

In the one test dropped from 1.5 m, the ear-mounted prototype sensor acceleration was 110 % of the centre of gravity peak value. However, this was down to an out-lying single point after the main peak had occurred. Otherwise it would also have been 85 % of the centre of gravity peak resultant acceleration value. The acceleration results from this test are shown in Figure 9-34.



**Figure 9-34: Measured accelerations from a 1.5 m drop test onto 50 mm Confor™ foam**

Also in the 1.5 m drop test, the prototype 'A0' accelerometer was attached to the head with a fabric plaster. Figure 9-34 shows that the resultant acceleration from the A0 sensor is a close approximation of the centre of gravity resultant acceleration. However, it is of a slightly longer duration and has a slightly higher (119 %) peak value. It should be noted that whilst the FIA Institute and centre of gravity data is synchronised via the use of the same trigger, there was no possibility of providing a trigger input for the prototype sensor system. Therefore, the pulses were aligned 'by eye' in the plotting of the chart.

The attachment of the sensor system to the head via a headband was less appropriate for assessing the severity of loading to the head. The shape of the acceleration curves was less well modelled by the headband accelerometer. Also the peak resultant acceleration value was much more variable than from the other accelerometer attachment options used. It varied from being 91 to 230 % of the peak value as measured by the sensors at the centre of gravity. This seems to reflect the poor coupling of the sensors to the head and the ability for them to gently slide in relation to the skin and to be subjected, potentially, to shorter duration loading when finally constrained. This secondary deceleration event generates a higher acceleration than the initial contact between the head and foam covered loading platform as the hard sensor circuit board contacts the head or an adjacent sensor.

### **9.3.1 Discussion of positioning results**

The practicalities of this test series, the timing and ability of the prototype data acquisition system to collect data in higher severity events, mean that a limited evaluation of the sensor system and positioning options can be provided. Despite these limitations several general findings seem apparent.

- The data acquisition unit for the prototype system must be improved so that it is possible to conduct impact event trials at higher severities.
  - The unit itself should be made more resistant to shock events. This is important for a final product as participants should not be exposing themselves to the inconvenience of wearing such a system without assurance that the data will be collected reliably.
  - The solutions should not be complex; for instance, using non-removable memory and a stronger cable attachment and lighter cables.
- The FIA Institute system was shown to be effective in capturing the general traits of acceleration events at these severities.

- The prototype system is not as robust as that used by the FIA Institute.
  - As well as the data acquisition unit, the moulding process damaged one of the two sensors in the prototype ear-piece.
    - In this case, the sensors had already been cast in an earlier version of an ear-piece. Therefore it could have been that the removal process had weakened a connection; as much as the moulding process. However, the fact remains that by the time of the first data collection event, the sensor was no longer working and it is unclear as to what exactly had damaged the unit.
- The dynamic properties of the silicone used in the FIA Institute ear-pieces and the prototype system ear-pieces are different.
  - The FIA Institute ear-piece gave a closer relationship to the centre of gravity accelerations than the prototype version.
  - The prototype ear-piece attenuates the applied acceleration based on the small series of tests and severities investigated here.
    - It doesn't seem that the Proguard ear plug material is particularly softer than the FIA Institute ear-pieces, which may have contributed to some attenuation.
    - Perhaps the fit of the prototype ear-pieces in the ear of the dummy was looser than with the FIA Institute ear-pieces. This may well be the case given that the ears of the dummy were cast around the FIA Institute ear-pieces, but the prototype ear-pieces had been remoulded from a cast of an earplug used to capture the shape of the dummy's ears.
  - Further investigation of the dynamic properties of the earplug silicone material is needed to understand this behaviour and to help in selecting another alternative.

- It is possible to select hardness properties associated with commercial ear-pieces and it would be interesting to see to what extent the dynamic properties can be matched by low cost alternatives.
- Findings on the merits of fitting accelerometers to the head via a fabric plaster and the neck or cheek are mixed.
  - This is a very quick and convenient attachment method
  - The shape of the acceleration pulse is described, but small differences in duration or precise magnitude can be evident from this test series
  - With selection of a particular type of tape/adhesive a robust attachment can be provided, at least up to the severities generated in this small test series.
  - However the sensor unit is attached to the skin there remains decoupling between the skull and the skin during an impact event. This skin to skull motion will influence the measurements obtained from the sensor and the ability to predict acceleration at the centre of gravity of the head.
- Findings on the merits of fitting accelerometers in a headband are clear. This is not a reliable method for capturing the loading to the head. It allows too much separation of the sensor from the head and enables secondary contacts unrelated to the motion of the head itself.

#### **9.4 Results from field trials**

To determine if it was a reasonable proposition to collect acceleration and angular rate data from sports participants using the prototype low-cost in-ear system a couple of participants were recruited to try that.

As the trials included human participants, the TRL ethical procedure was followed. Participants were recruited via an advertisement at TRL on an internal network area. One of the restrictions placed upon the study during ethical review discussions was that the trial should avoid contact sports where a direct blow to an ear-piece could be expected. It also provided advice to avoid non-contact sports where hearing was required for safety. Therefore, the advertisement was written to recruit two people who were participating in a sport in which they would routinely expect to wear a full, rigid shell, helmet that covered the ears. As a result of this a motorcyclist and an American football player agreed to try wearing the system during an event and report on the convenience they experienced. A participant information sheet was provided and is reproduced in Appendix A, together with the participant advertisement and consent form.

To facilitate this, bespoke instrumented ear-pieces were made for the two trial participants according to the procedure described above:

- they both attended initial meetings to discuss the participant information, give consent to take part and mould their ears
- casts of the moulds were taken and then the ear-pieces were remoulded containing the sensors
- a second meeting provided the participants with the ear-pieces and instructions on their use
- finally a wrap-up meeting was held at which feedback was received regarding the use of the systems.

### **9.4.1 American football**

The risk of concussion in American football is particularly topical at this time. The film *Concussion* was released on the 25 December 2015. This film tracks the path of accomplished pathologist Dr. Bennet Omalu who, on the basis of autopsy findings reported on brain damage in football players who suffered repeated concussions in the course of normal play and applied the label 'chronic traumatic encephalopathy' to this scenario. This followed other concerns with, for instance, Taylor (1967) suggesting that because of the "dangerous" protective devices worn there were abundant records of college boys who never played American football again after a head injury. Very recently, the NFL (National Football League) has announced awards of \$ 30 million for research projects attempting to reduce the likelihood of concussion being sustained during American football.

In this trial, the participant agreed to wear the sensor system in the last game before a mid-season break. The usual and full protective equipment was to be worn during the game, including pads and a helmet. The pads provided an obvious location for attaching the data acquisition box. The box at the moment may be considered to be in its development phase, and is therefore, larger than would be ultimately necessary for the functions it performs. Even being this size, the data acquisition box was easily fitted on the pads without cause for concern. The specific style of pads contained a slot in the back in which the box was taped.

The wires leading to the ear-pieces were routed through the pads, out from the shirt, up the neck and under the helmet. The ear-pieces were provided with sufficient length of cables so that they did not restrict movement of the head, beyond the constraints imposed through wearing the pads and helmet themselves. However, according to the participant, he was worried about damaging the system every time the helmet was removed. In a sport such as American football it is not realistic to expect that a player wears their helmet for the whole game. The breaks for different plays and with offensive, defensive and special teams lead to situations where players are resting on the side-lines for long enough that they would expect to remove their helmet. Therefore, improvements to the robustness of the wiring and connections, or at least

the perception of robustness, would be needed for a full trial of data collection in this setting.

An anticipated comment (given the necessary wearing of earplugs) was that the ear-pieces influenced, and decreased, the ability of the player to hear. Hearing wasn't a requisite for this team as significant tactical 'calls' were made through the use of hand signals and there are few crucial sounds made throughout 'plays'. However, the participant did comment that sounds were muffled and this did present a potential inconvenience in that not all instructions would be received clearly. He noted that whilst the sounds were muffled, he could still hear ambient noises sufficiently so as to avoid any disorientation.

The participant also reported that it was awkward to fit the ear-pieces. They were confused as to which piece was left and right and more generally had trouble locating the pieces accurately in the ear. As one might expect, there were no mirrors on the football pitch to aid with this process. Despite this the participant expressed that, once fitted, the ear-pieces were not uncomfortable.

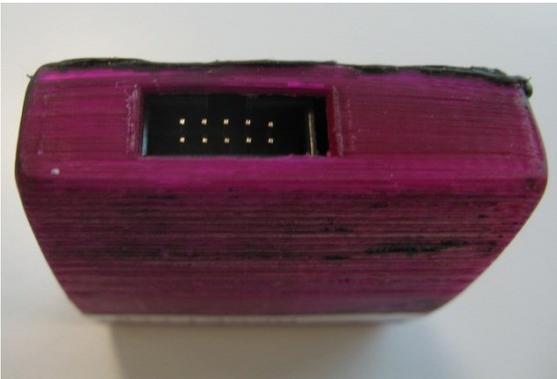
Apparently, there was one fundamental further issue with data collection from the American football trial. This was that the participant was concerned over the time available for data recording. The location of the data acquisition unit on the pads meant that it could not be easily accessed without first removing, the helmet, shirt and then pads. The normal progression for the participant was to get changed ready to play football, then warm-up, then play the four quarters of the match and then warm down, all before removing his shirt in a way that would facilitate access to the data acquisition box.

### 9.4.2 Motorcycling

Another trial participant was recruited to wear the instrumented ear-pieces whilst motorcycling. Bespoke ear-pieces and the other hardware necessary for data acquisition were provided.

Initial feedback was provided by the participant. He had tried wearing the sensors one day, but had noticed that the ear-piece cables had pulled free from the data acquisition box whilst he had been wearing them. As mentioned in Section 9.3, this causes the box to jam in a mode where no further data are sampled and the existing data from the session cannot be cached to disk. The box can be retrieved from this situation by resetting it, at the cost of the data from the session (it is not finished appropriately to allow future access).

The eight-pin port in the data acquisition box, for the cable connection to the ear-pieces can be seen in Figure 9-35. The corresponding terminus for the cable is shown in Figure 9-36. This latter figure also shows the small retaining clip to hold the cable in place, which is easily overcome.



**Figure 9-35: Box housing the data acquisition unit as used in the trials**



**Figure 9-36: Connector on the cable linking the ear-pieces with the data acquisition unit**

Following the identification of this issue, the motorcyclist was able to complete three rides whilst wearing the ear-pieces and collecting data. In each case the ride lasted for about 30 minutes.

No significant head contact events were noted by the participant. Therefore the data corresponded to normal riding conditions. Specifically, the ride was described as:

- 3-5 minutes of pre-roll (getting from garage to the road)
- 20-25 minutes of riding (heavy traffic in some places)
- 3-5 minutes cooling down until the unit could be switched off.

This duration of ride created approximately 1.8 million data samples from each of the 12 sensitive axes. Data files were cut into 5 minute (8.3 Megabyte) tranches. When imported into Excel™, these files generated a Spreadsheet of approximately 73 Megabytes. Therefore, this quantity of data is manageable using conventional means, readily available to everyone with a modern computer.

As noted in Section 9.1.5, the responses from the sensors were noisy. There were random fluctuations in the accelerometer data in excess of  $\pm 2 g$ . Despite there being no significant head contact or loading events, the peak accelerations from the trial data were between  $10.7 g$  and  $-8.2 g$ . The peak angular velocities were between  $412$  and  $-366$  degrees per second. These results indicate that these sensors in the current configuration (circuitry and ear-piece mounting) would not be suitable for investigating impacts of severity lower than these values; for instance, less than about  $\pm 10 g$ . As such events could not be detected reliably above other noise.

### **9.4.3 Summary from user trials**

Some clear conclusions can be drawn already from the extremely limited trials conducted. These are summarised briefly here:

- The process of obtaining bespoke moulded ear-pieces for participants is a reasonable one

- It seems relatively straight-forward for somebody to use the mould-your-own ear plug kits to generate a mould of their ears
- After several trials, a procedure for casting the ear plugs and creating other positive impressions has been found and validated
- It is possible to fit sensor units in the earpieces without creating hard or sharp edges protruding through the sides
- Whilst an audiologist may offer a more professional service, the moulding process demonstrated here can be followed to produce results in a shorter period and potentially with less inconvenience for a participant
- Whilst they may not be easy to fit in the ears, once there the ear-pieces produced for this study are comfortable to wear.
- In some ways the data acquisition option provided to accompany the ear-pieces can be worn and used
  - Being of a small size, a American football player and a motorcyclist thought that it would be possible to wear the sensors and data acquisition box whilst taking part in their activity
- In other ways the data acquisition option provided to accompany the ear-pieces needs improvement to make it robust for use in such applications
  - The inadvertent removal of the ear-piece cables is unacceptable as it jeopardises all data captured to that point
  - A threshold trigger should be incorporated to help in allowing the system to be active for long periods without draining the battery and to make post-event analysis a more sensible proposition in terms of effort
  - A user's manual would be helpful to interpret the coded messages that can be flashed through LEDs on the side of the unit. For instance, indicating that there is a problem accessing the SD card.

- If the ear-pieces were intended to be used to detect impacts around the onset severity for minor head injuries only, then consideration would need to be given to improving the signal to noise ratio
  - Filtering may be necessary to smooth random fluctuations in the data (though care would have to be taken to preserve the significant characteristics of the response during the loading event).
  - Otherwise, it might be that the power supply decoupling capacitors could be improved in specification or location on the circuit board. Here care would have to be taken to ensure that circuit board size did not increase so as to compromise the mounting within the ear-canal.



## **10 Discussion – Application in transport safety**

The original ambition of this study was to generate understanding of the fundamental mechanisms governing head injury in road traffic accidents. Inadequacies of the existing data to support such research were shown through a novel analysis. It was hoped that a new sensor system could be developed to capture information from events where a head is impacted. This led to the research question, 'Is it possible to validate advanced head injury criteria and head models using additional (new) head injury case data so as to make their application more robust in efforts to mitigate future injuries?' In response to this question, a sensor system planned for use in motorsports was evaluated. Furthermore a prototype sensor system was developed and its suitability for use in generating novel head injury case data was evaluated. Several limitations have been found with the prototype system and its potential use in this application; though, potentially, most of them could be resolved with additional effort. However, the issues do not prevent some comments being made with regard to the ability of this research to address the data need and respond to the research question posed.

### **10.1 Usability**

Concussion is consistently the most commonly reported injury in professional rugby (England Professional Rugby Injury Surveillance Project Steering Group, 2015). There are also a large number (~ 300,000; Mihalik et al., 2007) of American football players expected to receive a mild traumatic brain injury each year in the U.S. Given some high-profile recent examples of players being suspended for concussion and the potential cumulative effects of concussive and sub-concussive blows (Hazrati et al., 2013), it seems reasonable to expect that some sports participants would be willing to wear instrumentation to collect data regarding such events. A range of technical options for data collection are now appearing on the market. Some are high fidelity solutions that can be mounted to the helmet, e.g. HITS. However, most seem to be lower fidelity systems (temporarily mounted to a helmet or head-band) used for counting head contact events and recommending a medical evaluation for that player

when a threshold is passed. This second type of system does not offer access to the accurate event data that could be achieved through the wearing of instrumented ear-pieces, particularly for high severity events beyond the acceleration ranges for which those instrumentation systems were designed.

If the moulding process adopted in this study can be employed successfully, then there would be very little inconvenience for a participant to receive ear-pieces. These can be returned in less than a day. Therefore the wearing of an ear-piece does seem to be a viable alternative to other methods for attaching a sensor to the head or helmet.

The few responses received so far suggest that wearing an ear-piece is unobtrusive in the short-term and generally does not interfere with other activities. This may be different in sports or other applications where hearing is required for participation, orientation or safety. For those sports where these issues are not a primary concern, then ear-piece instrumentation seems to be a valid approach for generating data relating to head injury mechanisms.

It should be noted that this research was not able to demonstrate the benefit of fitting instrumentation into the ear rather than elsewhere on the head or helmet, robustly. It would have been interesting to compare in-ear sensor data with that from sensors positioned in the mouth (in a mouthguard), on the neck (for instance, via the xPatch) or attached to a helmet. A comparison of ear and tooth mounting strategies was recently completed in the U.S. (Christopher et al., 2013), showing agreement with each other, but not using the accelerometer mounting methods available now. Therefore, it is suggested that this comparison forms a necessary part of any purchasing strategy for a sensor system user group. From a biomechanical perspective, it is suggested that further evaluations should be undertaken to guide the development of the most useful sensor placement approaches, giving the best possible accuracy of data.

## 10.2 Injury prediction

For use in transport safety, a new injury criterion would need to relate to the prediction of the primary injury types that need attention. The remaining high priority injuries requiring prevention for car occupants and vulnerable road users, at least in terms of societal costs, are severe (for example: diffuse brain injury, traumatic subdural haemorrhage, and traumatic subarachnoid haemorrhage). This is different to the situation in sports, as mentioned above, where the frequency of concussive and sub-concussive blows makes them the priority for research and injury mitigation; and therefore is likely to require a different dataset for risk analysis. To provide a reasonable prediction of risk, the criterion values should be related to both injury and non-injury cases. Therefore, data must be collected from conditions where these injuries may occur. This research has shown that valuable data can be captured from conditions representing a high severity motorsport event using ear-piece accelerometers. It is excellent to see that such a system is now being used in Formula 1. In the event of a head contact occurring for a driver, it is hoped that this system will provide valuable data for the research community in the future. However, if data collection is limited to Formula 1 or top-level motorsports, then it may take a substantial period to collect sufficient data to address the research need. About 20 cases are needed to support regression for each variable in injury risk function analyses, then for the simplest analysis 20 cases are needed, ideally with 10 where an injury occurred and 10 with no injury. Assuming that there might be 1 or 2 substantial head contact events in a year in Formula 1 then it could be 10 or 20 years before this data collection avenue yields sufficient information for robust statistical analysis. A further complication arises if the data cannot be released immediately due to confidentiality issues. In that case, further delays can be foreseen.

Therefore it seems a reasonable proposition that other sports where high impact events may occur should be targeted for additional data collection. It is hoped that a future iteration of the prototype sensor system developed alongside this research aids that process. Typically, it is likely to be motorsports where sufficient energy is routinely available and there is a risk of severe head injury. The regular use of a

helmet and the proven ability to wear instrumented ear-pieces under a helmet makes it seem reasonable to expect that more motorsport participants could adopt the wearing of sensors in the future.

One of the confounding factors identified within this research was the wearing of a helmet. Therefore, it should still be a research goal to collect data from events where a helmet is worn and those where a helmet is not worn. There would need to be injury and non-injury cases in both types of event and data in sufficient quantity to identify and control for other variations between the two groups. No regular source of information on impact events where no helmet is worn, but there is a substantial likelihood of a severe head injury occurring, has been determined. Instead it would be necessary to show that the variety of impact conditions from the helmeted head group can be extrapolated to the impact conditions seen by car occupants and vulnerable road users. Generally, padding is used to provide some protection for vulnerable heads such as in car interiors and airbags; so it is not unreasonable to expect that data should be transferable, to some extent (provided that it is measured at the head, not on a helmet).

### **10.3 Risk validation**

Before being used in the process of designing safer transport systems there is a need to demonstrate that a new criterion can be used to drive designs in the correct way. There should be some validation that systems designed to offer more protection are associated with a lower predicted risk of injury. For instance, with the current criteria there is an intuitive inference regarding linear accelerations that softer head impacts would be associated with lower acceleration values and a lower risk of head injury. Such a simple relationship may not be evident with new criteria. Therefore, there is a need to demonstrate this relationship by other means. This has not been done reliably for existing kinematic criteria.

The availability of six degree-of-freedom head motion data together with information about injury outcome would be immensely valuable in this pursuit of validating injury risk estimates. However, to provide data tailored to the car occupant or pedestrian or pedal cyclist it may be that specific accident events must be reconstructed that represent typical real-world impact conditions. It is not clear in this process how in-ear instrumentation will be of benefit, unless there is such a take-up of a commercial system that many car occupants, pedestrians and pedal cyclists begin wearing them throughout their normal daily transport activities. Despite the advances in technology seen during the duration of this research, the widespread adoption of high-fidelity, wearable instrumentation in this way still seems unlikely at the moment.



# **11 Summary**

## **11.1 Introduction**

Road traffic injuries claim many lives each year and have a huge impact on health and development and with regard to societal cost. To continue efforts to prevent or mitigate these injuries it is necessary to understand how the general loading to the head during a road traffic collision relates to loading of the specific structures prone to injury and the risk of injury. With previous research, there has already been much iteration of the kinematic theories concerning the motion of the head, the skull and intracranial contents, and the relation to injury. However, there has been reliance in transport safety system design on the relatively simple measures of linear acceleration and perhaps the Head Injury Criterion (HIC). Now, with a focus on injuries to the intracranial contents, there is a need for a new kinematic criterion to guide design. However, the existing proposals have not been well validated. One of the reasons for this is that the data available to validate new criteria are not suitable. No single study is large enough to investigate all confounding factors in head injury outcome. When several studies are combined by using the data available from the literature, little additional statistical power is generated because the experimental designs are often not compatible and not all factors are always reported. This research therefore has shown that additional data are needed for the purpose of validating the link between general head kinematics and specific injury risks.

## **11.2 Addressing the research question**

The research question was posed was; is it possible to validate advanced head injury criteria and head models using additional (new) head injury case data so as to make their application more robust in efforts to mitigate future injuries?

At present there are substantial concerns over concussion in sports and the potential impairments that could be caused after mild traumatic brain injuries. This has led to several instrumentation systems being developed to detect a potentially concussive blow to the head. However, in transport systems there is also a need to consider more severe injuries and to generate predictive models based on high fidelity data. Therefore, beyond the development of these systems, efforts were directed within this research to the specification of a high fidelity, high impact severity unit that could be used to provide head impact data for future injury criterion development and validation to support injury prevention strategies for road traffic accidents. The potential for such a solution to address the research question is summarised in the following sections.

### **11.3 Contributions of the study**

The strengths and weaknesses of an in-ear accelerometer system used by the FIA Institute and now employed in Formula 1 racing were determined through experimentation with that system. Adoption and use of this technology provides the potential to investigate high impact severity loading events and their potential to cause head injuries. However, there are two primary reasons why this source of data will provide only a part of the solution to the data need. Firstly, since the sensors were launched there have only been 20 to 22 wearers of the system (at any one time), during about 20 events a year. Whilst there have been some high profile head impacts during this time, the safety of the sport means that head contact events are still relatively rare. Hence it will take many more years until sufficient data are available with which to construct statistically reliable injury risk estimates. Secondly, as the participants are all 'stars' with fanatical following for them and the sport, it is likely that somebody will be able to link results to a certain person and event; even if they are published anonymously. It is difficult to see how data can be provided to third parties for analysis whilst also honouring confidentiality for the people involved in the sport. As such, there could be delays between a head impact event occurring and data from that event being made available for analysis.

Therefore, in addition to the Formula 1 system, a low-cost solution was developed with the aim to give similar sensor performance for a wider market of potential wearers. This was developed in line with the specification presented in Section 7.

The prototype sensor system was evaluated in a small series of drop tests and also in a very small real-world data collection trial.

#### **11.4 Limitations of the study**

This evaluation identified a series of issues that need to be resolved before the system can be used to generate valuable data. Key issues are that:

- The data acquisition unit for the prototype system must be improved so that it is possible to conduct impact event trials at higher severities. For this purpose the unit should be made more resistant to shock events.
- The prototype system ear-pieces are not as robust as those used by the FIA Institute and this needs to be improved.
- The prototype ear-piece attenuates the applied acceleration and therefore the material used in the ear-pieces needs to be changed.
- The inadvertent removal of the ear-piece cables is unacceptable as it jeopardises all data captured to that point.
- A threshold trigger should be incorporated so that data are only recorded immediately around a substantive blow to the head or loading picked up by the sensors.

However, the basic process of moulding an ear-piece for a participant, including accelerometer and angular rate sensor instrumentation, seems to be reasonable. It gives little inconvenience to the participant and seems to be unobtrusive during American football and motorbike riding trials. As such the concept seems appropriate for further investigation and warrants modification of the prototype system developed here.

## **11.5 Recommendations for application in transport safety**

A working in-ear system seems to offer a potential solution to obtaining elusive data regarding the kind of impact events that could cause head injuries for road users. Other than expensive accident reconstructions, which are also associated with inherent limitations and potential sources of error, it is not clear from where else such valuable information could be derived. As such this research has identified a potential solution for collection of data pertaining to the mechanisms of head injury in road traffic accidents. It is recommended that this approach is adopted as quickly as possible and as widely as possible to support a growing need for the data it can generate. Two practical systems have been evaluated within this work. One system has been adopted by Formula 1 and has already been effective in generating data from head impact events. A second system was generated in this research to support a more widespread access to this technology. This latter system used a similar approach to placing the sensors in the ear, but without the professional moulding of the ear-piece for the individual wearer. However, whilst this technology was demonstrated in the evaluation documented in Section 9, a viable system is not ready immediately, but could be following modifications to the prototype system evaluated.

## **11.6 Recommendations for future work**

After modifying the system to address the issues identified throughout this research, it is recommended that such a sensor system is worn by sports people other than those participating in Formula 1. Ideally this would demonstrate a suitably low risk for the wearers that a trial could be conducted in a situation where a protective helmet is not worn by the participants. Before that time, one could imagine sensors being worn during horse racing or equestrian events as the next trial group.

As sensor technology moves towards smaller and higher fidelity devices inexorably, it is important that opportunities are taken to adopt that technology in potential sensor systems. The advantages of a higher data sampling frequency in the fidelity of the potential data from these systems has been demonstrated in this research and therefore should be kept in mind for future system specifications. Additionally, progress will be made with the accuracy and robustness of angular rate sensors or accelerometers. This is important with respect to obtaining the six degree of freedom measurements the injury criteria development research calls for.

Additionally, it is still suggested that further validation work is necessary to confirm the advantages of having the sensors mounted in the ear canal, over other commercially available alternatives. This will involve further extension of the validation work already undertaken and making use of the latest version of prospective sensor and data acquisition systems.



## 12 Conclusions

Road traffic injuries claim many lives each year and have a huge impact on health. To make further advances in head injury protection, it will be necessary to understand how the general loading to the head during a road traffic collision relates to the risk of specific injuries.

Analysis of existing data sources indicated that additional data are needed for validating the link between general head kinematics and specific injury risks. This finding led to the research question, "Is it possible to validate advanced head injury criteria and head models using additional (new) head injury case data so as to make their application more robust in efforts to mitigate future injuries?"

Efforts were directed towards the development of two instrumentation systems to quantitatively measure the severity of a blow to the head and generate new data. Hence, two practical systems were evaluated within this work.

1. In-ear accelerometers, as used in Formula 1;
2. A low-cost alternative solution, developed to the prototype phase.

The evaluation identified a series of issues that need to be resolved before the prototype system could be used to generate valuable data. However, the basic process of moulding an ear-piece for a participant, including accelerometer and angular rate sensor instrumentation, seems to be reasonable.

In-ear sensor systems seem to offer a potential solution to the collection of data pertaining to the mechanisms of head injury in road traffic accidents. It is recommended that this approach is adopted as quickly as possible and as widely as possible to support a growing need for the data it can generate. It is recommended that a similar system is worn by sports people in addition to those participating in Formula 1 and top-level motorsports.

System developments should take advantage of sensor technology improvements. Any new system will need equivalent validation to that developed here.

It remains necessary to demonstrate the fidelity associated with placing sensors in the ear, compared with other placement options.

## **Acknowledgements**

The work described in this thesis was carried out in the Safety and Technology Group of the Transport Research Laboratory. The author is grateful to David Hynd and Andrew Parkes for their support and advice to this point in the research. The author is also grateful to the TRL Academy and Engineering & Technology Division for providing funding for the PhD research effort, to Andrew Hannay for his technical help with the prototype electronics and to the FIA Institute for provision of hardware to test.

I am also extremely grateful for the understanding and compassion shown by colleagues, friends and family as the research competed with other commitments throughout the past years. Particular thanks to Maggie who has had to cope with all of the 'fall-out' from me getting stuck into this.

## **Dedication**

I dedicate this thesis to Robyn and Theodore. The best distractions from study I could hope for.



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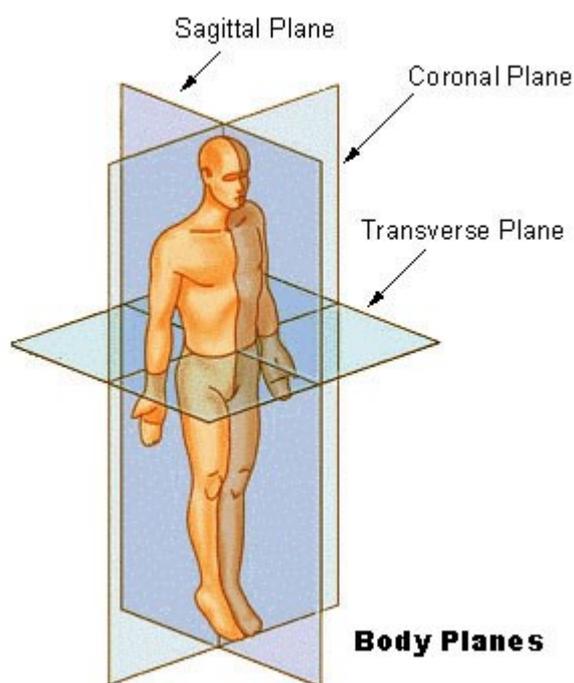
## Glossary of terms and abbreviations

**AAAM** – Association for the Advancement of Automotive Medicine.

**Abulia** – The lack of will or willpower, or the initiative to act decisively.

**AIS** – Abbreviated Injury Scale (AAAM, 2008).

**Anatomical planes** – A sagittal plane is a vertical plane passing through the body dividing it into right and left sections. The median sagittal plane passes through the centre of the body giving equal right and left halves (see Figure G-1). Coronal planes are also vertical planes through the body but which are at right angles to the sagittal planes. Finally, transverse planes are horizontal and at right angles with both the sagittal and coronal planes.



**Figure G-1: Anatomical planes**

**Aphasia** – Loss or impairment of the capacity to use words as symbols of ideas.

**Apoptosis** – The process of programmed cell death.

**APROSYS** – Advanced PROtection SYStems Project, an FP6 (Sixth Framework) Project for the European Commission.

**Astrocytic** – Relating to astrocytes which are star-shaped glial cells involved in support around the blood-brain barrier, provision of nutrients, maintenance of extracellular ion balance, etc.

**Atrophy** – The physiological process of wasting away, through the breakdown and reabsorption of tissue.

**Auditory meatus** – The ear canal.

**BrIC** – Brain rotational Injury Criterion.

**Caudal** – Of, at, or near the tail or hind parts. It can mean posterior or inferior depending on the axis or body part being described.

**CP** – Combined Probability.

**Crosstalk** – Values measured in one axis when oscillations are applied in a perpendicular direction.

**CSDM** – Cumulative Strain Damage Measure.

**DAI** – Diffuse Axonal Injury.

**DDM** – Dilational Damage Measure.

**Decerebrate posturing** – An abnormal body posture that involves rigid extension of the arms and legs, downward pointing of the toes, and backward arching of the head.

**Dendrites** – Extension of a nerve cell along which the impulses are transmitted.

**Diencephalon** – The diencephalon is located deep in the brain underneath the cerebrum and above the pituitary gland. It is the caudal (posterior) part of the forebrain which contains the thalamus and the hypothalamus. It is the link between the nervous system and the endocrine system.

**EDR** – Event Data Recorder (alternatively ADR; Accident Data Recorder)

**Epithelial** – Membranous (like a skin).

**EuroSafe** – European Association for Injury Prevention and Safety Promotion.

**Extravasation** – To pass bodily fluid out of its proper place.

**FIA** – Federation Internationale de l'Automobile.

**FIM** – Functional Independence Measure.

**FMVSS** – Federal Motor Vehicle Safety Standard.

**g** – acceleration shown in units representing the acceleration due to gravity on the Earth's surface.

$$1g = 9.8 \frac{m}{s^2}$$

**GAMBIT** – Generalised Acceleration Model for Brain Injury Threshold.

**GCS** – Glasgow Coma Scale.

**Genu** – The anterior end of the corpus callosum.

**Glia** – Glia, or the neuroglia, are the fibrous and cellular non-nervous supporting elements of the nervous system.

**GOS** – Glasgow Outcome Scale.

**GOSE** – extended Glasgow Outcome Scale.

**GSI** – Gadd Severity Index.

**Hemiparesis** – Paresis (slight paralysis, loss of muscular power) of one side of the body.

**HIC** – Head Injury Criterion.

**HIP** – Head Impact Power.

**HITS** – Head Impact Telemetry System.

**Hydrocephalus** – Also called hydrocephaly, an increase in the volume of cerebrospinal fluid within the skull.

**HYGE™** - a producer of reverse accelerator systems for applying high 'g' accelerations. The subject is reversed with respect to the accelerator or sled which is then accelerated towards them so as to simulate a deceleration impact event.

**Hz** - Hertz; the unit of frequency, defined as one cycle per second.

**IBH** - Instrumented Boxing Headgear.

**ICD** - International Classification of Diseases.

**IDB** - The European Injury DataBase.

([http://ec.europa.eu/health/data\\_collection/databases/idb/index\\_en.htm](http://ec.europa.eu/health/data_collection/databases/idb/index_en.htm))

**Inion** - The most prominent projection of the occipital bone low down at the back (posteroinferior) of the skull.

**ISO** - International Organization for Standardization.

**ISS** - Injury Severity Score. Calculated by taking the highest AIS severity code in each of the three most severely injured ISS body regions, squaring each AIS code and adding the three squared numbers together for the ISS.

**IVH** - Intraventricular haemorrhage.

**JHTC** - JARI (Japan Automobile Research Institute) Human Head Tolerance Curve.

**Lagrangian strain** - The difference between the current and original length, divided by the original length.

**LHS** - Left hand side.

**MEMS** - MicroElectroMechanical Systems.

**Mesencephalon** - The mesencephalon, or 'mid-brain' is a part of the brain stem. It is the short, constricted portion which connects the pons and cerebellum with the diencephalon and cerebral hemispheres. It is associated with vision, hearing, motor control, sleep/wake, arousal (alertness) and temperature regulation.

**Microglial** - Microglia are a type of glial cell forming the active immune defence in the central nervous system.

**Mitochondria** – An organelle found in cells which is responsible for respiration and energy production.

**MSC** – Mean Strain Criterion.

**NFL** – National Football League {official site: [www.nfl.com](http://www.nfl.com)}; is the highest level professional American football league.

**NHTSA** – National Highway Traffic Safety Administration.

**NMSC** – New Mean Strain Criterion.

**Oligodendrocytic** – Relating to the oligodendrocytes which provide the myelin sheath for axons.

**Parietal** – Forming a wall.

**Petechial haemorrhage** – A minute rounded spot of haemorrhage on a cross-sectional surface of an organ.

**Plexus** – A network of interlacing nerves or anastomising (reconnecting) blood vessels or lymphatics.

**PMHS** – Post mortem human subject.

**PRHIC** – Power Rotational Head Injury Criterion.

**RHS** – Right hand side.

**RIC** – Rotational Injury Criterion.

**RMDM** – Relative Motion Damage Measure.

**RMS** – ‘root, mean, square’ or the quadratic mean is the square root of the arithmetic mean of the squares for a set of numbers.

$$x_{RMS} = \sqrt{\frac{1}{n}(x_1^2 + x_2^2 + \dots + x_n^2)}$$

**Rostral** – The top, or head, of the brainstem (i.e. towards the cerebrum, as opposed to the spinal cord).

**SF-36** – Short Form 36.

**SFC** – Skull Fracture Correlate.

**SIMon** – The Simulated Injury Monitor finite head model.

**Soma** – The cell body of a neuron.

**SUV** – Sports Utility Vehicle.

**TEC** – Translational Energy Criteria.

**THIM** – Translational Head Injury Model

**TRL** – The UK's Transport Research Laboratory; TRL Limited.

**von Mises Stress** – The von Mises - Hencky criterion is a formula for calculating whether the stress combination at a given point will cause failure. Three orthogonal stresses are combined into an equivalent stress (index number), which is then compared to the yield stress of the material. If the "von Mises stress" exceeds the yield stress, then the material is considered to be at the failure condition (definition from various internet sources).

**WHO** – World Health Organization.

**WSUHIM** – Wayne State University Head Injury Model.

## **Appendix A Field trial participant information**

### **A.1 Advertisement for trial participants**

#### **Advert**

### **Participants needed to help with TRL project**

#### **Background**

We are currently undertaking research to produce a wearable micro-accelerometer. We need help to evaluate the prototype that has been produced.

#### **What we need**

Do you take part in any activity that requires a full helmet to be worn, like a motorsport or motorcycling, perhaps ice-hockey or American football? The activity should include the wearing of a helmet which covers your ears.

We will be asking you to use the micro-accelerometer while you take part in this normal activity. Through the fitting of a bespoke (custom-moulded) ear-piece the micro accelerometer will be placed in your ear. Your opinion will then be sought on whether the accelerometer was a burden during the activity, or not.

#### **Time Scale**

We require the activity to be taking place before the end of November.

If you are willing and able to help out, please contact Jolyon Carroll ([jcarroll@trl.co.uk](mailto:jcarroll@trl.co.uk) / 0564) or Julie Austin ([jaustin@trl.co.uk](mailto:jaustin@trl.co.uk) / 0137).

## **A.2 Participant information sheet**

### **Information for research participants**

#### **Mechanisms of head injuries, in-ear accelerometer trial:**

Head injuries account for a large proportion of sporting injuries and injuries caused by road traffic collisions. To aid research into the key mechanisms responsible for the injuries it is necessary to understand the loading to the head during injurious and non-injurious events. This research is necessary to support further advances in helmets and other safety system performance, as well as in the adoption of procedures to prevent potentially injurious events from occurring in the future.

TRL has developed a low-cost sensor system which could be used to measure the acceleration of a head during a sporting event. It is intended that the sensors are fitted in the participant's ears to provide a close linkage with the head. Whilst this system is at a prototype stage, it is developed sufficiently to consider use in a sporting application and requires evaluation of its suitability as a wearable device and the level of inconvenience it imposes for a sports participant.

This trial comprises a step towards determining if the sensor system is a viable tool for gathering data about events where the head is loaded. The aim is to show whether or not it is reasonable to expect a participant to wear such a sensor system during various activities. It is a final piece in the PhD studies of Jolyon Carroll, looking at the availability of sensors to capture high-fidelity data from head loading events.

The trial will run over a single event only, for each participant recruited.

You have been approached because your suggested activity comprises a candidate event where ear-pieces can be worn under a helmet. There may also be a remote chance that your head could be loaded during routine participation or in the event of an accident. The likelihood of an injurious impact occurring, however probable or

improbable, has no bearing on participant selection as the trial is focussing on convenience for the participant and not the generation of data, at this stage.

As a participant in this trial you will be expected to:

- Participate in an initial meeting (potentially via telephone) to enrol in the study and arrange future logistical requirements
- Have your ears cast ready for moulding of the in-ear system
- Participate in a second meeting to setup the sensor system ready for use
- Assuming that feedback is positive and agreement is reached to use the system during an 'event', complete that whilst wearing the system
- Participate in a wrap-up meeting to allow; retrieval of data and reporting on the aspects associated with wearing the device

By using a custom moulded ear-piece it is assumed that no additional risk of injury will be generated during the event. If this gives any cause for concern to you, then the trial must be stopped immediately and terminated on that basis.

The two part silicone used to mould the ear-piece is medical grade and is therefore considered to be, in the most part, non-allergenic. The manufacturer has never had problems with anyone before being allergic to this product; however, they recommend that the trial excludes potential participants if they have dermatitis. Therefore, the presence of broken skin, dermatitis or psoriasis around the ears will preclude a potential candidate from participation.

Please Note: The ear-pieces should be cleaned often in warm mild soapy water and dried thoroughly before use. This is important if they are fitted and removed multiple times before data collection during the activity or event. The ear-pieces should have a smooth exterior surface without any cracks or holes in it. They should not be worn if there is any doubt about the structural integrity of the units.

In the unlikely event of a bang to the head or a head injury occurring; the data will still be retrieved and held by TRL. If you do not wish for these data to be used in any future research, that is up to your discretion. As mentioned, this trial is predominantly about the comfort associated with the sensor system, not data generation.

No direct benefits are offered for participating.

Once the study ends, your feedback will be collated and used in conjunction with the reporting of the sensor development in Jolyon's PhD thesis. Without your objection, the data will also be used to help set a non-injurious baseline level of head loading where we can be certain no injury is likely to occur.

No personal data will be kept alongside the sensor system outputs. Only a statement regarding the type of event undertaken and the injury outcome will be kept with the sensor data – that is to acknowledge and record that 'no injury was sustained during the trial period'.

Support for Jolyon's PhD and this trial is being provided by TRL.

You are able to withdraw from the study at any point and are encouraged to do so if:

- the sensor system makes you feel at any greater risk of injury than under normal circumstances
- you experience any discomfort wearing the sensors or in the moulding process

If you have any concerns or queries, please contact me:

Jolyon Carroll

PhD student and Principal Researcher, Safety and Technology Group,  
Engineering and Assurance Division, TRL;

Crowthorne House, Nine Mile Ride, Wokingham, Berkshire, RG40 3GA:

Telephone: 01344 770564

Email: [jcarroll@trl.co.uk](mailto:jcarroll@trl.co.uk)

### A.3 Participant consent form

#### Consent form

#### Mechanisms of head injuries, in-ear accelerometer trial:

Jolyon Carroll;

PhD student and Senior Researcher, Safety and Technology Group, Engineering and Assurance Division, TRL;

Crowthorne House, Nine Mile Ride, Wokingham, Berkshire, RG40 3GA:

**Please initial box**

1. I confirm that I have read and understand the information sheet for the above study and have had the opportunity to ask questions.
  
2. I confirm that I do not have dermatitis around the ear, do not have an ear infection, perforated ear drum or have had ear surgery.
  
3. I understand that my participation is voluntary and that I am free to withdraw at any time, without giving reason.
  
4. I agree to take part in the above study.

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Name of Participant	Date	Signature
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Name of Researcher	Date	Signature
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