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1 **Finite element analysis of the effect of loading curve shape on brain**
2 **injury predictors**

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4 Original Article

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11 **Finite element analysis of the effect of loading curve shape on brain** 12 **injury predictors**

13 Prediction of traumatic and mild traumatic brain injury is an important factor in
14 managing their prevention. Currently, the prediction of these injuries is limited to
15 peak linear and angular acceleration loading curves derived from laboratory
16 reconstructions. However it remains unclear as to what aspect of these loading
17 curves contributes to brain tissue damage. This research will use the University
18 College Dublin Brain Trauma Model (UCDBTM) to analyze three distinct
19 loading curve shapes meant to represent different helmet loading scenarios. The
20 loading curves were applied independently in each axis of linear and angular
21 acceleration, and their effect on currently used predictors of TBI and mTBI.
22 Loading curve shape A had a late time to peak, B an early time to peak and C had
23 a consistent plateau. The areas for all three loading curve shapes were identical.
24 The results indicate that loading curve A produced consistently higher maximum
25 principal strains and Von Mises Stress than the other two loading curve types.
26 Loading curve C consistently produced the lowest values, with loading curve B
27 being lowest in only 2 cases. The areas of peak Von Mises Stress and Principal
28 strain also varied depending on loading curve shape and acceleration input.

29 Keywords: Brain injury; concussion; finite element modelling

30 **1. Introduction**

31 In the past, researchers using animal models have linked mechanisms of injury to
32 damaging levels of brain deformation caused by linear and angular loading curves
33 (Viano et al., 1989; Kleiven, 2007). These loading curves reflect head motion and are
34 represented by three dimensional linear and angular head kinematics resulting from
35 either a blow or resulting from inertia (O'Donoghue, 1999). Research has supported this
36 phenomenon showing that different injuries can be created by changing the
37 characteristics of the linear and angular acceleration loading curve (Gennarelli et al.,
38 1979; Adams et al., 1981; Kleven 2003). To reduce the risk of brain injuries, helmets
39 work by changing the characteristics of the linear acceleration curve. Specifically,

40 helmets lengthen the duration of the impact and lower the magnitude in an effort to
41 prevent severe brain trauma.

42 To prevent these injuries from occurring, researchers attempt to establish a threshold
43 whereby the damage would occur. Currently, the engineering parameter used to predict
44 the extent and severity of these injuries is peak resultant linear acceleration. This is
45 reflected in the use of peak linear acceleration for the certification of safety devices such
46 as helmets (Hoshizaki and Brien 2004). While the use of peak resultant linear
47 acceleration has been successful in reducing the prevalence of traumatic brain injury
48 (TBI) in sport, mild traumatic brain injury (mTBI) remains prevalent (Wennberg and
49 Tator, 2003). This suggests that linear acceleration alone is not capturing the spectrum
50 of brain trauma.

51 Recent research has further demonstrated the limitations of using peak resultant
52 linear acceleration to characterize the protective capabilities of football and hockey
53 helmets (Post et al., 2009; 2010). Using a Hybrid III headform and neck Post et al.
54 (2010) impacted helmets in a series of centre of gravity and off axis impacts. From this
55 research they showed that for certain locations similar values for linear acceleration can
56 be achieved with differing rotational accelerations. When the linear and rotational
57 acceleration outputs were used for a finite element model analysis it was discovered that
58 low linear acceleration values when high rotational acceleration was present can
59 produce potentially injurious levels of maximum principal strain when compared with
60 cadaver tolerance data. This research and others suggests that it may be prudent to
61 consider the characteristics of the linear and rotational acceleration loading curves such
62 as total duration and not just peak values when determining risk of injury from injurious
63 brain deformations (Yoganandan et al., 2008; Post et al., 2010). Further research has
64 been conducted using FEM to identify possible thresholds of injury for brain tissue

65 deformation through reconstructions of real world impacts with promising results
66 (Willinger and Baumgartner, 2003; Zhang et al., 2004; Kleiven, 2007). However, the
67 researchers were focused on identifying peak variables and did not consider how the
68 input loading curve characteristics and loading axis influenced brain deformation.

69 Improvements in computer modeling have allowed the use of finite element analysis
70 as a tool for predicting head injuries and subsequently improve the design of protective
71 equipment (Zhang et al., 2001; King et al., 2003). Finite element modeling allows for
72 the measurement of brain deformation of an impact through the 3 vectors of linear and 3
73 vectors of angular acceleration over time for a full representation of the event. This
74 allows for the inclusion of direction and magnitude as well as other loading curve
75 characteristics as influencing factors of brain deformation (Horgan and Gilchrist, 2003).

76 The purpose of this research was to investigate the influence of linear and angular
77 acceleration loading curve shape on the location and severity of brain tissue
78 deformation. This research has applications to not only protective technology design but
79 also in understanding how the model used in this study responds to controlled inputs.
80 Currently protective technologies and helmets in particular are certified against linear
81 acceleration when brain injury has been shown to be influenced by both linear and
82 angular acceleration. Finite element modelling allows for the interpretation of linear and
83 angular acceleration loading curves and how they influence brain deformation. This
84 unique ability of finite element modelling will likely be used to inform manufacturers
85 how to design helmets to be more protective for a wider range of brain injuries in the
86 future. As a result it is essential that there is a fully understanding of how the finite
87 element model in use is influenced by aspects of the loading curves used as input. The
88 current study uses 3 loading curves which were taken from helmet impacts to examine
89 how the characteristics of those loading curves influenced the resulting brain

90 deformation. The end result could hopefully inform researchers as to how to minimize
91 the production of damaging brain deformation by avoiding certain acceleration loading
92 curve shapes when designing helmets for various sports.

93 **2. Methodology**

94 ***2.1 Finite element model***

95 The finite element model used in this research was the University College Dublin
96 Brain Trauma Model (UCDBTM) (Horgan and Gilchrist, 2003; Horgan and Gilchrist,
97 2004). The geometry of the model was from a male cadaver as determined by
98 Computed Tomography (CT), Magnetic Resonance Tomography (MRT) and sliced
99 contour photographs (Horgan and Gilchrist, 2003). The head and brain are comprised of
100 the scalp, skull, pia, falx, tentorium, cerebrospinal fluid (CSF), grey and white matter,
101 cerebellum and brain stem (table 1, 2). This version of the UCDBTM does not include
102 elements representing the cerebral blood vessels.

103 The model was validated against intracranial pressure data from Nahum et al. (1977)
104 cadaver impact tests and brain motion against Hardy et al.'s (2001) research. Further
105 validations accomplished comparing real world brain injury events to the model
106 reconstructions with good agreement (Horgan and Gilchrist, 2003).

107 A linearly viscoelastic material model combined with a large deformation theory
108 was chosen to model the brain tissue. The behaviour of this tissue was characterized as
109 viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation
110 modulus (Horgan and Gilchrist, 2003). The compressive behaviour of the brain was
111 considered elastic. The shear characteristics of the viscoelastic behaviour were
112 expressed through the formula:

$$113 \quad G(t) = G_{\infty} + (G_0 - G_{\infty})e^{-\beta t}$$

114 where G_{∞} is the long term shear modulus, G_0 is the short term shear modulus and β is
115 the decay factor (Horgan and Gilchrist, 2003). The hyperelastic material model used for
116 the brain in shear was expressed as:

$$117 \quad C_{10}(t) = 0.9C_{01}(t) = 620.5 + 1930e^{-t/0.008} + 1103e^{-t/0.15} \text{ (Pa)}$$

118 where C_{10} and C_{01} are temperature-dependent material parameters, and t is in seconds
119 (Horgan, 2005).

120 The CSF layer was modelled using solid elements with a low shear modulus and a
121 sliding boundary condition between the interfaces of the skull, CSF and brain was used.
122 The model is comprised of 7,318 hexahedral elements representing the brain, and 2,874
123 hexahedral elements representing the CSF layer (Horgan and Gilchrist, 2003).

124 **2.2 Input loading curves**

125 To evaluate the effects of the loading curve shape and applied vector on brain
126 deformation three sample loading curves which represent shapes which are commonly
127 produced in helmet testing were created, all with equivalent areas (figure 1, 2). Loading
128 curve A, which had a late 6015 rad/s^2 / 150 g peak at 4.4ms meant to simulate a soft
129 material response. This loading curve would represent a situation where the material
130 was too soft to manage the energy of the impact and would produce a late spike in the
131 resulting linear and rotational acceleration. Loading curve B, which had an early 6015
132 rad/s^2 / 150 g peak at 0.9ms to simulate a stiff material response. This loading curve
133 would represent a situation where the material was too dense and did not absorb any of
134 the energy of the impact, producing an early spike in linear and rotational acceleration.
135 Finally, loading curve C which had a lower 2215 rad/s^2 / 55 g peak at 0.5ms, to simulate
136 a plateau curve. This loading curve represents what would be considered an 'ideal'
137 response for an energy absorbing material in helmet design, where there is a quick rise
138 to a low plateau where the energy is managed over a longer time without bottoming out.

139 The same loading curve was modified into m/s^2 for the linear inputs. All three loading
140 curve types were applied to the UCDBTM centre of gravity individually in the X, Y and
141 Z axes. The values were meant to represent the possible peak range to create an mTBI
142 as reported by Schreiber et al. (1997), while the plateau (loading curve C) is below
143 known traumatic brain injury thresholds but consistent in total area with loading curves
144 A and B. Each of the three loading curves was run independently using ABAQUS 6.7
145 finite element solver through the three axes of linear and angular acceleration. The X-
146 axis is facing forward from the head CG, the Y-axis to the left of the head and the Z-
147 axis vertically upwards.

148 The resulting brain deformation will be measured by commonly used brain
149 deformation metrics which have been proposed to have a correlation to brain injury
150 (Zhang et al., 2004; Kleiven 2007). In this study the brain deformation metrics chosen
151 were maximum principal strain and Von Mises stress.

152 **3. Results**

153 ***3.1 Maximum principal strain***

154 Overall, the maximum principal strain results indicated that loading curve A
155 produced the highest values (table 3). For the angular acceleration inputs, loading curve
156 A produced the highest strains in the grey matter for the X axis (0.119), white matter for
157 the Y axis (0.143) and grey matter for the Z axis (0.116). For linear acceleration inputs,
158 loading curve A produced the highest values of strain in the brain stem for the X axis
159 (0.164) and in the white matter for the Z axis (0.280). For the Y axis, loading curve B
160 produced the highest strain in the brain stem (0.124). The lowest values for maximum
161 principal strain were commonly produced by loading curve C and were found in the

162 cerebellum for angular and linear inputs (0.013 – 0.038), except for the linear Z axis
163 input where the lowest value was in the grey matter (0.039).

164 **3.2 Von Mises stress**

165 The Von Mises Stress results indicated that loading curve A produced the highest
166 stress values (table 3). For the angular acceleration inputs, loading curve A produced the
167 highest strains in the grey matter for the X axis (5,796 Pa), white matter for the Y axis
168 (6,646 Pa) and in the grey matter for the Z axis (5,650 Pa). For the linear acceleration
169 inputs, loading curve A also produced the highest Von Mises Stress values in the white
170 matter for the X axis (7,107 Pa) and in the brain stem for the Y axis (6,363 Pa). The
171 grey matter was the location of highest stress for the Z axis linear acceleration (11,730
172 Pa). The lowest values were produced by loading curve C for both linear and angular
173 acceleration inputs and were commonly found in the cerebellum (578 – 1791 Pa),
174 except for linear input in the Z axis which was found in the grey matter (1,612 Pa).

175 **3.3 Maximum principal strain and von Mises stress curves**

176 Figures 2 and 3 show the maximum values of maximum principal strain and Von
177 Mises stress over time. The elements experiencing the largest deformation changes over
178 the course of the simulation and the values reported here are for the elements
179 experiencing the largest magnitude of deformation. The values are from the single
180 element in the model which incurred the highest magnitude of deformation over the
181 duration of the simulation. The X, Y and Z axis curves of the stress and strain response
182 to the linear and angular acceleration inputs show that the time to peak for loading curve
183 B is shorter than that for the other loading two curve types (figure 2 and 3). Areas of
184 high maximum principal strain were associated with corresponding areas of large Von
185 Mises stress except for linear X and Y loading inputs, where the peak values in white

186 matter and brain stem were close. The peak values for maximum principal strain and
187 Von Mises stress produced by curve B was consistently lower than those for loading
188 curve A. The linear acceleration inputs produced the largest values overall for
189 maximum principal strain and Von Mises stress for the X and Z inputs, with angular
190 acceleration influencing the Y axis responses.

191 ***3.4 Response by location***

192 The different loading curve (A, B and C) results were compared across brain region
193 to analyze the influence of the type (linear or angular) and direction of loading curve
194 input (x, y or z) on the magnitude of deformation in different locations of brain model.
195 The loading curve results were ranked per region of the brain to examine which location
196 incurred the largest deformation for each type of loading curve input. The cerebellum
197 consistently showed the lowest maximum principal strain and Von Mises stress
198 response for all inputs. The X axis linear acceleration loading curves produced the
199 highest values in the white matter and brain stem regions. The X axis angular
200 acceleration response produced high stress and strain response exclusively in the grey
201 matter region. The Y axis linear acceleration produced highest values in the brain stem,
202 with the Y axis angular acceleration producing the highest values in the white matter.
203 The Z axis linear acceleration inputs produced the highest values in the brain stem. The
204 Z axis angular acceleration loading curves produced the largest values of stress and
205 strain in the grey matter.

206 **4. Discussion**

207 The purpose of this study was to investigate the effects of loading curve shape and
208 direction of the loading on brain tissue stress and strain using the University College

209 Dublin Brain Trauma Model (Horgan and Gilchrist, 2004). The research in this area has
210 been limited to animal models that lacked the control that can be achieved using finite
211 element modeling. Recent studies have identified low correlation between linear
212 acceleration and angular acceleration peak values with measured brain injury
213 parameters (Post et al., 2010). This study was undertaken to better understand the
214 relationship between the loading curve shape of linear and angular acceleration and
215 maximum principal strain and Von Mises stress for brain tissue.

216 ***4.1 Location and Direction***

217 The simulations in this study isolated each direction and type of acceleration in an
218 attempt to elucidate differences in location of peak stress and strain. Upon analysis of
219 the results the shape of the loading curve influenced the peak value, time to peak and
220 the location of strain and stress in the brain as defined by the UCDBTM. When loading
221 curve direction is analyzed independently, the regions of highest stress and strain also
222 vary, with the angular acceleration producing high values in either the grey matter for
223 the X and Z axis rotations or white matter for the Y axis rotation. The linear
224 acceleration inputs in the X, Y and Z direction consistently produced the highest values
225 in the brain stem region. From a modelling perspective this difference is likely a result
226 of the interaction of the tissue material characteristics with the different types of loading
227 curve inputs. The fact that the material behaves differently under linear and angular
228 loading curve inputs is an example of this. Different regions of brain tissue would
229 respond differently to these inputs due to the characteristics defining those regions such
230 as material characteristics in compression, shear and tension as well as geometry and
231 boundary conditions. An example of these complex interactions is shown by the Y axis
232 linear acceleration producing high values in the brain stem and Y axis angular
233 acceleration in the white matter. These are fundamentally different motions and as a

234 result the model produces different responses in different brain regions. Also the grey
235 matter showed high strains for the X and Z axis angular acceleration and white matter
236 for the Y axis angular acceleration. From a biological perspective the grey matter would
237 be less sensitive to directional inputs as it is thought to be comprised of non-directional
238 tissues. The white matter is comprised of directional axonal tracts and would be more
239 susceptible to shearing accelerations. These biological differences are represented in the
240 UCDBTM by a difference in shear characteristics; however the model does not consider
241 directionality of tissue in its current format. These differences are likely a reflection of
242 the interaction of the loading curve input characteristics and the complex interactions
243 between the tissue material characteristics and loading conditions within the model.
244 Within the literature, this variation in level of response to direction is supported by
245 Kleiven (2003), who demonstrated a directional dependence of subdural hematoma
246 using finite element modeling approaches, as did Bradshaw and Morfey (2001) for
247 diffuse axonal injury. Using animal studies Gennarelli et al. (1979) and Adams et al.
248 (1981) also demonstrated that direction influences the area and type of injury.

249 ***4.2 Loading curve shape and peak value***

250 The loading curve A condition consistently produced the highest values of
251 maximum principal strain and Von Mises stress for all conditions. Loading curve C
252 produced the lowest values, which were commonly located in the cerebellum. While
253 loading curve A and B had identical peak values; loading curve C had a lower peak
254 which explains the lower response associated with that condition. Loading curve A had
255 a longer time to peak than loading curve B which may explain the resulting large stress
256 and strain values. The maximum principal strain and Von Mises stress output curve
257 shapes for the elements that produced the peak values in the brain have similar profiles
258 to the linear or angular loading curve shapes. Loading curve C output showed a slower

259 rise to peak time without any noticeable plateau producing the lower stress and strain
260 curves. This would suggest a relationship between the loading curve shape and the
261 output curve shape of maximum principal strain and Von Mises stress. The largest
262 values in maximum principal strain and Von Mises Stress were influenced by the linear
263 accelerations and not the rotational accelerations which are in disagreement with
264 research investigating these influences. It is possible that the relative level of linear
265 acceleration (150 g) was higher than that of the rotational acceleration (6,015 rad/s²)
266 producing larger response in maximum principal strain and Von Mises stress. It is
267 difficult to use equivalent linear and rotational acceleration inputs as they produce
268 different motions and mechanisms to the same injury. Also human variation makes it
269 difficult to choose equivalent magnitudes of these parameters as subdural hematoma has
270 been produced at 4,500 rad/s² (Lovenhielm, 1974) and at 130 g (Willinger and
271 Baumgartner, 2003). In hybrid III impacts a value of 140 g can create a rotational
272 component of 12,000 rad/s² (Rousseau and Hoshizaki, 2009; Rousseau et al., 2009),
273 indicating comparing 150 g to 6,015 rad/s² may be an unrealistic scenario in terms of
274 brain tissue response. An ideal solution to these difficulties in analyzing linear and
275 acceleration loading curves would be to have acceleration loading curves that produced
276 a head injury.

277 The high values and location of maximum principal strain in this research were
278 found in the same location as the highest values of Von Mises stress. This would imply
279 that the two are correlated and perhaps it would be redundant to use both in brain tissue
280 deformation. This finding warrants further research. The response shown by this model
281 supports the conclusions of Yoganandan et al.'s (2008) research on biphasic
282 acceleration deceleration pulses using a 2 dimensional finite element model. They
283 reported strains that were produced were region and pulse shape specific.

284 The purpose of this research was to analyze how different loading curve shapes
285 could influence the production of maximum principal strain and Von Mises stress
286 within the University College Dublin Brain Trauma Model. The peaks of these loading
287 curves are commonly used as predictors of injury for protective technologies such as
288 helmets. However it has been indicated by authors that protective technologies must be
289 developed according to brain deformation and not peak linear or angular acceleration
290 (King et al., 2003). This would allow for the interaction between the acceleration
291 loading curves generated at impact and the brain tissue to be accounted for. If a better
292 understanding of these loading curves could be gained, as shown in this research, it
293 could be used to lower the potentially injurious brain deformations incurred from an
294 impact by designing protective devices to limit those types of loading curves and types
295 of acceleration. Future research in this area should focus on the limitations of the
296 modelling and upon attaining further understanding of how loading curves from an
297 impact influence the brain deformations and brain injury metrics which have been
298 suggested to be associated with injuries such as traumatic brain injury and concussion.
299 While this study investigated only a few loading curve characteristics, further studies
300 are warranted using impact loading curves which resulted in an identifiable brain injury
301 to investigate how these parameters influence maximum principal strain.

302 Limitations of this study revolve around the parameters used to define the
303 University College Dublin Brain Trauma Model (UCDBTM). The finite element model
304 (UCDBTM) is an approximation intended to simulate human responses to injurious
305 loading. The material characteristics, constitutive properties assigned to the cerebral
306 elements are taken from cadavers and as such may not represent the true nature of
307 central nervous system tissue. Also the inclusion of cerebral blood vessels in the
308 UCDBTM would likely result in different magnitudes of stress and strain from this

309 analysis (Ho and Kleiven, 2007). In addition this model uses a hyperelastic material
310 model for the brain tissue in shear, however the results would likely change depending
311 on the material law used or if linear elasticity was used instead. The influence of these
312 changes for this particular model has yet to be quantified. The geometry of the model is
313 taken from CT and MRI images from a single subject. The current mesh density used in
314 this model was optimized for validity of results compared to computational time. A
315 change in the density of the mesh used would provide different results, although the
316 exact magnitude of those differences has not been quantified. While validations for
317 finite element models of the human brain in general have been conducted, a complete
318 verification of its predictions for human response has not been accomplished for all
319 parts of the brain. As a result the deformation of the brain represented by the simulation
320 can only be taken as an estimate and not an absolute.

321

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324

325 **Conflict of interest statement:**

326 There are no conflicts of interest for this research

327 **5. References**

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