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Lumped element model of THOR-NT and PMHS abdomen under seatbelt and impactor loading

1

Romain Desbats, François Bermond, Sabine Compigne, Stéphane Nicolle, Philippe Vezin

2

3 **Abstract** Abdominal injuries represent a small proportion of crash injuries but it increases considerably
4 with regard to serious to severe injuries. It is important to assess abdomen injury risk using a biofidelic crash
5 test dummy, especially under seatbelt loading. A lumped element model has been used to simulate seatbelt
6 and impactor loading cases on Post Mortem Human Subjects (PMHS) subject and on the THOR dummy. The
7 model consists of two blocks of one spring and one damper with a mass between the blocks and a mass at the
8 front of the model. The front block has the spring and damper in series, whereas they are in parallel for the rear
9 block, to reproduce the abdomen and back force respectively. The mechanical parameters of the model have
10 been optimised to fit the force-penetration response of PMHS and of the dummy. The results show that the
11 PMHS response is more damper-dependent, whereas the dummy response is more spring-dependent.
12 Moreover, unlike the PMHS, the THOR dummy model needs a non-linear spring in order to fit the impactor test
13 data. The conclusions are that the dummy abdomen should be modified to make it more viscously deformable
14 and to include pelvis design modifications in order to prevent a too stiff response.

15

16 **Keywords** abdomen response, biofidelity, lumped element model, THOR dummy

17

18

I. INTRODUCTION

19 Abdominal injuries represent only a small proportion of all injuries caused by car crashes compared to those
20 of the head, chest or limbs, but their proportion increases considerably with regard to serious to severe injuries
21 (AIS 3 or more): 8% of AIS3+ injuries, 16.5% of AIS4+ injuries and 20.5% of AIS5+ injuries [1]. Besides, it is also
22 shown that rear occupants have a 1.7 times higher risk of suffering an injury to the abdomen than front
23 occupants [2]. Therefore, to provide the best protection to all car occupants, assessing the risk of abdominal
24 injury using a biofidelic crash test dummy is important, especially under seatbelt loading known to be the most
25 common source of AIS3+ injuries to the digestive system [1]-[3].

26

27 Abdomen biofidelity of crash test dummies under seatbelt loading is not included in biofidelity targets. The
28 biofidelity of the THOR dummy proved to be limited under seatbelt and impactor loadings, as shown in [4] and
29 [5] where force-penetration responses of the THOR-NT (reprocessed data from [6]) were compared to PMHS
30 corridors. Two impactor conditions and two seatbelt conditions were considered in [5]. For the impactor cases,
31 the dummy response is above the PMHS corridors after 80 mm whereas for the seatbelt cases, the dummy
32 response is below the PMHS corridors. Finite element models, as well as lumped element models, are tools to
33 help the analysis of human surrogates abdomen response such as PMHS and dummy. The first lumped element
34 model developed to simulate human behaviour under impact is the thorax model from [7]. Reference [8] used
35 this model to simulate an impactor loading case, as well as a seatbelt loading case to the thorax. A simplified
36 version of the same model was used in [9], where the mechanical parameters were optimised to match a given
37 thoracic response. A lumped element model was used to describe the abdomen response under seatbelt tests
38 in [4]. The model consisted of a spring in parallel with a damper and a mass at the front for the dummy case
39 only. The mass value was tuned to fit THOR dummy abdomen behaviour. The PMHS version of this model was
40 enhanced in [10]: a second stage was added to the model, linked with a mass to the first one. This allowed
41 modelling the force between the subject and the back support of the test bench during a seatbelt test.

42

43 The objective of this study is to continue the study from [4] with more loading conditions and to quantify
 44 the biofidelity of the THOR dummy through a comparative analysis of the responses to seatbelt and impactor
 45 loadings of the PMHS and the THOR dummy. A common lumped element model will be used to reproduce the
 46 PMHS and THOR responses by optimising the model's mechanical parameters. The use of a lumped element
 47 model allows to identify mechanical properties more quickly than a FE model, due to a shorter computation
 48 time and the reduced number of parameters. The differences between the PMHS and the THOR responses
 49 could be explained and quantified by the variations between parameters values.
 50

51 II. METHODS

52 *Test data*

53 *Seatbelt*

54 The seatbelt tests selected from the literature and applied to the lumped element model are taken from [8-10]
 55 for the PMHS tests and from [5] (replication of tests from [11]) and [4] for the THOR dummy. These are fixed-
 56 back tests where the belt is placed on the abdomen at the umbilicus level and retracted toward the back of the
 57 subject thanks to pretensioners (or by a piston for one of two condition in [10]). Three conditions from the
 58 literature are common to PMHS subjects and the dummy: B and C conditions from [11] (different
 59 pretensioners) and condition 2 from [4].
 60

61 For the PMHS tests, the average penetration across all the subjects for a given condition was applied to the
 62 model. The average force was compared to the model response. Two tests were performed for each condition
 63 in [5]. The average penetration of those two tests was computed, differentiated to obtain velocity and then
 64 differentiated a second time to determine acceleration. The three signals were filtered in order to avoid noise
 65 that would distort the model computations. Average penetration, velocity and acceleration were applied to the
 66 model and the average force was compared to the model response. One test per condition was performed in
 67 [4]. Fig. 1 shows those input conditions applied to the lumped element model overlaid together. It can be seen
 68 that all the input conditions do not have the same time duration and for some of them the data is not available
 69 over the full time range of the test.
 70

71 *Impactor*

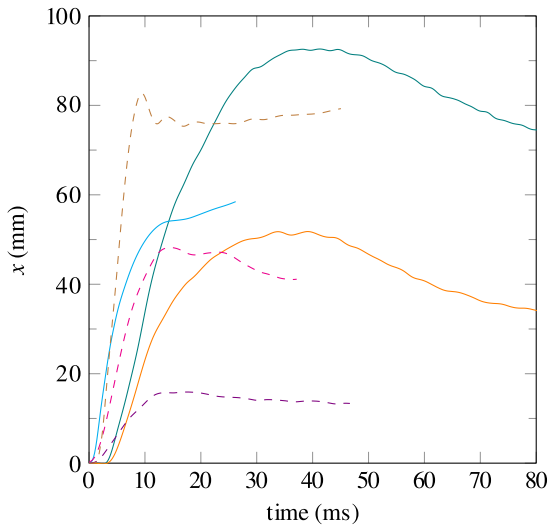
72 Six impactor conditions were applied to the model. For the PMHS subjects, two conditions were taken from
 73 [12] and [13]. The tests with a 64 kg mass from [12] have been excluded. For the THOR dummy two conditions
 74 were taken from [5]. There is only one common configuration for the PMHS and the dummy, the 6 m/s
 75 (nominal velocity) 32 kg test condition. For each configuration, the penetration data from the different tests
 76 were averaged and the corresponding initial velocity was taken as input condition v_0 . Table I shows the
 77 considered input velocities.
 78

79

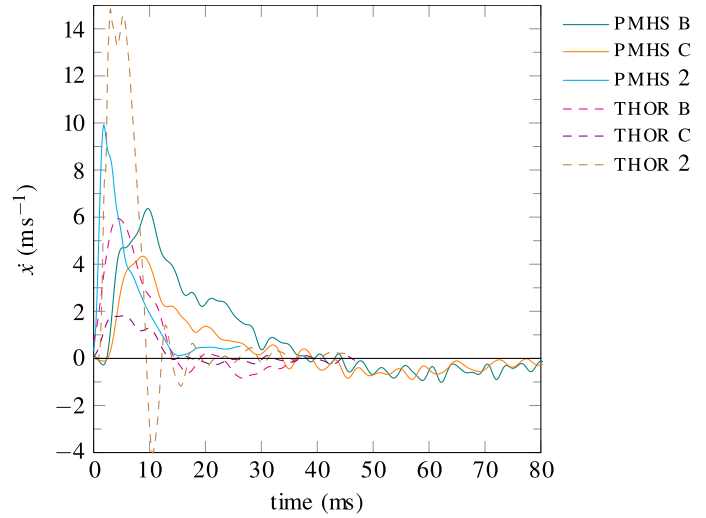
TABLE I
 INITIAL IMPACT VELOCITY VALUES

	condition	v_0 (m/s)	m_i (kg)	kinetic energy (J)
PMHS	Cavanaugh et al. 1986 6.1 m/s	6.1	32	595
	Cavanaugh et al. 1986 10 m/s	10	32	1600
	Hardy et al. 2001 6 m/s	6.5	48	1028
	Hardy et al. 2001 9 m/s	8.5	48	1745
THOR	Compigne et al. 2015 3 m/s	3.0	32	141
	Compigne et al. 2015 6.1 m/s	6.3	32	628

80



(a) Abdomen penetration.



(b) Abdomen penetration velocity.

Fig. 1. Input conditions for seatbelt loading conditions common to the PMHS and the THOR dummy.

B: B condition from Foster et al. 2006

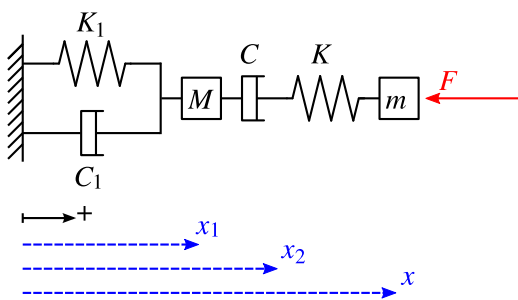
C: C condition from Foster et al. 2006

2: configuration 2 from Trosseille et al. 2002

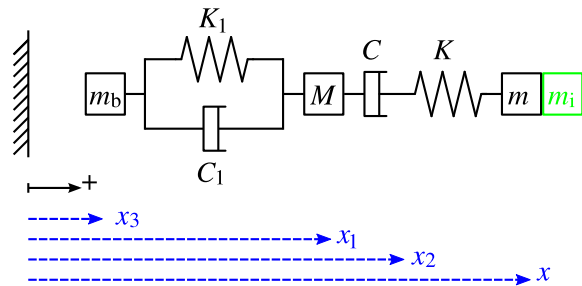
81

82 **Lumped element model**

83 A lumped element model was used to reproduce PMHS and THOR-NT force-penetration responses, in order to
 84 compare the mechanical parameters creating this response. The two stages of the model are meant to
 85 represent the behaviour of the organs, on one hand, and of the flesh and skin, on the other hand. Each stage of
 86 this model consists of a spring associated with a damper in order to represent a visco-elastic behaviour. This
 87 model is based on the work of [10], with some modifications. Linear elements were used instead of
 88 components with a cubic relationship, and a mass has been added at the front of the model to represent the
 89 flesh and skin mass. A mass has also been added at the back of the model to represent the global subject mass.
 90 This mass is fixed to model a seatbelt loading and let free to model an impactor test. An important modification
 91 of the structure of the model was carried out in order to represent both seatbelt and impactor loading cases.
 92 Indeed, the original model from [10] had the springs and dampers in parallel (Kelvin-Voigt model) and therefore
 93 could model a seatbelt test but not an impactor test. Since the extremity of the model has a non-zero initial
 94 velocity, the fact of having a damper in parallel created a non-zero initial reaction force. This is in contradiction
 95 with the test data. Therefore, it was chosen to transform the front stage of the model into a spring in series
 96 with a damper (Maxwell model). Fig. 2 shows the models for seatbelt and impactor loading cases.
 97



(a) Seatbelt.



(b) Impactor.

Fig. 2. Simplified abdomen model for seatbelt and impactor loading cases.

98

99 **Model equations and resolution**

100

101 *Seatbelt*

The variables of the model can be seen on Fig. 2(a). It was chosen to impose the x displacement and its derivatives \dot{x} and \ddot{x} as input conditions. The model response was considered as the force F . They are taken from the test data and imposed on the model. Equations 1(a) and 1(b) are the equations of motion of the system, obtained by isolating the mass M and the point where x_2 is measured (massless node linking the spring and the damper in series), respectively.

$$\begin{aligned} -M \cdot \ddot{x}_1 &= K_1 \cdot x_1 + C_1 \cdot \dot{x}_1 - C \cdot (\dot{x}_2 - \dot{x}_1) & (a) \\ 0 &= -K \cdot (x - x_2) + C \cdot (\dot{x}_2 - \dot{x}_1) & (b) \end{aligned} \quad (1)$$

Equations 1(a) and 1(b) are two coupled differential equations where the unknowns are x_1 and x_2 and where x_2 is used to compute F from Equation 2 (obtained from isolating the mass m) and x_1 is used to compute the force between the back of the subject and the test bench, as shown on Equation 3.

$$F = m \cdot \ddot{x} + K \cdot (x - x_2) \quad (2)$$

$$F_{\text{back}} = K_1 \cdot x_1 + C_1 \cdot \dot{x}_1 \quad (3)$$

Impactor

The model for impactor case is detailed on Fig. 2(b). The initial impactor velocity, $v_0 = \dot{x}(t=0)$, is imposed to the model. Equations 4(a) to 4(d) give the equations of motion of the model. The interaction force between the impactor and the subject is computed according to Equation 5. The differential equations systems are solved with a Runge-Kutta method programmed in the Scilab software.

$$\begin{aligned} -(m + m_i) \cdot \ddot{x} &= K \cdot (x - x_2) & (a) \\ 0 &= -K \cdot (x - x_2) + C \cdot (\dot{x}_2 - \dot{x}_1) & (b) \\ -M \cdot \ddot{x}_1 &= -C \cdot (\dot{x}_2 - \dot{x}_1) + K_1 \cdot (x_1 - x_3) + C_1 \cdot (\dot{x}_1 - \dot{x}_3) & (c) \\ -m_b \cdot \ddot{x}_3 &= -K_1 \cdot (x_1 - x_3) - C_1 \cdot (\dot{x}_1 - \dot{x}_3) & (d) \end{aligned} \quad (4)$$

$$F = (m + m_i) \cdot \ddot{x} \quad (5)$$

Cubic non-linear element as used in [10] proved to be difficult to use in this study: a cubic stiffness has little effect on small compression values but great effect on higher compression values due to the nature of the cubic relation. It was therefore difficult to fit all the test data with such elements. Good non-linear components would include more than one parameter in order to amplify differently low and high compression values. It was therefore decided to use only linear components in the model, if they are sufficient to fit the test data, so as not to add parameters to the model if not necessary. However, it was not possible to fit the THOR dummy impactor test data with linear elements. It was therefore decided to replace the element K with a non-linear spring that shows a strain dependence as seen in Equation 6. The nonlinear stiffness browses successively two regimes of compression and is characterised by three parameters: a linear stiffness K_0 ; a compression limit d ; and a power p (imposed as an integer number). For low compression values, when $(x - x_2) \ll d$, the term between curly brackets in Equation 6 reduces to unity and the response of the spring is linear, $F_{\text{spring}} = K_0 \cdot (x - x_2)$. At moderate to high compression, a stiffening sets in and the restore force becomes higher than the K_0 linear contribution.

$$F_{\text{spring}} = K_0 \cdot (x - x_2) \cdot \left\{ 1 + \left(\frac{|x - x_2|}{d} \right)^p \right\} \quad (6)$$

143

144 **Determination of model parameters**

145

146 *Masses determination*

147 Due to the abdomen representation described above, the masses of the model had to be estimated for the
148 human subjects. It was chosen to use the THUMS model described in [14] to measure those masses. The
149 THUMS model has a representation of the abdomen internal geometry and densities for a 50th percentile male
150 (175 cm height, 77 kg mass) which allows to estimate the masses repartition according to the lumped element
151 model structure. The abdomen was isolated from the rest of the model by drawing a 3D box between L1 and L5
152 lumbar vertebrae. The abdomen was divided in three areas as shown on Fig. 3. The *abdomen content* area is
153 meant to represent the organs that will move during the impact and will correspond to the mass M in the
154 simplified model. The *front flesh* area represents the tissues that create the initial inertia of the abdomen when
155 subjected to seatbelt loading and corresponds to the mass m . The *back* area represents the back of the subject
156 (including the spine) that deforms little during the compression of the abdomen. However, the mass m_b , which
157 is used only in an impactor test condition, would be greater than the mass of the *back* area since more than
158 this body part moves during a free back impact. The calculated masses on the THUMS model (from the selected
159 parts densities) are $M = 5.7$ kg and $m = 0.92$ kg.
160

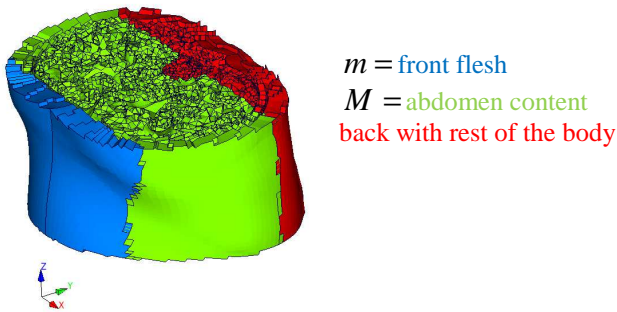


Fig. 3. Masses repartition of the THUMS model.

161

162 Therefore, $M = 6$ kg and $m = 1$ kg were used for all the PMHS subjects. An exception was made for
163 configuration 2 from [4], where m has been set to 0.5 kg due to the low body mass of the PMHS subjects (45
164 kg and 50 kg). Table A.III. (see Appendix) presents the anthropometry of the considered PMHS subjects. The
165 mass m_b was computed as $m_b = BM - M - m$, where BM is the average body mass of the PMHS subjects
166 of the considered test condition. The masses for the THOR dummy were measured on the dummy v1.0 FE
167 model (see [15] for v0.0). The part densities were taken from [16]. The mass of the two foam blocks of the
168 abdomen together is approximately 0.3 kg. Therefore, M was set to 0.2 kg and m was set to 0.1 kg. The mass
169 m should also account for a portion of the dummy jacket, but it is not possible to estimate this contribution.
170 The mass m_b was computed as $m_b = 78.3$ kg $- M - m = 78$ kg, since the total mass of the dummy is equal to
171 78.3 kg according to [17].
172

173

173 *Optimisation loop*

174 The response of the model for a specific loading condition is computed with a Scilab program. Based on
175 manually adjusted initial values, the parameters values are optimised for each test configuration with the Scilab
176 "optim" function until the model response (the force signal) matches the test data, minimising the criteria

177
$$f = \sum_{i=1}^N (F_{\text{model}}(i) - F_{\text{test}}(i))^2$$
 with N the number of data points. The goodness of fit is then estimated by

178 calculating cross-correlation coefficients and the variation range of the parameters is determined.
179

180

180 *Goodness of fit assessment*

181 The assessment of the goodness of fit between the model response and the reference test data is obtained by

182 calculating amplitude ratio, shape factor and phase shift as described in [18]. Equations 7(a) to 7(c) define the
 183 three ratios with $x(t)$ being the signal to compare to the reference $y(t)$ and $\|x\| = \int_{-\infty}^{+\infty} x^2(t) \cdot dt$ the norm of x .
 184 The integrals of digital signals are computed as: $\int_{-\infty}^{+\infty} x(t) \cdot dt = \sum_{i=1}^{N-1} T \cdot \frac{x_{i-1} + x_i}{2}$ with T the sampling rate
 185 (numerical integration by trapezoidal rule). A perfect fit between the two signals would lead to an amplitude
 186 ratio and a shape factor of 1 and a phase shift of 0.
 187

$$\text{amplitude ratio} = \frac{\|x\|}{\|y\|} \quad (\text{a})$$

$$\text{shape factor} = \frac{\int_{-\infty}^{+\infty} x(t) \cdot y(t) \cdot dt}{\sqrt{\|x\| \cdot \|y\|}} \quad (\text{b}) \quad (7)$$

$$\text{phase shift} = h \text{ so that } \frac{\int_{-\infty}^{+\infty} x(t) \cdot y(t+h) \cdot dt}{\sqrt{\|x\| \cdot \|y\|}} \text{ is minimal} \quad (\text{c})$$

189

190 *Sensitivity analysis*

191 In order to compare the model parameters for different conditions, it was necessary to estimate the range in
 192 which the optimised parameters can vary without significantly changing the model response result. The
 193 optimised values of the model parameters (spring stiffnesses and damping parameters) and the masses values
 194 were subjected to variations by steps of 10% of their value, ranging from 0.1 to 2 times the initial value. For
 195 each parameter value the three correlation coefficients mentioned above between the model response and the
 196 test data were computed. The percentage of variation of the three coefficients was computed at each step. A
 197 variation of one of the coefficients exceeding 10% of the coefficient value computed with the optimised
 198 parameter was considered as the limit of the range of variation of the parameter. An exception was made for
 199 the phase shift coefficient. In order to avoid the artefacts due to small phase values, the value of phase shift
 200 used in these computations was divided by 10 ms and the coefficient $1 - \frac{\text{phase shift}}{10 \text{ ms}}$ was considered instead
 201 of the phase shift in ms, in order to have a coefficient equal to 1 in case of a perfect fit.
 202

203

III. RESULTS

204 *Model fit to test data*

205 Figures 4 and 5 show the model response fit to the test data under seatbelt and impactor loading respectively,
 206 for the common conditions between PMHS subjects and the dummy. This model response is obtained with the
 207 optimised mechanical parameters values reported in Tables A.I. and A.II., respectively (see Appendix). Non-
 208 normalised data have been used for the PMHS penetration and force response. The standard deviation
 209 corridors are indicated for the PMHS force data. The goodness of fit coefficients are reported at the bottom of
 210 each subfigure. Force-penetration graphs of those results are presented in Fig. A.1. (see Appendix).
 211

212 For the seatbelt conditions, the initial negative slope of the model force response for some tests is an
 213 artefact due to the negative slope in the \dot{x} data (see Fig. 1(b)). This may be a consequence of the filtering of
 214 penetration data to compute the input velocity. The THOR dummy response shows sharp peaks and an abrupt
 215 initial slope that the model can-not fit exactly, regardless of the potential time offset in the force signal from
 216 test data. For the impactor conditions, The non-linearity of the dummy response can be seen on Fig. 5(c) and
 217 (d), where the slope of the force-time signal changes around 10 ms, and on Fig. A.1. (c) and (d). Comparing
 218 Figures A.1. (c) and (d) for the PMHS data, it appears that the model predicts a higher penetration than
 219 measured in the test data.

220

221 For both seatbelt and impactor case the model allows to fit the test data, the response being mainly in the
222 standard deviation corridors for the PMHS, the amplitude and shape ratios stay between 0.8 and 1.1 and the
223 phase shift between ± 1.2 ms. This means that this model is suitable to analyse the human and dummy
224 abdomen response.

225

226 *Parameters values*

227 Figure 6 shows the parameters identified for each common PMHS/dummy condition along with their range of
228 variation for the seatbelt and impactor cases, respectively. The numerical values are reported in Table A.I. and
229 Table A.II. (see Appendix).

230

231 For the seatbelt case, the first thing to be noticed on Fig. 6(a) is that the K values for the PMHS subjects are
232 much higher than for the THOR dummy. The C values are higher for the dummy than for the PMHS subjects
233 (Fig. 6(c)), except for configuration 2 from [4] where they are equal. The C_1 values (Fig. 6(g)) are higher for the
234 dummy compared to the PMHS, except for the configuration 2 from [4], but since ranges of variation overlap
235 for most of the test conditions, it is concluded that the C_1 values are the same for the PMHS and the dummy.
236 Additionally, the K_1 values presented on Fig. 6(e) are similar between the PMHS and the dummy, but large
237 ranges of variation exist for all of the conditions. This high range of variation of the parameter K_1 is due to the
238 fact that x_1 is significantly lower than x for most of the conditions. Therefore, a given variation of K_1 would
239 have less effect on the abdomen force than the same variation of K . Regarding the results for configuration 2
240 from [4], the fact that C is equal for the PMHS and the dummy and that C_1 is lower for the dummy, can be
241 explained by the fact that the penetration velocity is the highest for this condition (Fig. 1(b)).

242

243 For the impactor case, since a non-linear spring was needed to fit the THOR test conditions, the effective
244 stiffness of these conditions is compared to the K values of the other conditions on Fig. 6(b). The effective

245 stiffness for the non-linear spring is computed as $K_{eff} = \frac{F_{spring}}{x - x_2}$ with F_{spring} taken from Equation 6, the force

246

given by the spring. K_{eff} is plotted against the relevant elongation, $x - x_2$. For a linear spring (PMHS case),

247

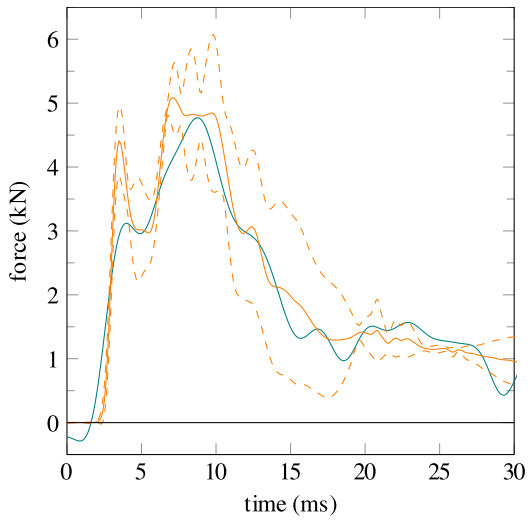
$K_{eff} = K$. For clarity, only optimised values are plotted, without range of variation. The K_{eff} values for the
248 THOR dummy are lower than the K values of the PMHS subjects on most of the penetration range. The C
249 values are higher for the THOR dummy compared to the PMHS subjects (Fig. 6(d)) except for 10 m/s condition
250 where the PMHS have a higher C value than the other conditions. The same trend is noted for the C_1
251 parameter (Fig. 6(h)), although there is a large range of variation for the 10 m/s condition. The K_1 values
252 presented on Fig. 6(f) seem higher for the dummy compared to the PMHS, but due to relatively large ranges of
253 variation of this parameter it can take the same values for all the conditions. The same observation can be
254 made for the high range of variation of K_1 than in the seatbelt case. The fact that the 10 m/s condition has a
255 higher C value compared to the other PMHS condition can be due to the fact that this is the highest velocity of
256 the test conditions.

257

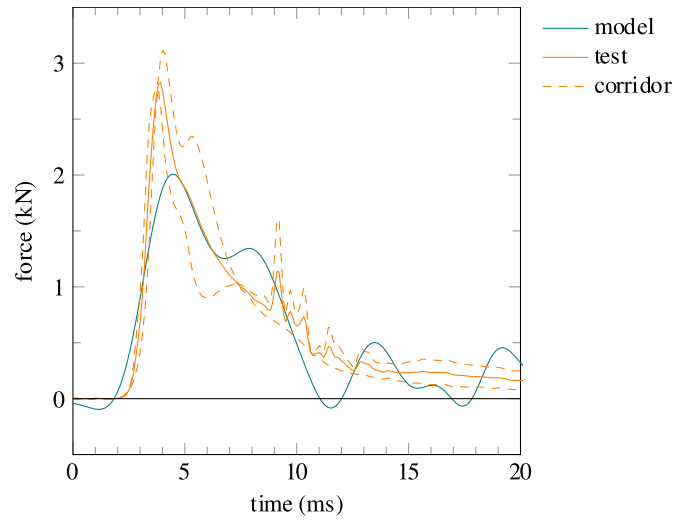
258 It stands out that the second stage of the model contributes less to the global penetration response of the
259 model since the displacement and velocity of the second stage, (x_1 and \dot{x}_1 for seatbelt, $x_1 - x_3$ and $\dot{x}_1 - \dot{x}_3$
260 for impactor), are lower than those of the front stage ($x - x_1$ and $\dot{x} - \dot{x}_1$). This leads to focussing on the K
261 and C parameters for the analysis.

262

263 For the conditions that are not common to the PMHS and the dummy, the optimised parameters are presented
264 in Fig. A.2. (see Appendix). The same trends and order of magnitudes as explained above can be noticed.

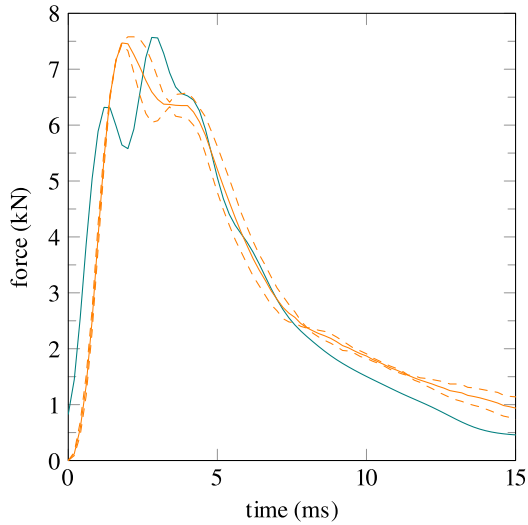


(a) PMHS B.



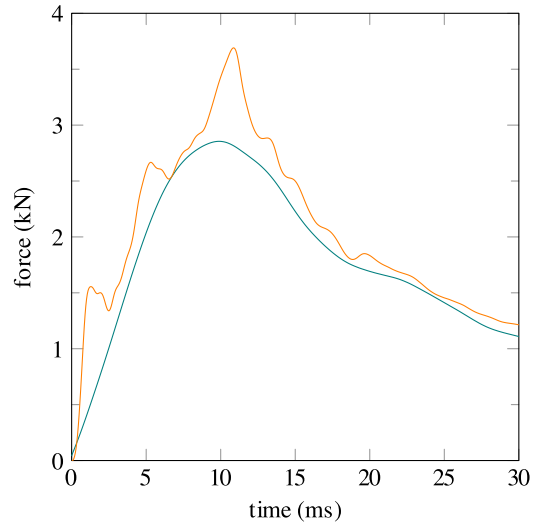
(b) PMHS C.

amplitude=0.86 shape=0.99 phase=0.05 ms



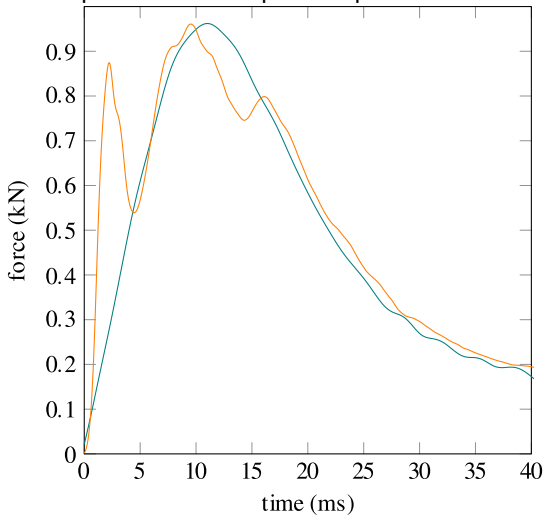
(c) PMHS 2.

amplitude=0.87 shape=0.95 phase=-0.1 ms



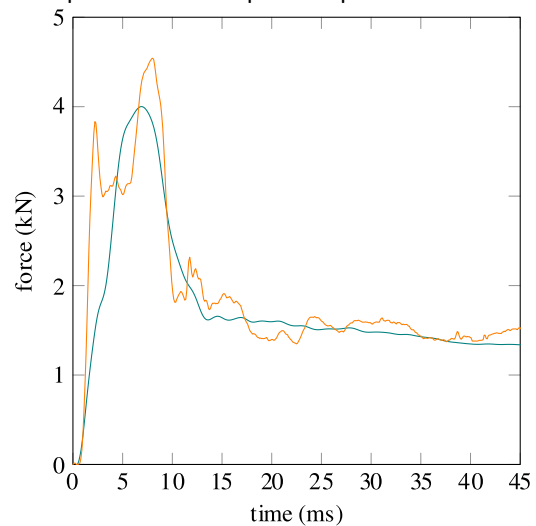
(d) THOR B.

amplitude=1.01 shape=0.98 phase=0.3 ms



(e) THOR C.

amplitude=0.80 shape=0.99 phase=-0.05 ms



(f) THOR 2.

amplitude=0.93 shape=0.97 phase=-1.2 ms

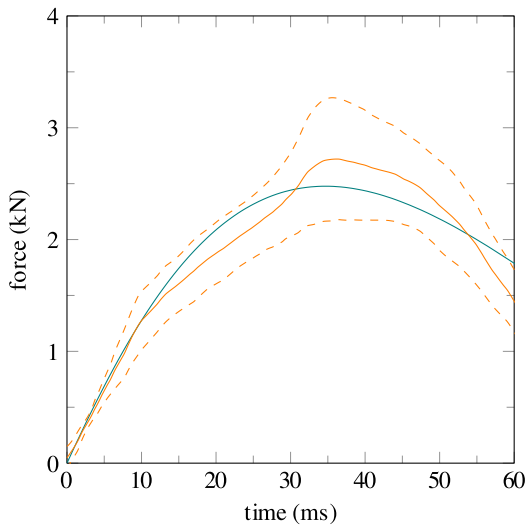
amplitude=0.86 shape=0.97 phase=-0.1 ms

Fig. 4. Fit of the model response to test data for seatbelt case.

B: B condition from Foster et al. 2006

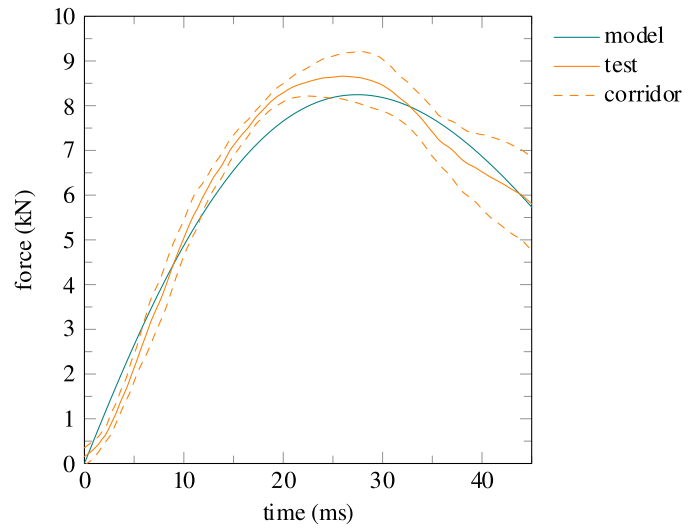
C: C condition from Foster et al. 2006

2: configuration 2 from Trosseille et al. 2002



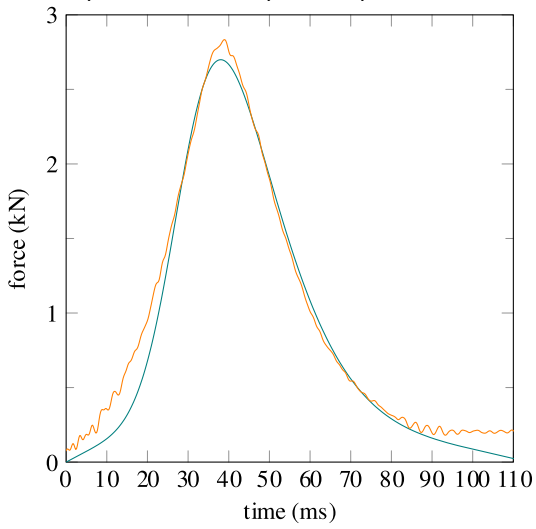
(a) PMHS 6.1 m/s.

amplitude=0.95 shape=1.00 phase=0 ms



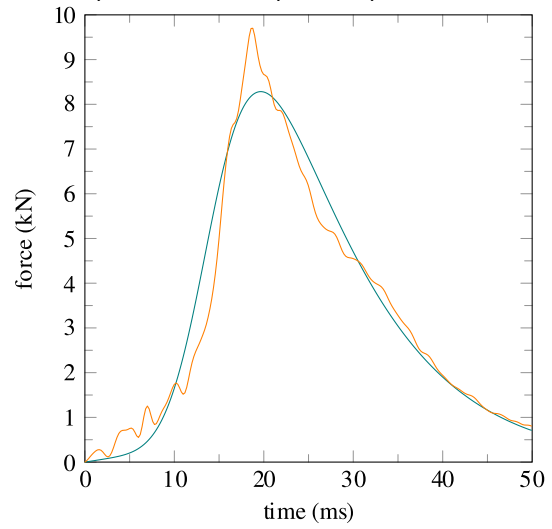
(b) PMHS 10 m/s.

amplitude=1.00 shape=1.00 phase=0 ms



(c) THOR 3 m/s.

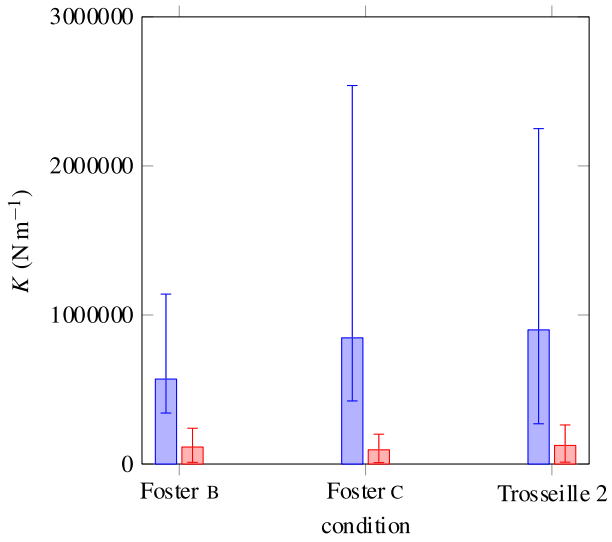
amplitude=0.94 shape=0.99 phase=-1 ms



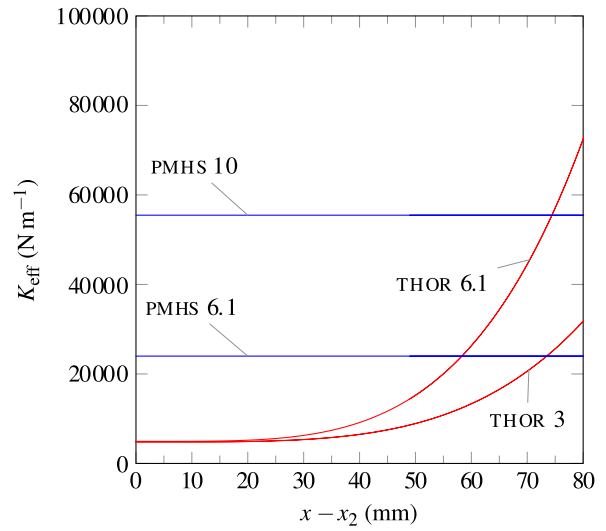
(d) THOR 6.1 m/s.

amplitude=1.04 shape=0.99 phase=0.4 ms

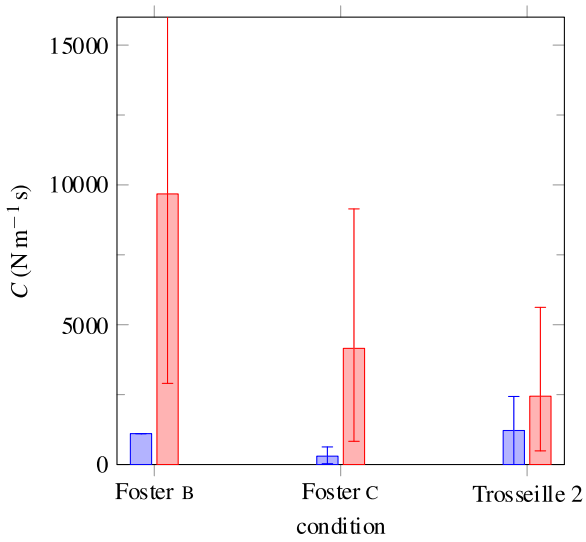
Fig. 5. Fit of the model response to test data for impactor case.
 PMHS 6.1 m/s: 32 kg 6.1 m/s condition from Cavanaugh et al. 1986
 PMHS 10 m/s: 32 kg 10 m/s condition from Cavanaugh et al. 1986
 THOR 3 m/s: 32 kg 3 m/s condition from Compigne et al. 2015
 THOR 6.1 m/s: 32 kg 6.1 m/s condition from Compigne et al. 2015



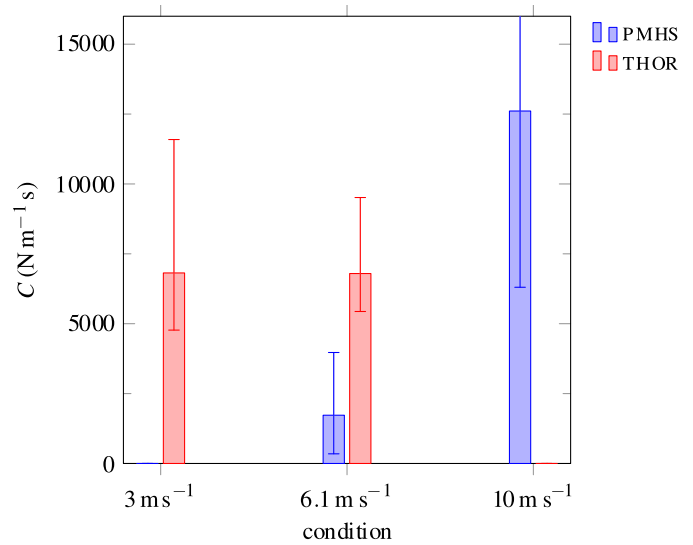
(a) K for seatbelt.



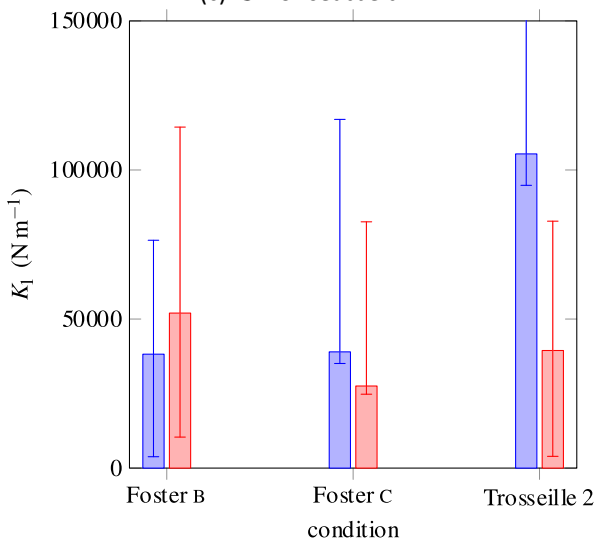
(b) K_{eff} for impactor.



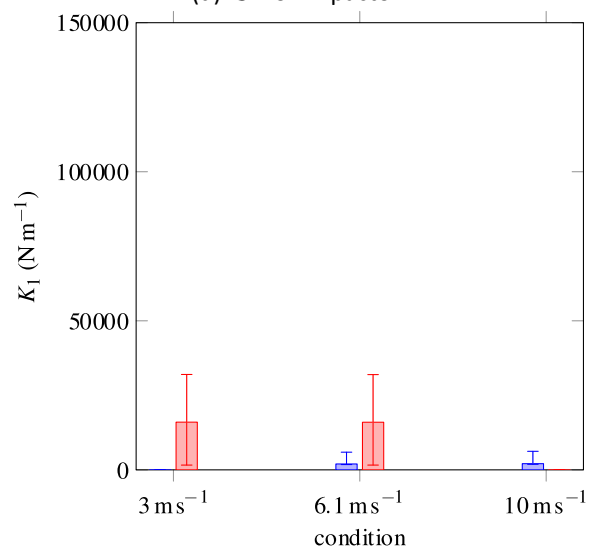
(c) C for seatbelt.



(d) C for impactor.



(e) K_1 for seatbelt.



(f) K_1 for impactor.

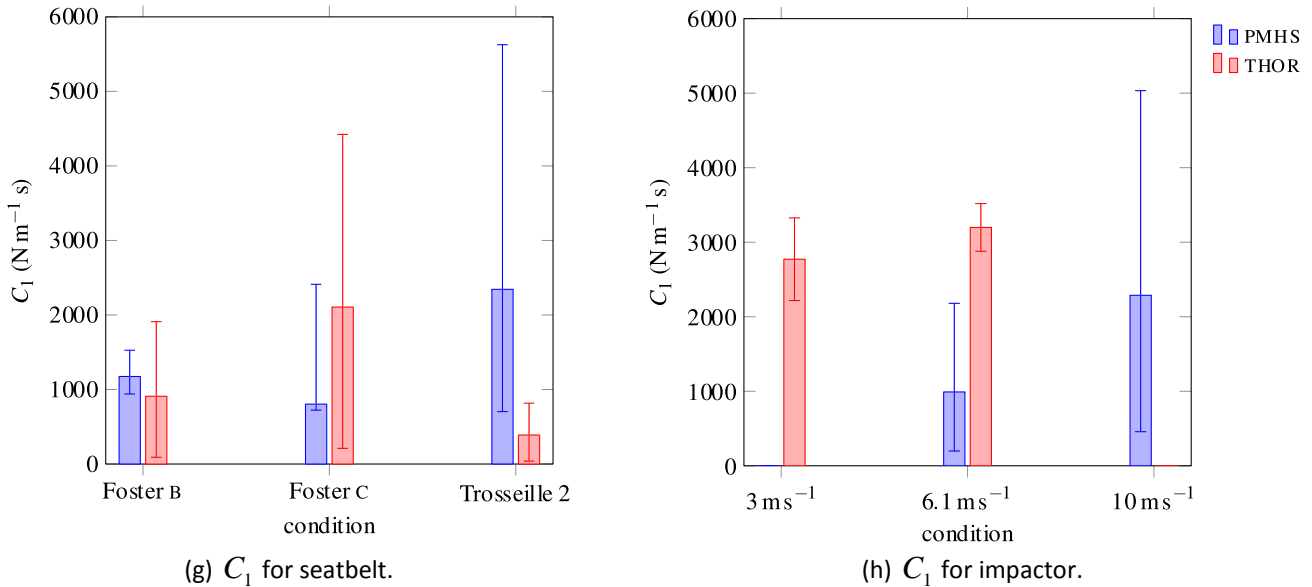


Fig. 6. Identified model parameters for seatbelt and impactor common loading conditions to PMHS and THOR.

B: B condition from Foster et al. 2006

C: C condition from Foster et al. 2006

2: configuration 2 from Trosseille et al. 2002

PMHS 6.1 m/s: 32 kg 6.1 m/s condition from Cavanaugh et al. 1986

PMHS 10 m/s: 32 kg 10 m/s condition from Cavanaugh et al. 1986

THOR 3 m/s: 32 kg 3 m/s condition from Compigne et al. 2015

THOR 6.1 m/s: 32 kg 6.1 m/s condition from Compigne et al. 2015

267

268

IV. DISCUSSION

269 Interpretation of the results

270 The parameters values between seatbelt and impactor cases can be compared on Fig. 6. The K values for the
 271 PMHS subjects (Fig. 6(a) and (b)) are very different between the seatbelt and impactor cases. Regarding the
 272 THOR dummy, the K values for the seatbelt case are the same order of magnitude as the effective stiffness
 273 values of the impactor case. The origin value of the THOR curves on Fig. 6(b) is the K_0 value. The linear
 274 stiffness of the dummy modelled with a non-linear spring is lower than the stiffness of the other conditions.
 275 The C values (Fig. 6(c) and (d)) match well between the seatbelt and impactor cases (except for the 10 m/s
 276 condition). K_1 values match between seatbelt and impactor cases for the THOR dummy, whereas values are
 277 much lower in the impactor case for the PMHS (Fig. 6(e) and (f)). However, the large range of variation (relative
 278 to the optimised value) for the PMHS in the impactor case allows K_1 to take any value to fit the data. C_1
 279 values are similar for PMHS and the THOR dummy (Fig. 6(g) and (h)) between the seatbelt and impactor cases
 280 and some large ranges of variation exists.

281

282 In order to understand the meaning of the parameters value comparison, one has to note that the influence
 283 of the parameters is not the same whether they are placed in series or in parallel in the model. For instance, a
 284 spring in parallel will have no effect when its stiffness is equal to 0, whereas a spring in series has no effect
 285 when its value tends toward infinity. It is the same for damping parameters. This explains the important range
 286 of variation toward high values for K and C concerning PMHS and THOR respectively. A high value for such a
 287 component in series means it deforms little. Therefore a higher value means it deforms even less, which does
 288 not necessarily change the global result.

289

290 The model in series has a characteristic time $\tau = \frac{C}{K}$. The model behaviour is elastic for times inferior to τ
 291 and viscous for times superior to τ . The characteristic times have been computed with the optimised K and

292 C values. It can be concluded that the penetration response of the abdomen is mainly caused by the
293 instantaneous deformation of the spring K for the THOR dummy and by the long-term deformation of the
294 damper C for the PMHS. For the seatbelt case, this is highlighted by Figures A.3. (a) and A.3. (b) (see
295 Appendix), which show the prominence of the K spring and C damper responses on the abdominal
296 penetration of the dummy and the PMHS, respectively. The PMHS characteristic time being low (less than 3 ms)
297 compared to the loading duration, the PMHS response is mainly viscous. On the opposite, the dummy
298 characteristic times are between 20 ms and 80 ms, higher than the loading duration, therefore the response is
299 mainly elastic. Figures A.4. (a) and A.4. (b) (see appendix) show the same phenomenon for the impactor case. It
300 is difficult, however, to compare the characteristic times with a non linear spring in the dummy case.
301

302 These findings correlate with the test data reported in [5]. For the impactor case the dummy force response
303 was higher than the PMHS responses and for the seatbelt case the dummy force response was lower than the
304 PMHS responses. This is explained by the previous findings. An impactor loading creates high abdomen
305 penetration, therefore the dummy force response is higher than the PMHS one due to the higher influence of
306 K (the spring non-linearity enhances this phenomenon since high penetration increases the spring force
307 contribution). On the other hand, a seatbelt loading creates less penetration (compared to the impactor
308 loading) and a sharp velocity peak, therefore the dummy force response is lower than the PMHS one due to the
309 lower influence of C .
310

311 The fact that the K values are extremely different between the dummy and the PMHS subjects only for the
312 seatbelt case is also explained by the PMHS mainly viscous behaviour for the seatbelt condition, whereas for
313 the impactor condition the spring and damper contribution are more balanced. This may be due to the fact that
314 for the seatbelt condition, the displacement and velocity of the extremity of the model is imposed. The force
315 profile to fit having a similar shape than the velocity profile explains that the K value has less influence. The
316 results would perhaps be different if the force were imposed. In the impactor case, only the initial velocity of
317 the model's extremity is imposed, which imposes less constraints on the model. This also explains why the
318 PMHS K values are higher in the seatbelt case as compared to the impactor case.
319

320 This spring-driven behaviour of the dummy and the damper-driven behaviour of the PMHS highlighted by
321 the model are due to the nature of the constitutive material of the human subjects and the dummy. The
322 human abdomen is made of soft tissues that include a large proportion of water, therefore causing a viscous
323 behaviour whereas the dummy abdomen is made of foam and therefore has an elastic behaviour, although not
324 linear and strain rate dependent. The influence of the dummy spine and pelvis (more important than for the
325 human) explains the need for a non-linear spring in the impactor case. The optimised values of the d
326 parameter of the non-linear spring (42 mm and 52 mm) represent the level of compression after which the
327 non-linearity appears. This corresponds to the depth the abdomen foam has to be compressed before the
328 pelvis is involved (measured at approximately 45 mm on the dummy).
329

330 *Comparison with other studies*

331 References [4] and [10] also developed lumped element models of the abdomen for seatbelt loading. It should
332 be noted that in [4], the penetration of the seatbelt into the abdomen was estimated as the average of the left
333 and right belt strand displacement, although in [11] and [10] the belt penetration in the abdomen was
334 measured directly.
335

336 The stiffness parameters found in [4] were from 12000 N/m to 17000 N/m (except for the subject PRT035
337 showing a very low stiffness) and the damping parameters from 600 N/m.s to 900 N/m.s. The lumped element
338 model used for the THOR dummy in [4] had an extra mass compared to the model used for PMHS and the
339 spring was non-linear. It is difficult to compare those values with the K and C values of this study since the
340 first stage of the present model, where the displacement is imposed, consists of two components in series and
341 not in parallel as in [4]. It was concluded in [4] that the THOR abdomen was too stiff (approximately two times)
342 and not viscous enough (approximately four times) compared to the human subjects. The same conclusion can

343 be drawn in the present study for K and C , with the distinction that the components are in series in the
344 present model, therefore it is more correct to say that the abdomen response is more influenced by the spring
345 and less by the damper for the dummy compared to the PMHS. It was chosen to compare the parameters of
346 the THOR dummy and the human subjects based on the same lumped element model until it was necessary to
347 use a non-linear spring for the dummy under impactor loading.

348
349 Reference [10] applied a lumped element model to PMHS response under seatbelt loading. But as the front
350 stage of the model had K and C in parallel it is difficult to compare their values with those from the present
351 model. Non-linear elements were used for K , K_1 and C_1 , which also makes it difficult to compare them with
352 the values from this study. Moreover, in [10] the force from test data was applied to the model and the
353 penetration given by the model was compared with the test data. This is a different approach from the one
354 used in this study, where the penetration was applied to the model, as in [4] and [8], which is believed to be
355 more representative of a seatbelt loading with a piston or pretensioners. The two different methods could
356 potentially lead to different results with the same parameters, especially in cases where the elements of the
357 model are non-linear.

358
359 There is also a difference in the manner of determining the value of the parameters for given reference
360 data. In [4] and [10] the parameters were identified from the test data as sloped in a force-penetration diagram
361 for a stiffness parameter, for example. In this study an optimisation procedure was used, as in [7] and [9]. A
362 sensitivity analysis of the optimised parameters was also performed to predict the effect of the variation of the
363 parameters on the response of the model.

364 *Limitations*

365
366 Lumped element models applied to impact biomechanics have inherent limitations. They only represent a
367 one-dimensional loading and use simple mechanical parameters. A linear model proved to be sufficient to fit
368 the PMHS and dummy response until non linearity appeared for dummy impactor tests. The structure chosen
369 for the first stage of the present model (Maxwell model) theoretically represents a fluid behaviour. If loaded
370 with an imposed constant force (like at the end of a seatbelt loading, for instance), it would result in an
371 infinitely increasing penetration. A generalised Maxwell model with three elements (standard linear solid)
372 would account for those limitations.

373
374 As described above, the second stage of the model contributes less to the global penetration response of
375 the model. This causes the K_1 and C_1 parameters to have less influence on the response and more variability
376 than the K and C parameters. This limits the relevance of a two stages model.

377
378 The fact that for the PMHS subjects in the impactor case the model penetration is higher than the
379 penetration from test data raises a limitation of the model. The penetration is not imposed on the model,
380 unlike for the seatbelt case, and since the model parameters are optimised to fit the force from test data only,
381 the penetration response of the model does not necessarily fit the test data. This could be due to the fact that
382 the model does not take into account the abdomen depth, therefore allowing the predicted penetration to be
383 higher than the test penetration. This could also be due to the Maxwell model used for the first stage of the
384 model, where the C damping element does not unload. This can be seen on Fig. A.4. (b) where the
385 compression of C does not decrease during unloading. It would also be useful to have abdomen depth
386 considered in the lumped element model to account for the individual anthropometry variations between
387 PMHS subjects.

388 *Guidelines for improving the THOR dummy*

389
390 Although there are differences in the model parameters values between the seatbelt and impactor case, the
391 results show that the THOR dummy abdomen is more elastically and less viscously deformable compared to the
392 human abdomen. This behaviour is mainly explained by the first stage of the model, i.e. related to parameters
393 K and C since the second stage contributes less to the response as explained above.

394

395 Based on these observations, some improvements can be applied to the dummy at the material and
396 structural levels. The material properties of the lower abdomen foam should be modified to favour the viscous
397 behaviour instead of the elastic behaviour. The need of a non-linear spring for the dummy model at high
398 penetrations, whilst it was not necessary for the PMHS, means that stiffer parts (such as the pelvis) contribute
399 non-biofidelically more to the response than soft parts. This confirms the design changes of this region
400 implemented on THOR Mod Kit, which shortened the pelvis flesh at antero-superior iliac spines by around 20
401 mm. Furthermore, the difference in moving mass between the dummy abdomen and the PMHS has an
402 influence and therefore the dummy abdomen mass should be increased.

403

404

V. CONCLUSION

405 A lumped element model has been designed in order to reproduce the abdomen force-penetration
406 response of human subjects and the THOR dummy under seatbelt and impactor loading. This model proved to
407 be capable of fitting the different test data and allows an explanation of the differences between the abdomen
408 response of human subjects and a crash test dummy in terms of mechanical parameters. The viscous
409 contribution to the dummy abdomen deformation has to be increased relative to the elastic contribution in
410 order to improve its biofidelity. To achieve this goal, the material and structural modifications of THOR
411 abdomen proposed in this study will be tested on THOR Metric FE model released by NHTSA in 2015 (v2.1). The
412 ultimate goal will be to propose a new abdomen prototype.

413

414

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416 Trosseille et al. 2002 and from Lamielle et al. 2008 as well as Mr Craig Foster for providing the data from Foster
417 et al. 2006.

418

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VIII. APPENDICES

TABLE A.I.

IDENTIFIED PARAMETERS FOR SEATBELT LOADING CONDITIONS, INCLUDING SENSITIVITY ANALYSIS
(PARAMETERS VALUE RANGE WHICH KEEPS GOODNESS OF FIT PARAMETERS WITHIN $\pm 10\%$ RANGE)

		Trosseille et al. 2002 condition 2					
		K (N/m)	C (N/m.s)	K_1 (N/m)	C_1 (N/m.s)	M (kg)	m (kg)
superior limit		1350054	1217	210787	3282	10.2	1
optimised value		900036	1217	105393	2344	6	0.5
inferior limit		630025	1217	10539	1641	1.8	0.05
		Foster et al. 2006 B condition					
		K	C	K_1	C_1	M	m
PMHS	superior limit	1140043	1107	76428	1527	9.6	1.6
	optimised value	570021	1107	38214	1175	6	1
	inferior limit	342013	1107	3821	940	3	0.3
		Foster et al. 2006 C condition					
		K	C	K_1	C_1	M	m
superior limit		1692911	330	77983	1608	12	1.1
optimised value		846456	300	38992	804	6	1
inferior limit		423228	270	3899	80	1.2	0.9

Trosseille et al. 2002 condition 2						
	K	C	K_1	C_1	M	m
superior limit	137326	3178	43387	428	0.4	0.2
optimised value	124841	2445	39443	389	0.2	0.1
inferior limit	112357	1956	35499	350	0.02	0.01
Foster et al. 2006 B condition						
	K	C	K_1	C_1	M	m
superior limit	125792	19356	62396	1000	0.4	0.2
optimised value	114357	9678	51997	910	0.2	0.1
inferior limit	102921	6774	41597	819	0.02	0.01
Foster et al. 2006 C condition						
	K	C	K_1	C_1	M	m
superior limit	104873	4986	55076	2317	0.4	0.2
optimised value	95339	4155	27538	2106	0.2	0.1
inferior limit	85805	3324	2754	1896	0.02	0.01

THOR

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TABLE A.II.
IDENTIFIED PARAMETERS FOR IMPACTOR LOADING CONDITIONS, INCLUDING SENSITIVITY ANALYSIS
(PARAMETERS VALUE RANGE WHICH KEEPS GOODNESS OF FIT PARAMETERS WITHIN $\pm 10\%$ RANGE)

Cavanaugh et al. 1986 32 kg 6.1 m/s condition									
	K	C	K_1	C_1	M	m	m_b		
	(N/m)	(N/m.s)	(N/m)	(N/m.s)	(kg)	(kg)	(kg)		
superior limit	31179	2244	3949	1189	9	2	110		
optimised value	23984	1726	1974	991	6	1	55		
inferior limit	21585	1381	197	793	3	0.1	33		
Cavanaugh et al. 1986 32 kg 10 m/s condition									
	K	C	K_1	C_1	M	m	m_b		
superior limit	66570	25217	4149	2746	10.8	2	154		
optimised value	55475	12609	2075	2288	6	1	77		
inferior limit	49928	6304	207	1830	1.2	0.1	53.9		
Hardy et al. 2001 48 kg 6 m/s condition									
	K	C	K_1	C_1	M	m	m_b		
superior limit	37220	1384	81092	5847	12	2	80.4		
optimised value	37220	1259	40546	2923	6	1	67		
inferior limit	37220	1133	4055	1462	0.6	0.1	60.3		
Hardy et al. 2001 48 kg 9 m/s condition									
	K	C	K_1	C_1	M	m	m_b		
superior limit	73832	3338	20284	1873	9.6	2	74.1		
optimised value	67120	2781	10142	1561	6	1	57		
inferior limit	60408	2225	1014	1249	1.8	0.1	45.6		
Compigne et al. 2015 32 kg 3 m/s condition									
	K_0	d	P	C	K_1	C_1	M	m	m_b
	(N/m)	(mm)	(no unit)	(N/m.s)	(N/m)	(N/m.s)	(kg)	(kg)	(kg)
superior limit	5284	52	4	11588	32005	3326	0.4	0.2	101.4
optimised value	4804	52	4	6817	16002	2772	0.2	0.1	78
inferior limit	4323	52	4	4772	1600	2217	0.02	0.01	62.4
Compigne et al. 2015 32 kg 6.1 m/s condition									
	K_0	d	P	C	K_1	C_1	M	m	m_b
superior limit	6911	42	4	9514	31955	3519	0.4	0.2	109.2
optimised value	4936	42	4	6796	15977	3199	0.2	0.1	78
inferior limit	3949	42	4	5437	1598	2879	0.02	0.01	62.4

PMHS

THOR

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TABLE A.III.
ANTHROPOMETRY OF CONSIDERED PMHS SUBJECTS

Trosseille et al. 2002							
	configuration	test	gender	age (years)	body mass (kg)	stature (cm)	abdomen depth (mm)
	2	PRT038	F	64	45	149	206
		PRT039	F	86	50	150	207
Foster et al. 2006							
seatbelt	configuration	test	gender	age (years)	body mass (kg)	stature (cm)	abdomen depth (mm)
	B	B-1	M	85	81	169	360
		B-2	M	45	75	174	258
		B-3	M	59	62	169	261
	C	C-1	F	86	38	159	191
		C-2	F	86	38	159	191
Cavanaugh et al. 1986							
	configuration	test	gender	age (years)	body mass (kg)	stature (cm)	abdomen depth (mm)
impactor	low velocity	14	M	56	68	182	283
		19	F	43	53	159	231
		24	M	57	45	187	257
		28	F	57	75	163	275
		33	F	51	68	163	261
		37	M	50	88	169	332
	high velocity	57	M	64	90	184	322
		61	M	60	79	180	277

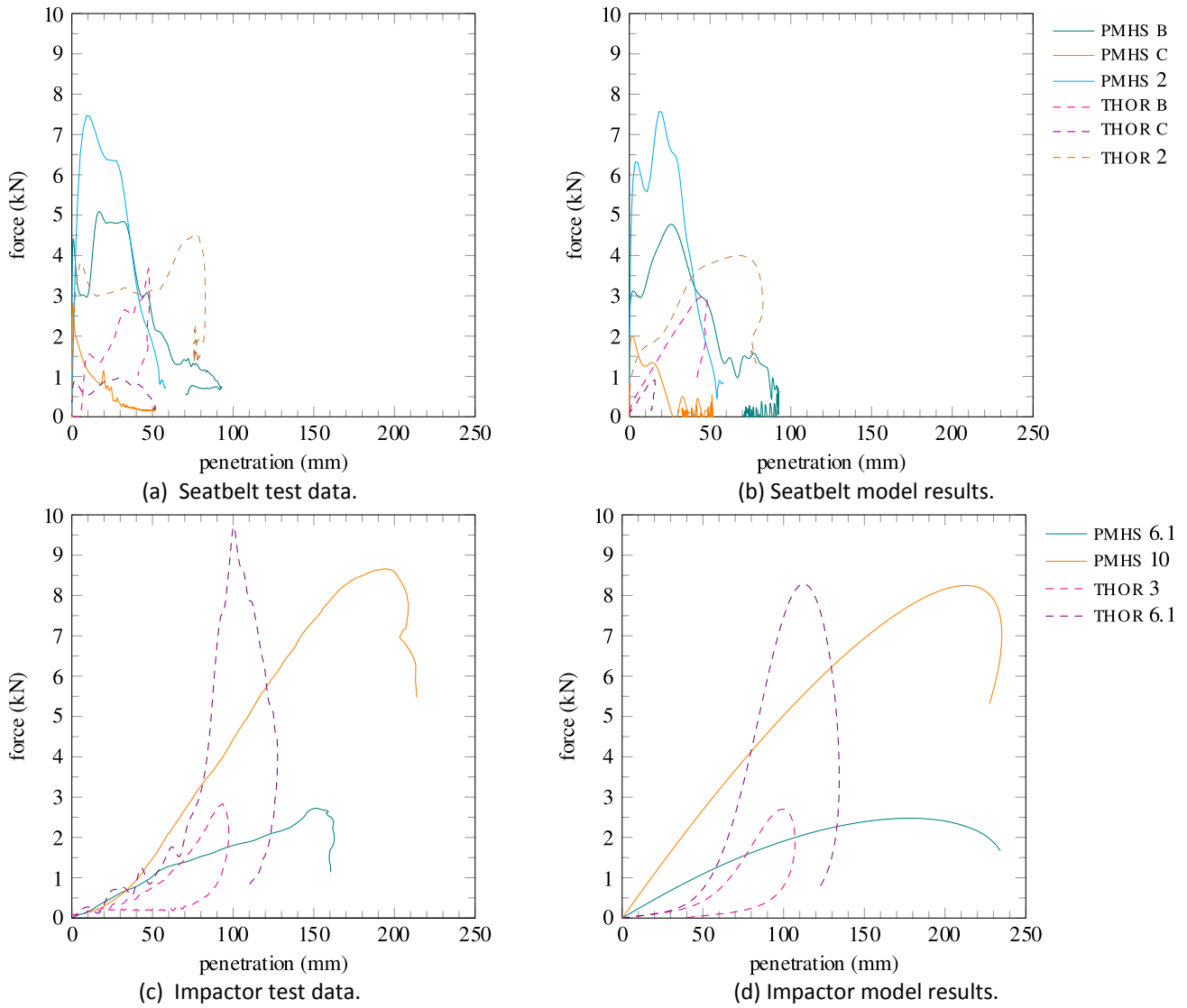


Fig. A.1. Force-penetration responses.

B: B condition from Foster et al. 2006

C: C condition from Foster et al. 2006

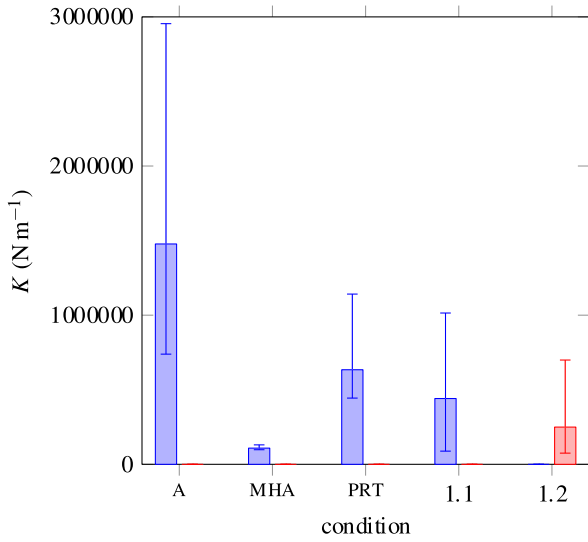
2: configuration 2 from Trosseille et al. 2002

PMHS 6.1 m/s: 32 kg 6.1 m/s condition from Cavanaugh et al. 1986

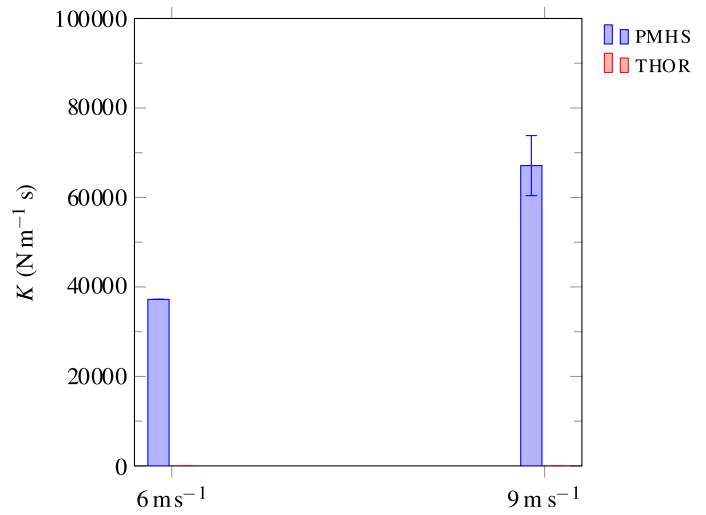
PMHS 10 m/s: 32 kg 10 m/s condition from Cavanaugh et al. 1986

THOR 3 m/s: 32 kg 3 m/s condition from Compigne et al. 2015

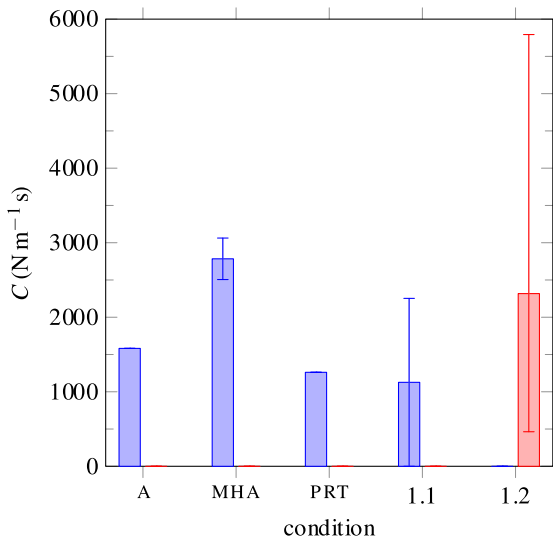
THOR 6.1 m/s: 32 kg 6.1 m/s condition from Compigne et al. 2015



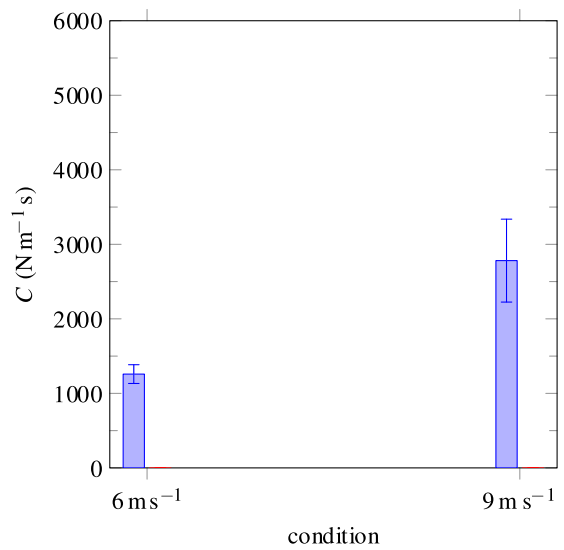
(a) K for seatbelt.



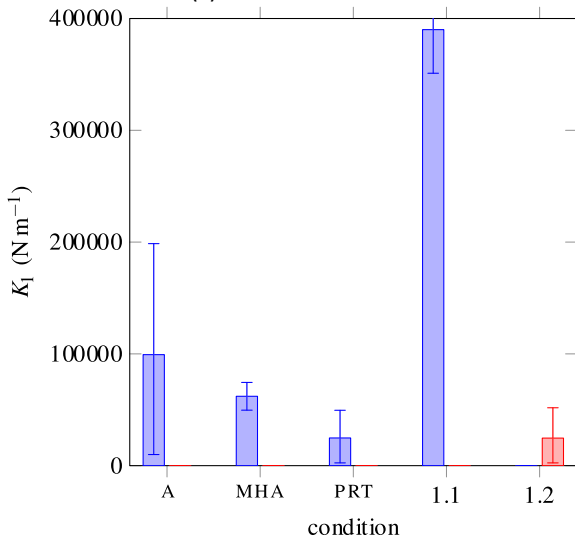
(b) K_{eff} for impactor.



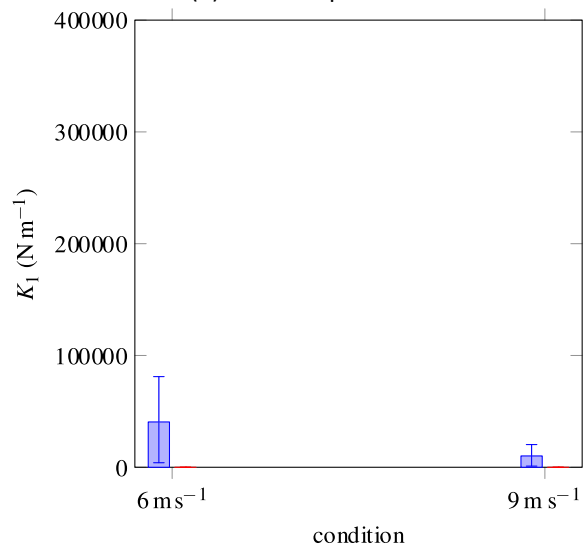
(c) C for seatbelt.



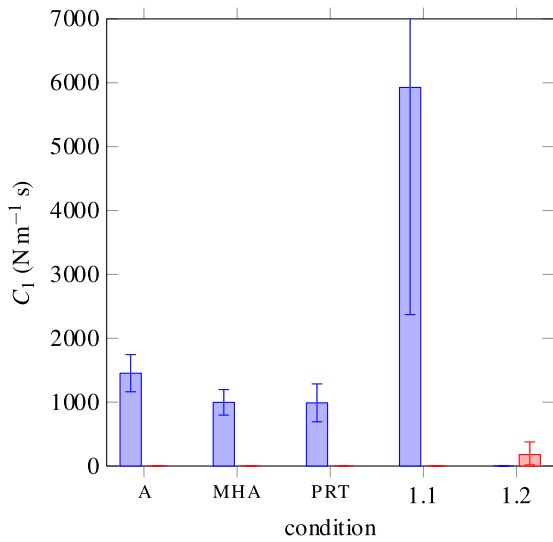
(d) C for impactor.



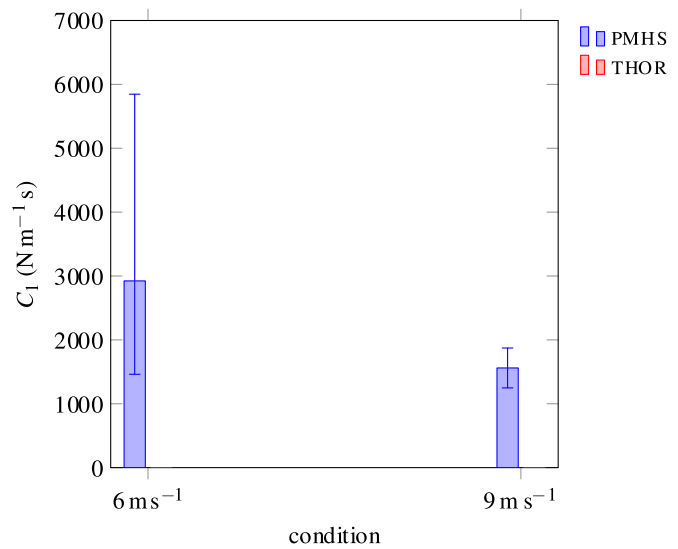
(e) K_I for seatbelt.



(f) K_I for impactor.



(g) C_1 for seatbelt.



(h) C_1 for impactor.

Fig. A.2. Identified model parameters for additional seatbelt and impactor loading conditions.

A: A condition from Foster et al. 2006

MHA: MHA condition from Lamielle et al. 2008

PRT: PRT condition from Lamielle et al. 2008

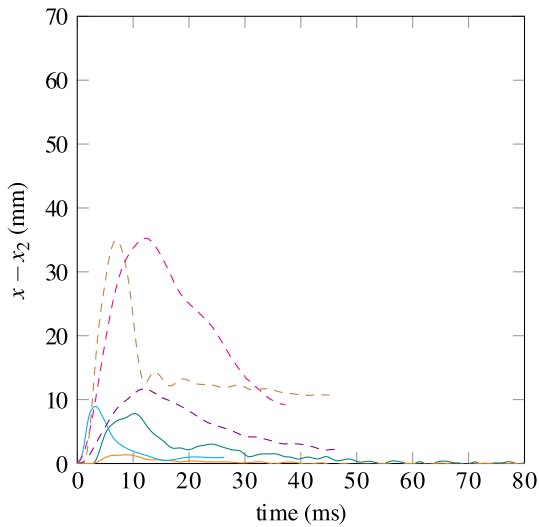
1.1: configuration 1 from Trosseille et al. 2002 (PMHS)

1.2: configuration 1 from Trosseille et al. 2002 (THOR)

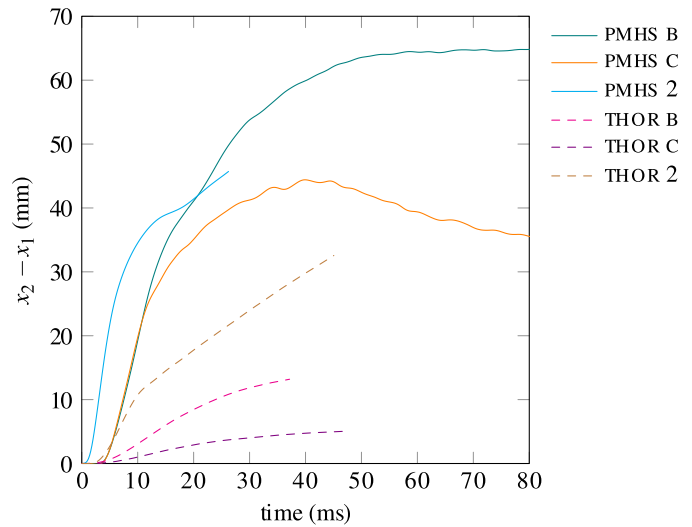
PMHS 6 m/s: 48 kg 6 m/s condition from Hardy et al. 2001

PMHS 9 m/s: 48 kg 9 m/s condition from Hardy et al. 2001

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(a) Deformation of K spring.



(b) Deformation of C damper.

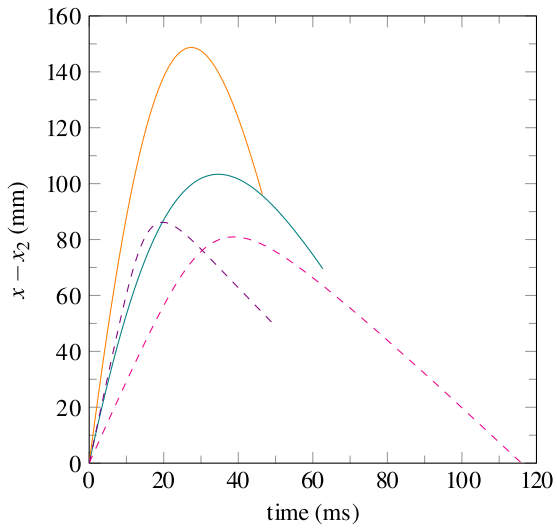
Fig. A.3. Displacement results for seatbelt loading conditions.

B: B condition from Foster et al. 2006

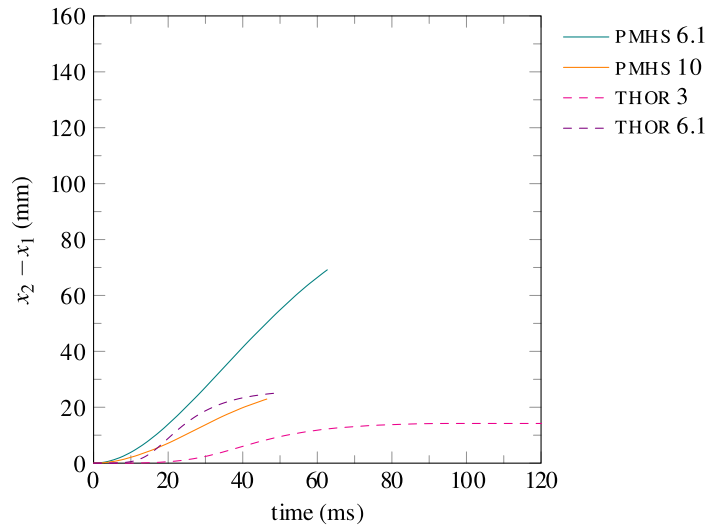
C: C condition from Foster et al. 2006

2: configuration 2 from Trosseille et al. 2002

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(a) Deformation of K spring.



(b) Deformation of C damper.

Fig. A.4. Displacement results for impactor loading conditions.
 PMHS 6.1 m/s: 32 kg 6.1 m/s condition from Cavanaugh et al. 1986
 PMHS 10 m/s: 32 kg 10 m/s condition from Cavanaugh et al. 1986
 THOR 3 m/s: 32 kg 3 m/s condition from Compigne et al. 2015
 THOR 6.1 m/s: 32 kg 6.1 m/s condition from Compigne et al. 2015