Abdominal injury analysis of a 6-year-old pedestrian finite element model in lateral impact

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Abstract: A previously developed finite element (FE) model of a 6-year-old pedestrian abdomen was used to analyse internal organs injuries in lateral impact tests in conjunction with scaling methods. The model was applied to reconstruct adult abdominal cadaver experiments in lateral impact to verify its biofidelity by comparing simulation results with scaled experimental response corridors. Simulation results showed that the abdominal force-deformation curves were well matched with the scaled experimental corridors in different impact speeds. The maximum values of abdominal impact force, deformation and viscous criterion (VC) were proportional to impact velocity. In terms of compression and viscous criterion, the paediatric abdomen had a 25% probability risk of AIS4+ (Abbreviated Injury Scale) abdominal injury in impact velocities of 6.7 m/s and 9.4 m/s. Judging by the first principal strain, contusion or rupture of the left kidney, stomach and spleen appeared in simulations of 6.7 m/s and 9.4 m/s, while liver rupture appeared only in simulations of 9.4 m/s. Predicted internal organ injuries were found to be consistent among the force, deformation, and VC basis injury criteria. The maximum abdominal impact force was inversely proportional to the impact angle, while the abdominal deformation was proportional to the impact angle. Therefore, the model can be further applied to analyse abdominal injuries for a 6-year-old human in pedestrian impact.
1 Introduction

The mortality rate of paediatric pedestrians aged 6–9 is the largest in traffic accidents, according to studies by DiMaggio and Durkin (2002) and Miller et al. (2004). Road traffic injury was the second leading cause of death for children aged 5–14, and one-third of these mortalities were found from pedestrian impact (Crandall et al., 2013). Paediatric pedestrians are regarded as a vulnerable group in road traffic users (Mou, 2004). There is a noticeable difference in the incidence rate of tissue injury between adult and paediatric pedestrians. Although the incidence rate of abdominal injury was lower than that of the head and limbs (Ivarsson et al., 2006), even minor injuries of abdominal internal organs can lead to death in children. As reported by Rouhana and Foster (1985), abdominal injury accounted for 20% of serious injuries (AIS≥4, Abbreviated Injury Scale) in side impact. Therefore, the study of paediatric abdominal injury biomechanics in lateral impact is of great significance on how to protect paediatric pedestrians, and to formulate biomechanically sound children safety regulations.
Studying children’s abdominal injuries through cadaver experiments is difficult because, among other things, it is difficult to obtain paediatric cadaver samples. To the best of our knowledge, there have only been a few paediatric abdominal experiments conducted so far, as reported by Kallieris et al. (1976), Kent et al. (2006, 2009), and Ouyang et al. (2015). Kallieris et al. (1976) conducted sled tests with different velocities (30 km/h, 40 km/h) using four cadaver samples aged 2.5, 6, 6 and 11. The tested subjects were restrained by a lap belt with a shield. They reported that several haemorrhages of muscles, discs and ligaments were noticed in autopsy reports except in the abdominal internal organs. Kent et al. (2009) loaded the upper/lower abdomen of a 7-year old cadaver to obtain abdominal posterior reaction force-deformation under belt restraint. The experimental results showed that the lower abdominal responses of the testing subject were consistent with the experimental response corridor of porcine conducted by Kent et al. (2006). Ouyang et al. (2015) conducted front abdominal impact experiments with nine paediatric corpse samples aged 2–14 at different impact velocities. Experimental results included abdominal impact force-deformation curves, time history curves of VC and abdominal visceral injury information. Paediatric abdominal impact response corridors were developed based on the experimental data (Ouyang et al., 2015). These studies mentioned above provided useful data and valuable references for model validation when developing paediatric abdomen FE models.

Although finite element modelling of human children has played a role in injury biomechanics, there are not many high biofidelic paediatric abdominal finite element models that have been developed so far. Mizuno et al. (2005) developed a finite element model of a 3-year-old child by scaling from a 50 percentile FE model. The validity of the scaled 3-year-old human model was verified by performing a reconstructing calibration test of the 3-year-old Hybrid III dummy by comparing simulation results with response corridors from dummy tests. The scaled model may not accurately represent the unique anatomical characteristics of children, since children are not simply scaling versions of adults. Therefore, it may be of significance to develop a finite element model of paediatric abdomen with detailed anatomical structures from real human child geometry.

A 6-year-old pedestrian abdomen finite element model with real anatomical structure has been previously developed by Lv et al. (2015). The model was validated by reconstructing abdominal frontal impact tests of paediatric samples conducted by Ouyang et al. (2015). Since most paediatric pedestrian injuries in traffic accidents involve lateral impacts, it is important to also verify the 6-year-old FE model with lateral impact experimental data. Then the FE model can be used for studying child pedestrian injury in omnidirectional impacts. Owing to the lack of lateral impact experimental data from paediatric abdominal samples, the model was used to reconstruct adult abdominal side impact experiments by Viano et al. (1989) with scaling method. The impactor mass used in the simulation was obtained by SAE (Society Automotive Engineers) scaling method calculated from the adult tests. The validity of the model was verified by comparing the simulation results with experimental corridors after scaling. In order to study the effects of impactor angles on abdominal injury responses, the angles of the impactor in the 4.5 m/s simulation were set as 0° (pure lateral impact), 15° and 30° respectively as seen in Figure 2. Internal organ injuries in different impact velocities and impact angles were predicted based on tissue-level of first principal strain, compression, and viscous criteria.
2 Materials and methods

2.1 Description of the 6-year-old pedestrian abdomen FE model

The model used in the current study is the one previously developed in Lv et al. (2015). It includes lumbar vertebra, intervertebral disc, end plate, diaphragm, stomach, liver, spleen, kidney, large intestine, small intestine, bladder, blood vessels, oesophagus, skin, ligaments and abdominal muscles as shown in Figure 1. The abdominal muscles include rectus abdominis, obliquus internus abdominis, latissimus dorsi muscle, erector spinae muscle and psoas muscle. The cancellous bone of the lumbar, abdominal internal organs, and muscles were modelled in solid element. Shell elements were used to model the cortical bone, end plate and skin. All ligaments were modelled by membrane elements.

Figure 1 6-year-old pedestrian abdomen finite element model

2.2 Material properties

Owing to the paucity of material parameters for paediatric tissues obtained from related studies, the material parameters used in the model were got by scaling from adult material properties, which can be found in the paper previously published by the same authors (Lv et al., 2015).

2.3 Formulation for data scaling

The impactor mass and paediatric abdominal impact force-deformation corridors were obtained with SAE scaling method (Mertz et al., 1989). The detailed process is as follows: equations (1) and (2) are derived from the law of conservation of momentum and energy, respectively.

\[ m_i v_i = (m_i + m_j) v_i \]  
\[ \frac{1}{2} m_i v_i^2 = \frac{1}{2} (m_i + m_j) v_i^2 + \frac{1}{2} K_s D^2 \]
where $D$ and $F$ are the value of abdominal deformation and impact force, $m_i$ and $v_0$ are the mass and initial velocity of impactor, $m_A$ and $K_A$ are the mass and stiffness of abdomen, respectively. The abdominal responses were obtained from equations (1) and (2):

$$D = v_0 \sqrt{m_i m_i / K_A (m_A + m_i)}$$

(3)

$$F = K_A D$$

(4)

$$\lambda_D = \frac{m_{IC} m_{IC} (m_{AC} + m_{IC})}{\sqrt[3]{K_A m_{AC} m_{AC} (m_{AC} + m_{IC})}}$$

(5)

$$\lambda_F = \frac{\lambda_A \lambda_D}{\lambda_F}$$

(6)

where the subscripts $C$ and $A$ represent child and adult, $\lambda_D$ is the ratio of child and adult abdominal deformation, $\lambda_F$ is the ratio of child and adult impact force. The abdomen stiffness is obtained by referring to the study of Ito et al. (2009).

$$K_A = \frac{E h b^3}{2 r^3}$$

(7)

$$\lambda_E = \frac{\lambda_A \lambda_B \lambda_h^3}{\lambda_F^3}$$

(8)

where $h$ is the abdomen height, $b$ is the abdomen width, $r$ is the abdomen equivalent radius, $\lambda_E$ is the ratio of abdomen stiffness between child and adult.

If consider the abdomen as a cylinder, then $\lambda_E = \lambda_B \lambda_h$ since $\lambda_D = \lambda_Y$ and $\lambda_Y = 1$ according to SAE scaling method (Mertz et al., 1989). Then the impactor mass of child test $m_{IC}$ can be expressed as in equation (9).

$$m_{IC} = \lambda_{IC} m_{IC} m_{IC} / [m_{IC} + m_{IC}(1 - \lambda_E)]$$

(9)

Yamada (1970) found that there was no significant relationship between the material properties of soft tissue and age, so $\lambda_E$ is set as 1. The values of $\lambda_A$ and $\lambda_Y$ are 0.735 and 0.695, according to Table 1. The impactor mass of paediatric abdomen tests is 8.31 kg, obtained from equation (9). The initial speed of impactor is set as 2.5 m/s, the diameter of impactor: $d_i = d_A \lambda_z = 150 \times 0.735 = 110$ mm. The values of $\lambda_D$ and $\lambda_F$ are 0.695 and 0.511.

| Table 1 | Abdominal parameters for six-year-old child and 50 percentile adult male |
|---------|--------------------------|--------------------------|
| **Sitting height (mm)** | **Hip breadth (mm)** | **Reference** |
| Six-year-old child (C) | 628 | 212 | GB/T 26158-2010 |
| 50 percentile adult male (A) | 855 | 305 | GB 10000-88 |
| C/A | 0.735 | 0.695 |

2.4 Simulation configuration set-up

According to the impact conditions by Viano et al. (1989), the impactor in the simulation was considered as a rigid cylinder. The collision position was located at a point below the xiphoid process 55 mm for the 6-year old, according to the information given by the adult impact experiment after scaling. In order to make the impact force through the centre of gravity of abdomen, the impactor was rotated forward by 30° in the horizontal plane, as shown in Figure 2. The speeds of the impactor were set as 4.5 m/s, 6.7 m/s and 9.4 m/s, which was consistent with the corpse experiments. Simulation results and scaled experimental corridors were compared to verify model biofidelity. Finally, internal organ injuries were predicted in terms of compression criterion, viscous criterion and first principal strain. In addition, in order to study the effects of impactor angles on abdominal injury responses, the angles of the impactor in the 4.5 m/s simulation were set as 0° (pure lateral impact) and 15° respectively as seen in Figure 2. The simulation analysis was completed using nonlinear collision simulation software PAM-Crash (ESI-Group, Paris, France).

Figure 2 Simulations of paediatric abdomen in lateral impacts

3 Results

3.1 Model validation in lateral impacts

Figure 3 shows the abdominal impact force-deformation curves in different impact velocities between simulations and scaled adult experimental corridor responses. In Figure 3, the black solid line and red dashed line represent scaled experimental response corridors and simulation results. It can be seen from Figure 3 that simulation responses
are located within the scaled experimental corridors, indicating that the simulation predictions are in good agreement with the experimental result. From Figure 3(b), we found that the abdominal model showed a marginally stiffer response initially (from 0 mm to 15 mm), since the simulation curve was located outside of the corridor a little.

**Figure 3** Force-deformation comparisons between simulations with scaling experimental corridors at different speeds: (a) 4.5 m/s, (b) 6.7 m/s, (c) 9.4 m/s
Table 2 shows the maximum impact force, maximum deformation, maximum VC and corresponding injury threshold. The paediatric abdominal injury threshold was obtained by scaling that of an adult from Ivarsson et al.’s (2004) research. Figure 4 shows the relationship between abdominal maximum impact force/deformation/VC and impact speeds from the simulations. The red solid line and black dashed line denote the linear fitting results of simulation responses and the corresponding injury thresholds. As shown in Figure 4, the maximum impact force, deformation and VC are proportional to impact velocity. It can be seen from Figure 4(a) that the maximum impact force in the 9.4 m/s simulation exceeds the injury threshold 1.82 kN, which indicates that there is a 25% probability risk of AIS4+ abdominal injury. From Figure 4(b, c), the maximum deformation and VC in the 6.7 m/s and 9.4 m/s simulations are above the corresponding injury thresholds (79 mm and 1.6 m/s), respectively.

Table 2  Simulation results of paediatric abdomen in different impactor velocities

<table>
<thead>
<tr>
<th>Impactor velocity (m/s)</th>
<th>Force (kN)</th>
<th>Deformation (mm)</th>
<th>VC(_{\text{max}}) (m/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>4.5</td>
<td>1.002</td>
<td>71.2</td>
<td>1.046</td>
</tr>
<tr>
<td>6.7</td>
<td>1.675</td>
<td>88.2</td>
<td>1.751</td>
</tr>
<tr>
<td>9.4</td>
<td>2.561</td>
<td>108.3</td>
<td>2.426</td>
</tr>
<tr>
<td>Injury threshold</td>
<td>1.820</td>
<td>79</td>
<td>1.600</td>
</tr>
</tbody>
</table>
Figure 4 Maximum force, deformation, VC of paediatric abdomen VS impact velocity of simulations: (a) maximum force, (b) maximum deformation, (c) $V_{C_{\text{max}}}$
Figure 4  Maximum force, deformation, VC of paediatric abdomen VS impact velocity of simulations: (a) maximum force, (b) maximum deformation, (c) VC$_{\text{max}}$ (continued)

Figure 5  The first principal strain contours of paediatric internal organs in different impact velocities: (a) stomach, (b) spleen

(c)

(a) 4.5 m/s, stomach

(b) 4.5 m/s, spleen

(a) 6.7 m/s

(b) 9.4 m/s
3.2 Internal organ injury

Figure 5 shows the contours of the maximum first principal strain of stomach and spleen with different impact velocities at the time $t = 13.5$ ms. As shown in Figure 5(a), the maximum first principal strain of stomach appeared at the side of gastric body. The first principal strain in the same part of the stomach increases with the increase of the collision velocity, and the compression deformation of the stomach is proportional to impact velocity. From Figure 5(b), the contour of the first principal strain of spleen is in a ‘monkey face’ distribution, and the maximum first principal strain of spleen located at the middle and lower edge of visceral surface, corresponding to the position of ‘monkey eyes’ and ‘monkey nose’, respectively. Contours of the maximum first principal strain for other organs are not shown for brevity, but it is worth mentioning the location of the maximum first principal strain of left kidney located in the contact surface between kidney and transverse process of lumbar vertebra.

**Figure 6** Impact velocity VS maximum first principal strain for paediatric abdomen internal organs: (a) liver, spleen and stomach, (b) kidney, large and small intestines
The ultimate strain of 30% for liver injury was obtained from macaque liver loading tests from Melvin et al.'s (1973) study, so the first principal strain threshold values of liver, spleen, stomach, and kidney were defined as 30% in the paper. Yamada (1970) found that the ultimate strain of intestines was 120%. Figure 6 shows the maximum first principal strain of paediatric internal organs from the simulations. From Figure 6(a), the maximum first principal strain of liver, spleen, and stomach increases with the increase of the impact velocity. Under the same impact velocity, the maximum first principal strain of spleen is larger than that of the liver and smaller than that of the stomach. In the simulations of impact velocity of 6.7 m/s and 9.4 m/s, the maximum first principal strain of stomach and spleen exceed the injury threshold by 30%, indicating that they may have a different degree of tissue contusion. In the simulations of impact velocity of 9.4 m/s, the maximum first principal strain of liver exceeds the injury threshold by 30%, indicating that contusion may occur. As shown in Figure 6(b), the maximum principal strains of kidney, large, and small intestine are proportional to impact velocity. Under the same impact velocity, the maximum first principal strain of the left kidney is larger than that of the right kidney, mainly because the left kidney is located on the collision side. The maximum first principal strain of the intestine is greater than that of the left kidney; the main reason is that the kidney (solid organs) has a stronger tendency to resist deformation than the intestines (hollow organs). In the simulations of impact velocities of 6.7 m/s and 9.4 m/s, the maximum first principal strain of the left kidney exceed the injury threshold of 30%, indicating that contusion may occur. The maximum first principal strain of the right kidney and intestines are smaller than the corresponding injury threshold of 30% and 120%, showing that no injury may occur.

3.3 Simulation results from different impact angles

Figure 7 shows the maximum force and deformation of paediatric abdomen in the simulations of impact angles of 0°, 15° and 30°. As shown in Figure 7, the abdominal maximum impact force was inversely proportional to the impact angle, and the abdominal deformation was proportional to the impact angle.

Figure 7 Maximum force, deformation of paediatric abdomen VS Impact angles of simulations: (a) maximum force, (b) maximum deformation
Figure 7  Maximum force, deformation of paediatric abdomen VS Impact angles of simulations: (a) maximum force, (b) maximum deformation (continued)

Figure 8 shows the maximum first principal strain of paediatric abdomen in simulations of impact angles of 0°, 15° and 30°. It can be seen from Figure 8 that the maximum first principal strains of the large intestine, small intestine and left kidney were inversely proportional to the impact angle, and the maximum first principal strains of the liver and stomach were proportional to the impact angle. Under the same loading conditions, the maximum first principal strain of the left kidney was smaller than that of the large intestine. With the increase of impact angle, the rate of change of the maximum first principal strain of large intestine was larger than that of the small intestine. In addition, the maximum first principal strain of the spleen in simulation of 15° was larger than that of the other simulations of 0° and 30°.

Figure 8  Maximum first principal strain of paediatric abdominal internal organs VS Impact angle
4 Discussion

4.1 Model validation

Owing to the lack of paediatric material parameters and cadaver experimental impact data, the material parameters used in the model were obtained by scaling method, and so was the experimental impact data. Through the reconstruction of adult abdominal lateral experimental impacts, simulation results are well compared with scaled experimental corridors. This not only proves the model validity (Figure 3), but also indicates that the scaling method may be a feasible approach to obtain paediatric material parameters at present. With the further development of human tissue material parameters and child experimental impacts data, the validity of the model will be further verified by reconstructing paediatric abdominal cadaver experiments with real paediatric materials in the future.

4.2 Visceral injury predictions

According to the impact force injury threshold, the paediatric abdomen has a 25% probability risk of AIS4+ abdominal injury in the simulation of the collision speed of 9.4 m/s (Figure 4a). The rupture of the abdominal aorta, kidney or liver and the laceration of the liver belong to AIS4+ abdominal injury, as defined by the American Automobile Medical Association (AAAM, 2005). In terms of the first principal strain as injury criterion, it is seen that the liver, kidney, spleen and stomach produced certain degrees of contusion or rupture (Figure 6a), which suggested that the abdominal area suffers an AIS4 injury. On the basis of compression and viscous criteria, the paediatric abdomen has a 25% probability risk of AIS4+ abdominal injury in the simulations of the collision speeds of 6.7 m/s and 9.4 m/s (Figure 4b, c). Using the first principal strain as a criterion, the left kidney, spleen and stomach may have contusion or rupture in simulation speed of 6.7 m/s; the left kidney, liver, stomach and spleen seem to show contusion or rupture in simulation speeds of 9.4 m/s. In summary, visceral injury predicted by first principal strain is consistent with that predicted by compression and viscous criteria; the latter can predict injury types and the former can predict the site and degree of visceral injury.

5 Conclusions

A 6-year-old pedestrian abdomen FE model has been validated using adult lateral cadaver impact data. Simulation results showed that the abdomen impact force-deformation responses are in agreement with the scaled experimental corridors. The maximum values of abdominal impact force, deformation and viscous index are proportional to the impact velocity. Under the same impact velocity, the maximum first principal strain of left kidney located at the collision side is larger than that of right kidney. Maximum first principal strains of hollow organs (intestines) are larger than that of solid organs (liver, kidney). Visceral injury predicted by first principal strain is
consistent with that predicted by compression and viscous criteria. In the future, this model can be applied in the development of a full size six-year-old paediatrics pedestrian finite element model, and provide references and support for establishing abdominal injury criteria for the protection of children from abdominal injury.

References


