The Influence of Shoulder and Pelvic Belt Floor Anchorage Location on Wheelchair Occupant Injury Risk: a simulation study

Lionel Cabrolier, Raymond D'Souza, Gina Bertocci, John Tiernan, Ciaran Simms

Abstract It has been previously shown using computational modelling of a 50th percentile adult seated in a commercial wheelchair that the occupant pelvic-belt angle with respect to the horizontal should be between 45°-60° to reduce pelvic loading and prevent submarining, instead of 30°-75° as recommended by voluntary ISO standards. This paper extends the previous work by using computational modelling to evaluate the influence of combined shoulder-belt and pelvic-belt position on a 5th percentile female, a 50th percentile male and a 95th percentile male occupant to study kinematics and loading in a standardized 20g frontal impact. The belt configuration is more realistically represented. The results show that, for all dummy sizes, the horizontal component of the pelvic-belt force remains broadly constant as the pelvic-belt angle increases from 30°-75°, but the vertical pelvic-belt force triples. A critical belt angle of around 45° was identified, below which predicted abdomen loading, pelvis rotation and head/neck injury risks are increased. The shoulder-belt force increased with pelvic-belt angle. Moreover, pelvic-belt angles higher than 60° greatly increased occupant excursions, abdomen loading and head/neck/chest injury risks. These results again indicate an optimum pelvic-belt angle in the region of 45°-60°, depending on the occupant size.

Keywords abdomen loading, injury criteria, pelvic-belt angle, wheelchair user

I. INTRODUCTION

Many wheelchair users remain seated in their wheelchairs during transit. Their numbers are significant, and we now estimate a minimum of 1400 wheelchair users in Ireland alone, taking at least 500,000 road trips annually [1-2]. In these cases, the safety features incorporated in standard vehicle seats and restraint systems must be emulated by the wheelchair and Wheelchair Tie-down and Occupant Restraint System (WTORS).

Frontal collisions dominate serious vehicle collisions, and the main focus of wheelchair safety research has been on preventing injury through occupant retention in frontal impacts [3-4], and developing crash protection for pediatric cases [5]. Ensuring the structural integrity of the wheelchair, tethering the wheelchair to the vehicle, and occupant retention in the wheelchair via WTORS are considered the principal features required to reduce wheelchair occupant injury risk in frontal impact. These objectives are embodied in the voluntary ISO standard 10542 (ISO 2001) (which tests the WTORS), ISO 7176–19 (ISO 2001) (which tests the wheelchair), and ISO 16840-4 (ISO 2008) (which tests seating systems independently of the wheelchair base).

A major aspect of wheelchair seating safety is facilitating adequate belt fit on the user's body to ensure good body restraint during an impact. The three-point pelvic and shoulder-belt system is designed to apply restraint forces to both the upper and lower torso to accomplish the required velocity change in a given severity of collision. For the pelvic belt, the restraint forces should be applied to the bony prominences of the pelvis rather than the abdomen to prevent underlying soft tissues being exposed to injury [6]. For the shoulder belt, forces should be applied away from the neck and the belt should fit the upper torso with a particular geometry (location and angle) that can be seen in Fig. 1 [12]. To achieve this, seats must accommodate the proper fit of occupant restraint systems and they must support the occupant throughout the crash so that the occupant restraints remain properly positioned [7].

The existing evidence does not explicitly show whether existing belt angle recommendations are problematic. However, recent work has documented the results of investigations involving 42 wheelchair-seated occupants, including a biomechanical analysis of each case. The study found that over 70% of the occupants were improperly restrained, either because of non-use or incomplete use of available belt restraints, or because the belt restraints were improperly positioned on the occupant's body. Over 60% of the occupants sustained

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significant injuries and almost 25% died. Thus the evidence indicates that the belt configuration is an important consideration for wheelchair occupant safety [8-9]. To address this issue, wheelchair-integrated occupant restraint systems have been proposed [10-11], similar to standard practice in conventional car seats, thus specifying the position of the shoulder and pelvic belt relative to the occupant. However, the overwhelming majority of restraint systems remain vehicle-mounted and this is likely to remain so for some time.

For vehicle-mounted restraint systems, the angle of the pelvic belt with respect to the vehicle floor (see Fig.2) is important as it influences the location of the principal belt contacts with the pelvis, abdomen and upper torso area as well as the magnitude of the belt loading on the body and thereby its kinematics and injury risks [17]. Voluntary standards such as ISO 10542 [12] stipulate that the allowable pelvic-belt angle should be between 30° and 75° with respect to the horizontal with a preferred zone between 45° and 75°, but these recommended pelvic-belt angles have not been formally evaluated. A recent study, based on a MADYMO 50th percentile male seated in a wheelchair subjected to a 20g frontal impact, showed a pelvic-belt angle between 45°-60° helped in reducing pelvic loading and preventing submarining. However, the study used a fixed shoulder belt anchorage position and a single occupant size. In existing wheelchair belt systems, the shoulder and pelvic belts are generally both connected to the same floor anchor point, and therefore moving this point influences both the pelvic and shoulder belt configurations and consequently the body kinematics and loadings. Accordingly, the goal of this work was to evaluate the influence of occupant size and a more realistic belt modelling approach on the predicted optimum pelvic-belt angle for wheelchair occupants in frontal impact. To do this, computational modelling of wheelchair occupant kinematics and loading in a standardized 20g frontal impact for three different adult occupant sizes was performed: 5th percentile female, 50th percentile male and 95th percentile male.

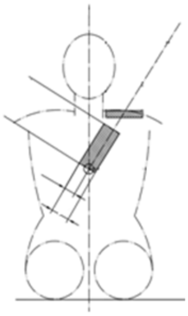


Figure 1 - Preferred zones for location of shoulder belt on occupant's torso, adapted from ISO 10542 [12]

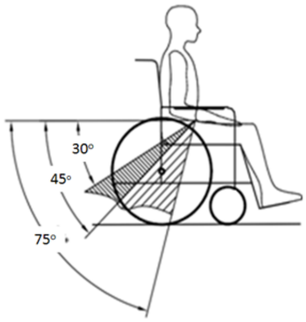


Figure 2 – Recommended and allowable wheelchair pelvic-belt angles, adapted from ISO 10542 [12]

II. METHODS

Computational Modelling

A previously validated MADYMO [13] model of an occupied adult manual wheelchair was used as a baseline for this work [14]. The model was originally developed to study wheelchair and WTORS loading, as well as potential occupant injury risk, making it ideal for the purpose of this study. The model simulates a frontal impact sled test of a Hybrid III 50th percentile male dummy seated in an adult manual wheelchair secured with a vehicle-anchored WTORS. Finite element components are used for the belt portions in contact with the dummy body (and wheelchair frame), and multibody belt components in series with the finite element components are used for the straight sections of the occupant restraint system and for the wheelchair tiedown belts, see Fig. 3.

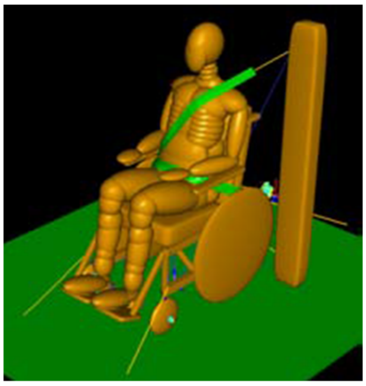


Figure 3 - The baseline wheelchair frontal impact model [14]

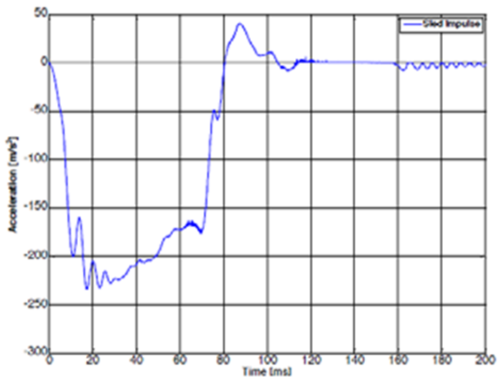


Figure 4 - 20g vehicle acceleration pulse compliant with ISO 10542

This model has been previously validated through comparison with sled test data including pelvic-belt and shoulder-belt force, and accelerations of the wheelchair CG, pelvis, chest and head, all of which were found to lie within acceptance ranges, indicating the model's ability to reproduce the sled test results. In this paper, the location of the belt floor anchorage point was varied parametrically to change the angle of the pelvic-belt connection to the vehicle floor with respect to the horizontal between 30° and 70° in 5° increments, in accordance with the permissible range in the voluntary ISO standard ISO 10542 [12]. The MADYMO belt fitting tool was then used in each case to find the corresponding shoulder belt configurations with respect to the ISO standard specifications. Previous research determined the influence of pelvic-belt angle on abdomen loading but this analysis was based on shoulder and pelvic belts modelled completely separately and using a fixed shoulder-belt floor mounting position (original 45° shoulder-belt position) [16]. In the present study a modified belt system was implemented: both the shoulder and pelvic belts (defined separately) are now linked together at a point close to the occupant's pelvis. A multi-body belt connects this point to the sled-floor anchorage point (called the "junction belt"). This enabled the use of a belt representation closer to a real system which consists of a single belt loop between the shoulder and the lap regions.

A 20g frontal impact crash pulse complying with the ISO 10542 standard was applied to the model as before (Fig. 4). In the baseline model, the occupant was represented by the 50th percentile male Hybrid III dummy model, which has a highly simplified representation of the pelvic and abdominal anatomy. The MADYMO models represent the physical Hybrid III dummies which model the lower torso with a cured cylindrical rubber section for the lumbar spine and a vinyl skin/urethane foam moulded over an aluminum casting for the pelvis. In the MADYMO Hybrid III dummy models this is represented as a single rigid body where the soft urethane foam coated over the hard aluminum casting has a force-penetration curve shown in Fig. 5. As a first approximation of different occupant sizes, the unaltered 5th percentile female, 50th percentile male and 95th percentile male MADYMO Hybrid III dummy models were used, and the influence of floor-anchor point location on the occupant's loading and injury risks was assessed. The baseline MADYMO dummy model is reasonable for predicting the overall occupant body response (loading and kinematic) in the event of a collision. However, to test how the belt interacts separately with the bony structures and the soft tissues of the abdomen region, the more realistic bony pelvis model created by McDonnell et al [16] was utilized. The same bony pelvis consisting of four ellipsoids was inserted in each dummy size, see Fig. 7.

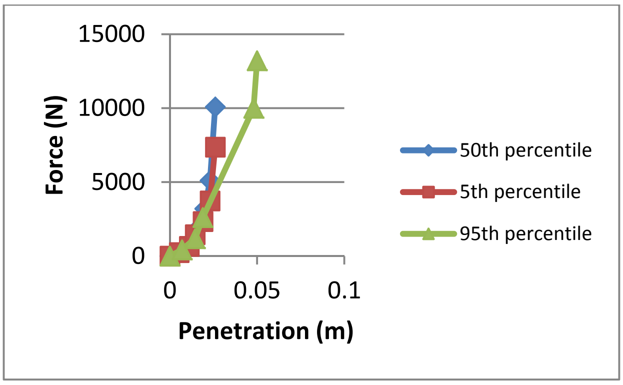


Figure 5 - Baseline MADYMO pelvis force-penetration contact characteristic for the three dummy sizes

Dummy Size	Scaling Factor
5th percentile small female dummy	0.87
95th percentile large male dummy	1.12

Table 1 - Scaling factors used to scale bony pelvis to each dummy size

A scaling factor was applied to adapt the dimensions and position of each dummy size, based on the pelvis ellipsoids used for the MADYMO 50th percentile male dummy. The scaling factor was determined by comparing principal ellipsoid dimensions in the abdomen/pelvis area to the initial 50th percentile male dummy. The scaling applied to the pelvis ellipsoids dimensions and positions are shown in Table 1, and the result can be seen in Figs. 6 and 7.

For each dummy model, the four bony pelvis ellipsoids were given a linear force penetration characteristic based on the slope of the line through the last two points of the corresponding curve (Fig. 5), i.e. the nominal hard tissue portion of the curve. The force penetration characteristic of the original abdomen ellipsoids was then modified to represent the soft tissue by using a linear characteristic based on the initial slope of the force penetration curve in Fig. 5, i.e. the soft tissue portion of the curve. Table 2 reports the stiffness used for the new bony pelvis (hard tissues) and abdomen ellipsoids (soft tissues) for each dummy size.

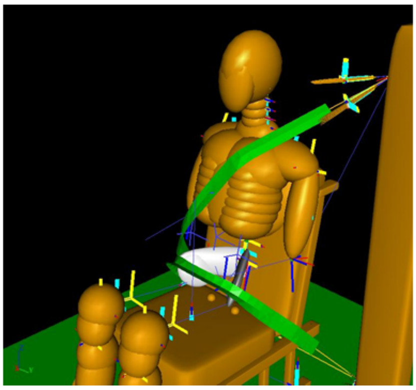


Figure 6 - Representation of the pelvic girdle in 50th percentile male amended MADYMO Hybrid III model

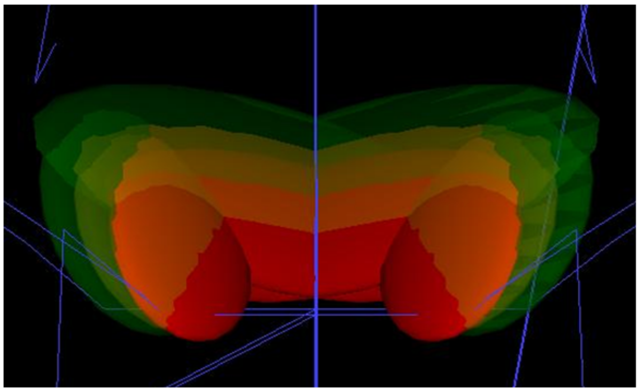


Figure 7 - Pelvis ellipsoids for 5th (Red), 50th (Yellow) and 95th (Green) percentile dummies

Dummy Size	Soft Tissues Stiffness	Hard Tissues Stiffness
5th percentile small female dummy	59.4kN/m	1.2MN/m
50th percentile mid-size male dummy	68.9kN/m	1.7MN/m
95th percentile large male dummy	81.6kN/m	1.7MN/m

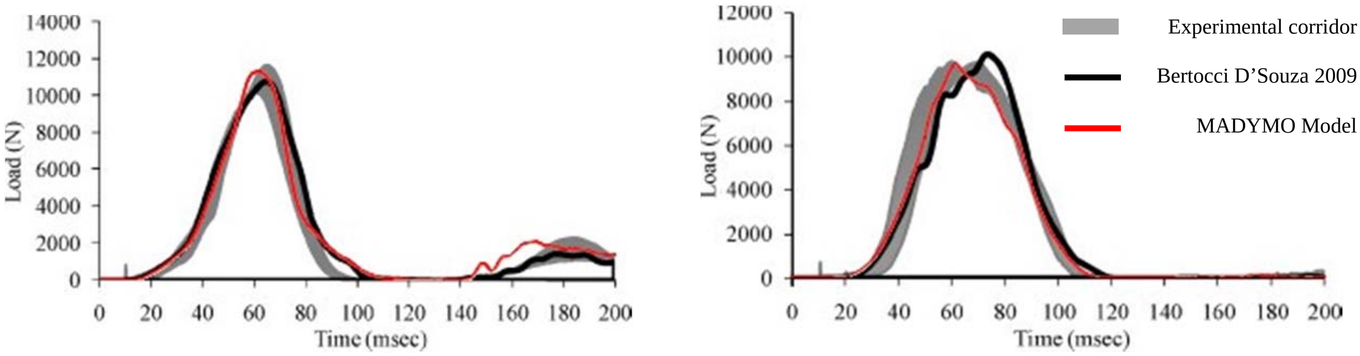
Table 2 - Hard / Soft tissues stiffness used to amended MADYMO models

Reassessment of the 50th percentile male model's ability to reproduce sled test results was first performed. Following this, for the three dummy sizes analyzed (5th percentile female, 50th percentile male and 95th percentile male), the results presented consist of:

1. Analysis of the horizontal and vertical components of the pelvic-belt force over a range of pelvic-belt angles from 30° to 75° for the following occupant representations:
   a. The baseline MADYMO Hybrid III dummy model which treats the pelvic-belt contact with the body using a single combined contact characteristic representing the bony pelvis and soft tissue of the abdomen in series.
   b. The amended MADYMO Hybrid III dummy model with separate contacts defined for the bony pelvis and for the soft tissue of the abdomen.
2. Analysis of the abdomen “soft tissue” compression forces for the amended MADYMO Hybrid III model.
3. Analysis of shoulder-belt forces over a range of pelvic-belt angles from 30° to 75°.
4. Analysis of injury risk criteria (HIC, NTF, Chest deflection).
5. The sagittal plane (XZ) pelvis, head and knee excursions and lower torso rotations for the three amended MADYMO Hybrid III models.

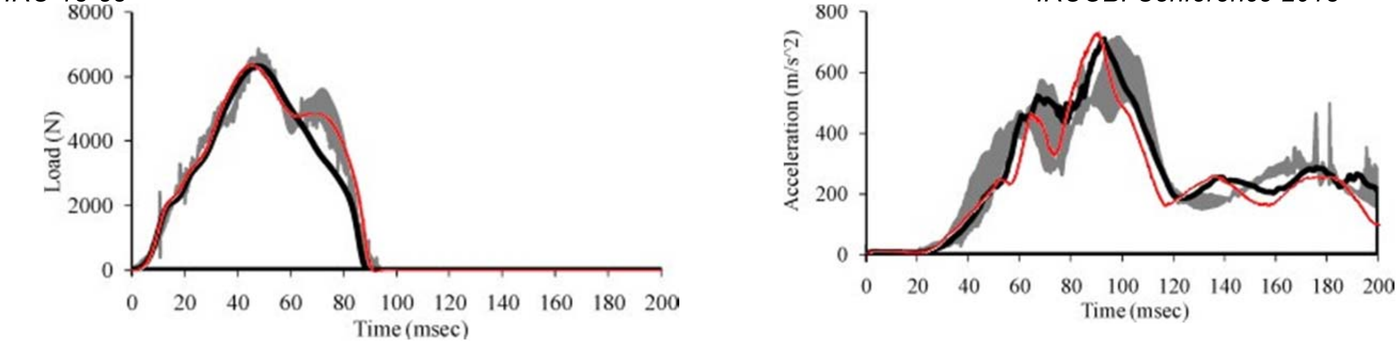
III. RESULTS

To test whether the predictive capability of the baseline model was retained with changes to the belt modelling, the shoulder, left pelvic-belt and tiedown forces time histories for the physical tests were compared to the current model results and the model predictions [14], see Fig. 8.



(a) - Shoulder belt loading

(b) - Left lap belt loading



(c) - Left rear tiedown loading (d) - ATD Head accelerations

Figure 8 – 50th percentile male amended Hybrid III dummy model validation results by comparison to previous MADYMO modelling [14] and sled test data.

The peak abdominal compression forces based on the soft tissue contact ellipsoid are presented for the 5th percentile female, 50th percentile male and 95th percentile amended MADYMO Hybrid III models in Fig. 10. Figure 11 presents the peak shoulder force versus pelvic-belt angle for the three amended Hybrid III dummy models.

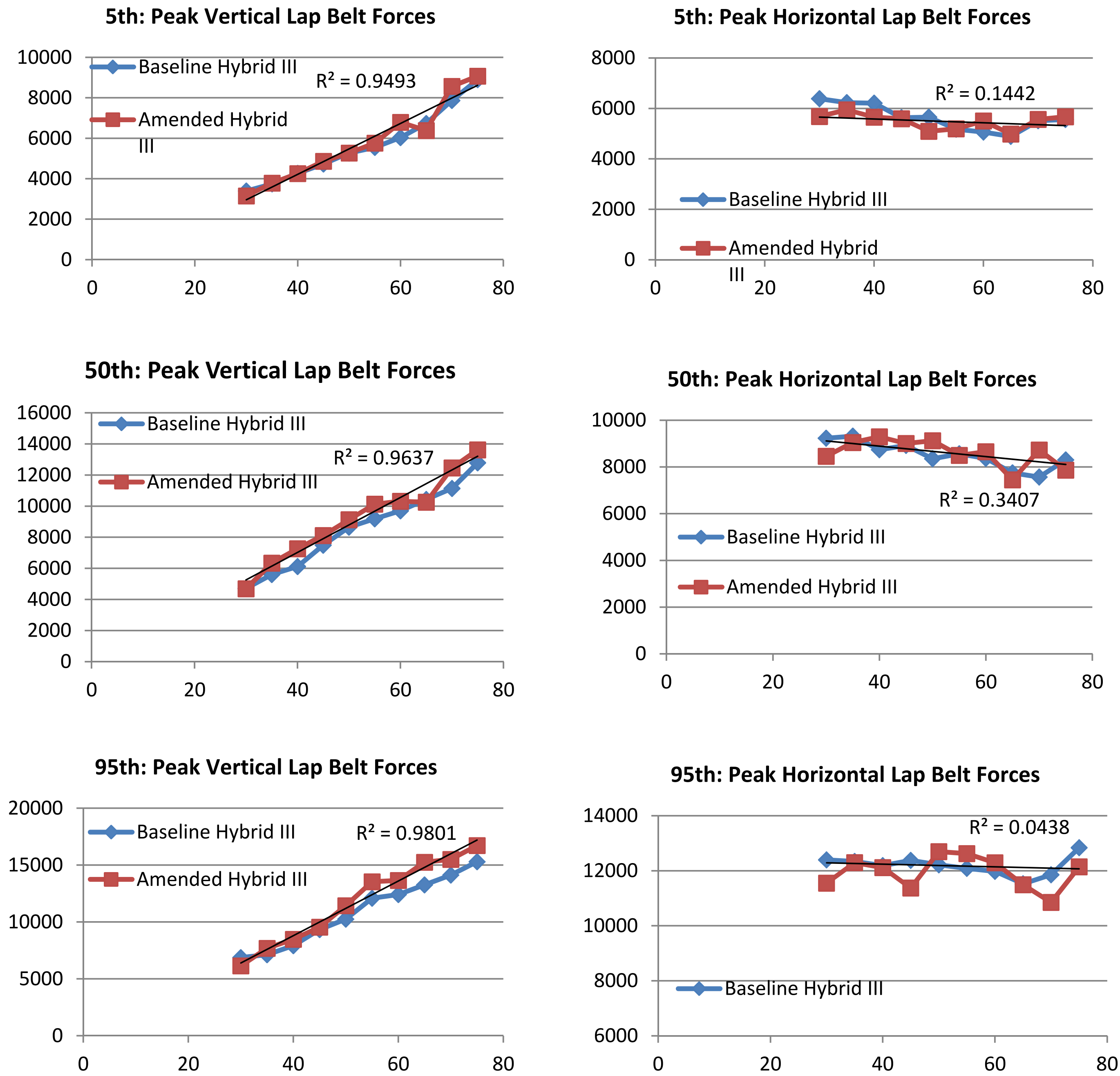
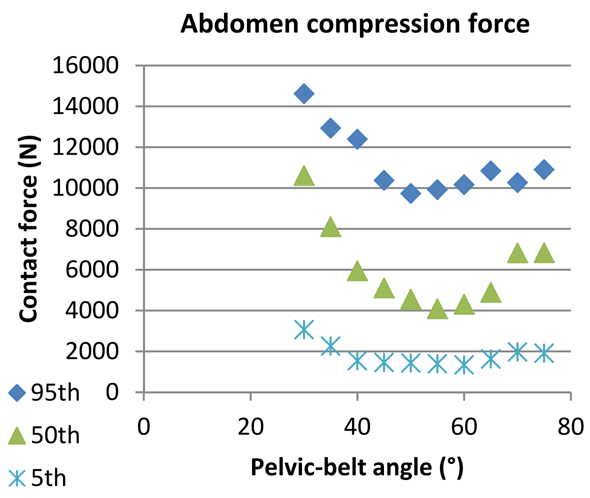


Figure 9 - Peak horizontal and vertical components of pelvic-belt force for 5th percentile, 50th percentile and 95th percentile baseline and amended MADYMO Hybrid III dummy models versus pelvic-belt angle.



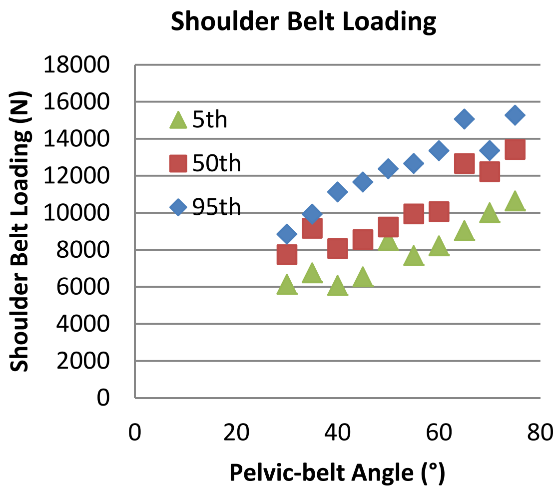
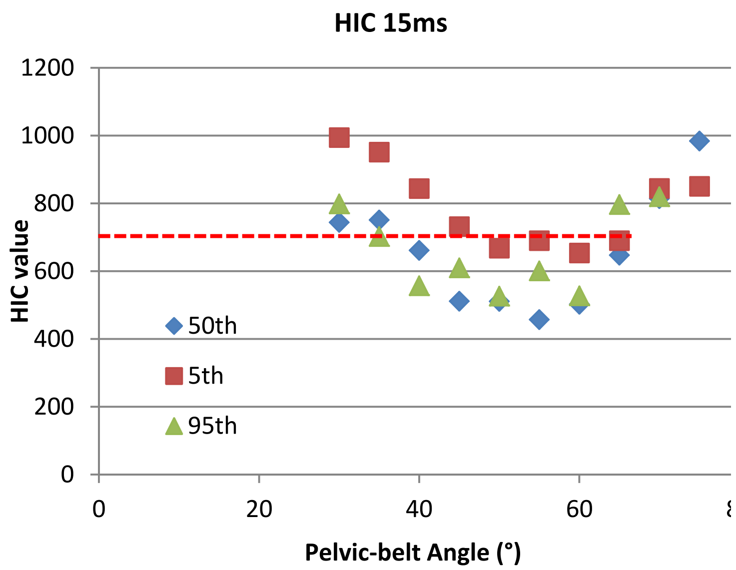


Figure 10 - Abdomen compression force for the three amended MADYMO Hybrid III dummy models.

Figure 11 – Peak shoulder belt loading versus pelvic-belt angle for three amended MADYMO Hybrid III dummy models.

The influence of pelvic-belt angle on the Head Injury Criterion (HIC 15ms) and peak head accelerations for the three dummy sizes are presented in Fig. 12 and Fig. 13 respectively.



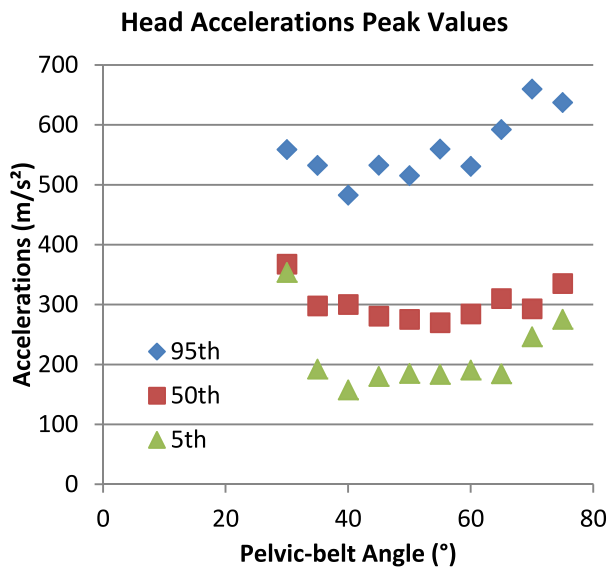


Figure 12 – Head Injury Criteria 15ms for the three amended MADYMO Hybrid III dummy models.

Figure 13 – Peak ATD head acceleration for the three MADYMO Hybrid III dummy models.

The Neck Tension Flexion (NTF) injury criteria while varying the pelvic-belt angle for the three modified MADYMO Hybrid III dummy models are shown in Fig. 14. Fig. 15 and Fig. 16 show the influence of pelvic-belt angle on the peak values of the axial neck tension and chest deflection for the three MADYMO Hybrid III models. Figure 17 shows snapshots of the instant of peak pelvic-belt loading for the three different MADYMO Hybrid III models with 50° initial pelvic-belt angle. The influence of pelvic-belt angle on lower torso sagittal plane rotation for three sizes of modified MADYMO Hybrid III models is shown in Fig. 18. The sagittal plane (XZ) displacement of the pelvis and the knee for the 5th percentile female, 50th percentile male and 95th percentile male modified MADYMO Hybrid III dummy models for different pelvic-belt angles is shown in Fig. 19.

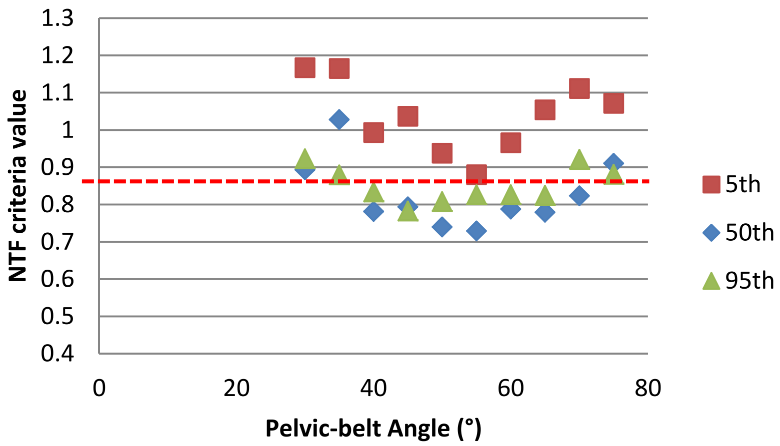


Figure 14 – Predicted Neck Tension Flexion injury criterion for amended MADYMO Hybrid III models

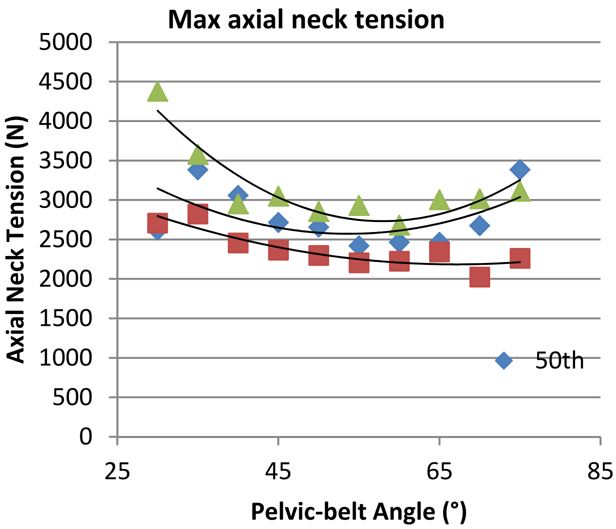


Figure 15 – Peak axial neck tension (three amended MADYMO Hybrid III dummy models)

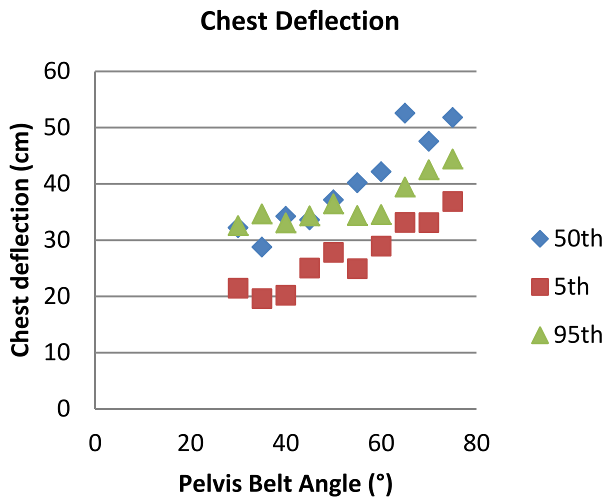


Figure 16 – Peak chest deflection (three amended MADYMO Hybrid III dummy models)

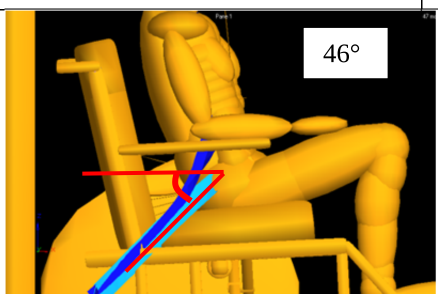
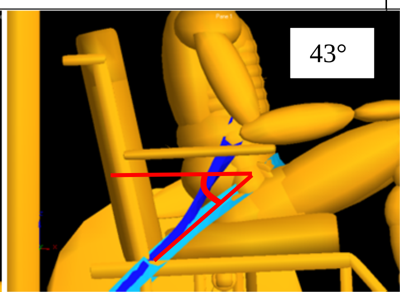
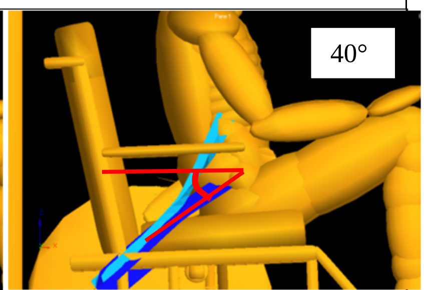
50° Test	5th	50th	95th
Pelvic-belt angle at peak loading (°)			
Excursion at peak loading (m)	(b) 0.3139	(c) 0.3388	(d) 0.3594

Figure 17 – Snapshots of pelvis excursion and pelvic-belt angle at peak pelvic-belt loading for the three amended MADYMO Hybrid III models (50° initial pelvic-belt angle)

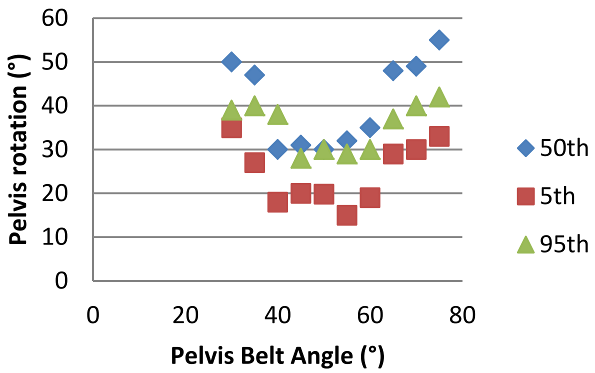


Figure 18 – peak lower torso rotation as a function of pelvic-belt angle (three amended MADYMO Hybrid III dummy models)

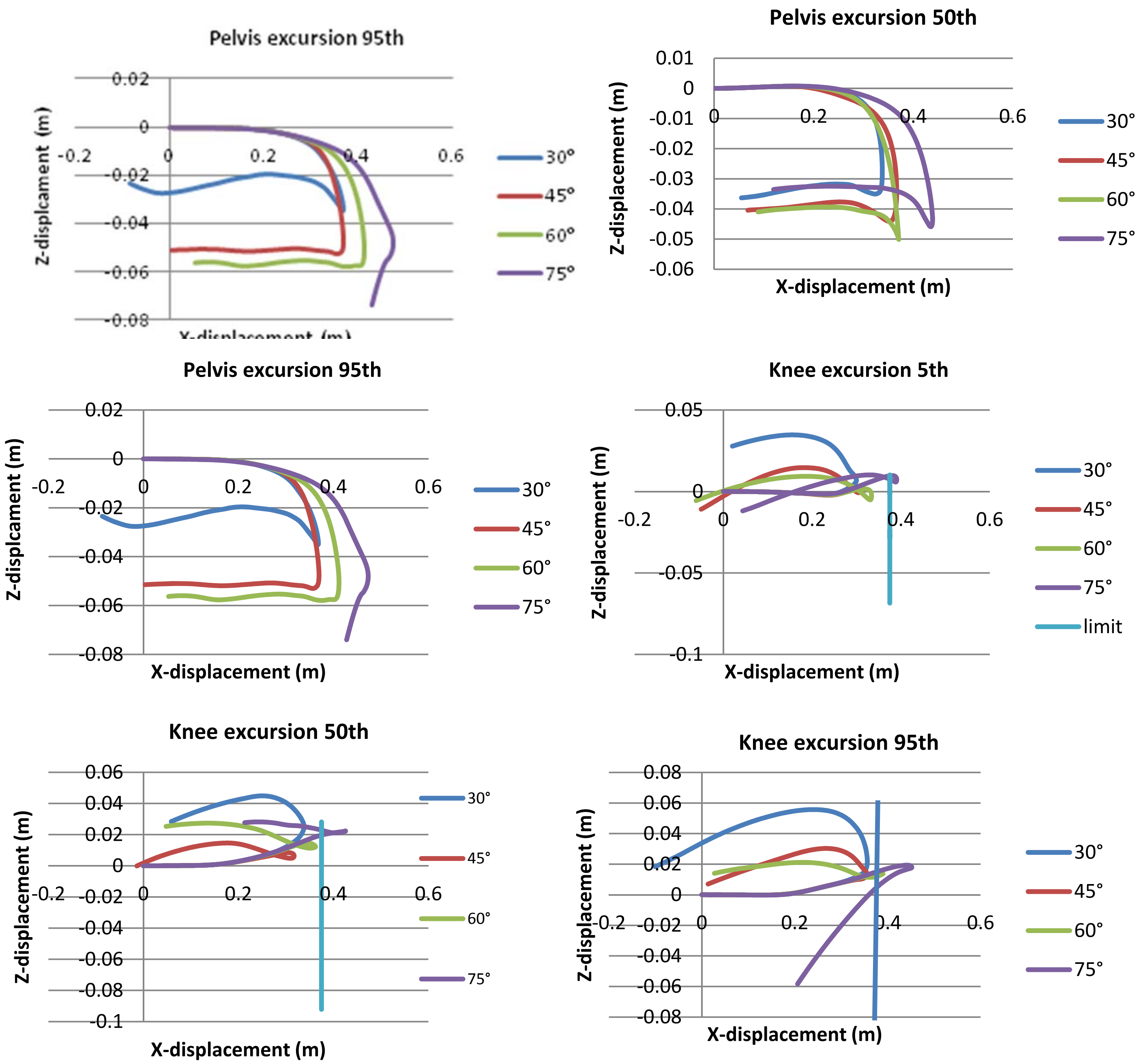


Figure 19 – Pelvis and knee sagittal plane trajectories at different belt angles

IV. DISCUSSION

It has been previously predicted [16] (using computational modelling of a 50% percentile adult seated in a commercial wheelchair) that the angle of the pelvic-belt connection to the vehicle floor should be between 45° and 60° instead of 30° - 75° as recommended by the voluntary ISO standard 10542 [12]. This proposed range of belt angles reduces the pelvic loading while preventing submarining. However this analysis was based on a fixed shoulder belt position and a single occupant size. In existing wheelchair belt systems, the shoulder and pelvic belts are both connected to the same floor anchor point, and therefore moving this point simultaneously influences the pelvic and shoulder belt configurations, and consequently the body kinematics and loadings. A simplified multibody approach was adopted to assess this. Furthermore, the influence of occupant size was assessed by extending the modelling to include the 5th percentile female dummy and the 95th percentile male dummy. Due to the nature of the modelling, the focus is on trends rather than absolute values.

Baseline model findings

The baseline MADYMO model employed in this paper [14] has been previously validated by comparison with sled test results and Fig. 8 shows that the model is largely capable of reproducing the time histories of the experimental results, and even of improving previously published model results for the pelvic belt, the shoulder-belt and the tiedown loading and the ATD accelerations. Thereby it was concluded that the model is appropriate to assess trends in the influence of pelvic-belt angle on occupant loading and kinematics.

Using the 5th percentile female, 50th percentile male and 95th percentile male MADYMO Hybrid III models, and including the shoulder-belt floor anchorage position with the pelvic-belt floor anchorage, we confirmed and expanded to a larger population range the earlier conclusions of McDonnell et al [16]. Maintaining the pelvic-belt angle between 45° and 60° is beneficial: the horizontal pelvic-belt force remains approximately constant (at about 5.5kN, 9kN and 12kN for the 5th, 50th and 95th percentile respectively) but the predicted vertical component of the pelvic-belt force, and hence the overall occupant loading, increases significantly. This can be seen in Fig. 9 (its value at a 30° pelvic-belt angle is approximately one third that at 70°). This is mainly because a more vertical pelvic belt is less efficient at counteracting the horizontal inertial force from the occupant during a frontal impact. The overall behaviour for the three dummy sizes is identical.

Amended model results

On the basis of maximum occupant loading alone (i.e. without separating hard and soft tissue loading), a more horizontal pelvic-belt angle would be more appropriate since this would reduce the magnitude of overall occupant body loading. However, in order to prevent submarining and soft tissue injuries, the contact force between the pelvic belt and associated soft tissue should be minimized. This force reduction is achieved by passing the load through the bony pelvic girdle rather than the soft tissue of the abdomen. The amended MADYMO models enable us to separately study hard and soft tissue loading in a simplified manner by separating the combined bone and soft tissue contact into two separate contacts, one of which represents the interaction between the pelvic belt and the soft tissue of the abdomen, and the other which represents the interaction between the pelvic belt and the bony pelvic girdle. This separation of the body contacts facilitates distinguishing between a desirable bone loading and an undesirable soft tissue loading from the belt in a generalized manner. Fig. 9 shows that the simplified pelvis representation introduced resulted in very similar belt loading for the three amended MADYMO Hybrid III dummy models compared to the baseline MADYMO Hybrid III model. This similarity indicates that the separation of the body contacts employed in this paper has not substantially altered the overall load interaction between the belts and the occupant.

Abdominal loading

Fig. 10 shows that for the three different dummies, the load transmitted directly to the soft tissues of the abdomen for the modified model shows a sharp reduction for increases in pelvic-belt angle from 30° to 40°. The predicted abdomen loading decrease is 1.62kN and 4.66kN respectively for the 5th percentile female and the 50th percentile male as the pelvic-belt angle is increased from 30° to 40°, and 6kN for the 95th percentile male dummy as the pelvic-belt angle is increased from 30° to 50°. After these angles, a more gradual decrease of the abdomen loading appears as the pelvic-belt angle is increased to 60°. Examination of the kinematics from the simulations revealed that, for the lower pelvic-belt angles, the belt failed to engage with the ellipsoids representing the Anterior Superior Iliac Spines (ASIS) and instead slipped over them, leading to the high 'soft tissue' contact load. The modified belt system used for this project included a shoulder and pelvic belt joined at the occupant's right side pelvis. It was observed that the shoulder belt pulled on the pelvic belt and so tended to slip more easily over the bony pelvis. Moreover, a distinction between the different dummy sizes can be made: the minimum belt angle to prevent submarining is higher for the larger dummy sizes. A heavier dummy creates a bigger inertial force and so larger belt forces (see Fig. 10). At peak pelvic-belt loading, larger excursions occur for the larger dummy, reducing the effective pelvic-belt angle, thereby increasing the opportunity for the pelvic belt to slip over the pelvic bones, see Fig. 17. This depends on friction and the shape of pelvic bones which are modelled approximately, but the trends shown here should be robust.

Moreover, Fig. 10 shows a clear increase of the abdomen loading when the increasing pelvic-belt angle is beyond 65°. At this steep pelvic-belt angle, the pelvic belt is less efficient at preventing the excursion of the pelvis, while the shoulder belt still applies restraint to the upper torso through the shoulders. The forward motion of the pelvic region therefore results in relative rearward rotation of the pelvis/torso and this can facilitate slippage of the pelvic belt over the pelvic bones. These large excursions significantly compromise the overall restraining action of the belts. Figure 18 corroborates the previous conclusion, showing large pelvic rotation in the sagittal plane for shallow and steep pelvic-belt angles, induced by a pelvic belt applying the load high on the abdomen.

Upper torso loading

For all three dummy sizes, Fig. 11 shows a shoulder-belt loading increasing by about 4kN/5kN/6kN respectively when increasing the pelvic-belt angle from 30° to 75° (27%) for the 5th/50th/95th percentile amended MADYMO Hybrid III models. This is due to greater occupant excursions that occur for higher pelvic-belt angles, as demonstrated by the sagittal plane pelvic excursions, see Fig. 19. For the 95th percentile male dummy a 30° pelvic-belt angle can result in a slightly larger horizontal excursion before the pelvic belt restrains the pelvis compared to a 45° pelvic-belt angle, because the belt slips over the ASIS and soft tissue compression of the abdomen facilitates further horizontal displacement. However, when the pelvic belt does not slip over the ASIS, increasing the pelvic-belt angle induces a pelvic belt misaligned with the horizontal inertial load of the body, leading to large horizontal excursions for each dummy size at a 75° pelvic-belt angle (0.40m, 0.44m and 0.47m for the 5th, 50th and 95th percentile dummy respectively). Fig. 19 also shows that the sagittal plane knee excursions increase with pelvic-belt angle. At large pelvic-belt angles, these excursions can exceed the allowable limits of 375mm specified in the ISO 7179-19 for the small adult female, midsize and large adult male ATDs.

This constant increase in the shoulder belt loading when increasing pelvic-belt angle can be considered as the reason for greater chest deflections (Fig. 16). A more horizontal pelvic-belt angle would be more appropriate to reduce chest loadings.

Head and neck injury risks

Figure 12 shows significant HIC scores, exceeding the FMVSS-208 allowable limit of 700 [17] at extreme pelvic-belt angles, even without direct head contact. The reliability of the HIC score in these cases is therefore questionable, but nonetheless the results show a parabolic trend for all three dummy sizes which indicates increased inertial head loading for lower and higher pelvic-belt angles compared to intermediate angles. This parabolic trend can also be seen for the peak head resultant acceleration (Fig. 13), the Neck Flexion Tension (NTF) (Fig. 14) and neck axial tension (Fig 15). Both the HIC and NTF plots (Fig. 12 and Fig 14) show that the 5th percentile female is more likely than the 50th and 95th percentile male to suffer severe injuries in the head/neck region due to the frontal crash. However upper torso injury risks are minimized for all three dummy sizes at pelvic-belt angles between about 45° and 60°.

Submarining risk

Viano and Arepally proposed risk criteria for submarining for occupants of automotive seats [15], according to which the pelvis and torso rearward rotation angle should not exceed 30 degrees, and the *H* point horizontal and vertical excursions should not exceed 25cm and 5cm respectively. Comparison of these limits to the results in Fig. 18 indicates that the angle limit is exceeded for a pelvic-belt angle of 30°, 70° and 75° for the 5th percentile female amended MADYMO model dummy, and 30°, 35°, 40°, 65°, 70° and 75° for the 95th percentile male amended MADYMO model dummy. It is unclear why the 50th percentile obtains higher values but nevertheless a general trend can still be recognized: Fig. 18 indicates pelvic rotation for pelvic-belt angles of 30°, 35°, 65°, 70° and 75°. Fig. 19 indicates that the horizontal excursion limits are exceeded in all cases, and the vertical limits are exceeded only for the 95th percentile male amended MADYMO model dummy for pelvic-belt angles of 45° and above. However, in the sled test that was the basis for validating the MADYMO model, the pelvic belt was at 45° and submarining was not observed. Therefore, it is unclear to what extent Viano's automotive seating-based criterion can be directly applied to wheelchair seating.

Limitations

There are several limitations to this work and the most important are listed here: the computer simulations were run using only one crash pulse, the influence of friction and pelvic bone geometry have not been analyzed and the Hybrid III dummy pelvis is also not an ideal reflection of the human pelvis. Accordingly, although the trends predicted appear robust and are unlikely to be substantially influenced by these approximations, it is recommended that experimental tests be performed to assess the veracity of these computational predictions.

Recommendations

Overall, these results indicate that a steeper pelvic-belt angle, between 45°-70° degrees (as recommended in the voluntary ISO standards), results in increased occupant belt loading for all three dummy sizes considered. Clearly the intention of the steeper pelvic-belt angle is to minimize the likelihood of the pelvic belt sliding over the top of the ASIS and hence causing large abdomen soft tissue loadings and submarining. The simulations presented here captured this for the three dummy sizes (Fig. 10), but also highlighted risks of head and neck injuries due to high inertial forces for low pelvic-belt angles (see Fig. 12-16).

The goal of this work was to predict a lower pelvic-belt angle limit to reduce pelvic-belt slippage and head/neck injury risks. Risks of pelvic-belt slippage is effectively eliminated once the pelvic-belt angle is above about 45° for the 5th percentile female and 50th percentile male dummies, and above 50° for the larger 95th percentile male dummy. Results showed that dummy size was a consideration to determine this critical angle, due to the difference in the inertial forces induced during the frontal impact. This study also found that keeping both the pelvic belt and shoulder belt connected to the same floor anchor point could induce high abdomen loading at high pelvic-belt angles. Even if high pelvic-belt angles resulted in lower abdomen loadings than shallow pelvic-belt angles, head and neck injury risks were significantly increased as were occupant body excursions, chest deflection and shoulder belt loading.

Thus the recommendations on the range of allowable and preferred pelvic-belt angles in the current ISO 10542 [12] does not seem to be appropriate for any of the three dummy sizes analyzed. Most of the results obtained suggest a smaller range of pelvic-belt angle of between 45° and 60° for the majority of people (5th percentile female dummy and 50th percentile male dummy) and 50° to 60° for heavier persons (95th percentile male dummy) rather than between 45° and 75°. This is illustrated by a qualitative rating shown in Table 3, which shows the benefits and drawbacks of different pelvic-belt angles for each dummy size. While the shallow belt angles (30°-40°) greatly reduce overall occupant loading, they provide insufficient protection to the abdominal region and to the head/neck region of the occupant. The high belt angles (60°-75°) appear worse than shallow angles, providing poor abdomen protection as well as high head, neck and torso injury risks and large excursions.

Dummy Size	5th	50th	95th	5th	50th	95th	5th	50th	95th	5th	50th	95th	Good
Belt Angle	30°-40°			40°-50°			50°-60°			60°-75°			Acceptable
Horizontal Lap Belt force	g			g			g			g			Poor
Vertical Lap Belt force	g			g			g	a	a	P			
Abdomen loading	p			g	a	p	g			g	P		
Shoulder belt loading	g			g			g	g	g	a	P		
Head - Neck Injury risk	p			a	g	g	g			P			
Upper Torso Injury Risk	g			g			g			a			
Occupant Body Excusions	g			g			g			a	P		

Table 3 – Qualitative rating of pelvic-belt angle for various parameters (5th percentile female, 50th percentile male and 95th percentile male dummies)

V. CONCLUSIONS

To the best of our knowledge, this is the first study to explicitly consider the effects of belt floor anchorage point location on wheelchair occupant overall loading and kinematics for different occupant sizes and for a realistic belt configuration. The results for three different dummy sizes (5^{th} percentile female, 50^{th} percentile male and 95^{th} percentile male) indicate that occupant restraint anchor points associated with pelvic-belt angles between 60° and 75° may be detrimental for the occupant loading, kinematics and injury risk. Results also indicate that a pelvic-belt angle below 45° for the 5^{th} percentile female and 50^{th} percentile male cases and below 50° for the 95^{th} percentile male strongly increase the risks of submarining and head/neck injury risks. A pelvic-belt angle between 45° and 60° (50° and 60° for heavier person) is recommended in order to minimize occupant loading kinematics and injury risks. However, only one crash pulse was analyzed in this simulation study, and further work should focus on experimental verification of these findings and translation to a range of crash pulses.

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