Scaling approach in predicting the seatbelt loading and kinematics of vulnerable occupants: How far can we go?

Bingbing Nie\textsuperscript{a,b}, Jason L. Forman\textsuperscript{a}, Hamed Joodaki\textsuperscript{a}, Taotao Wu\textsuperscript{a}, and Richard W. Kent\textsuperscript{a}

\textsuperscript{a}University of Virginia Center for Applied Biomechanics, Charlottesville, Virginia; \textsuperscript{b}State Key Laboratory of Automotive Safety and Energy, Tsinghua University, Beijing, China

\begin{abstract}
Objective: Occupants with extreme body size and shape, such as the small female or the obese, were reported to sustain high risk of injury in motor vehicle crashes (MVCs). Dimensional scaling approaches are widely used in injury biomechanics research based on the assumption of geometrical similarity. However, its application scope has not been quantified ever since. The objective of this study is to demonstrate the valid range of scaling approaches in predicting the impact response of the occupants with focus on the vulnerable populations.

Methods: The present analysis was based on a data set consisting of 60 previously reported frontal crash tests in the same sled buck representing a typical mid-size passenger car. The tests included two categories of human surrogates: 9 postmortem human surrogates (PMHS) of different anthropometries (stature range: 147–189 cm; weight range: 27–151 kg) and 5 anthropomorphic test devices (ATDs). The impact response was considered including the restraint loads and the kinematics of multiple body segments. For each category of the human surrogates, a mid-size occupant was selected as a baseline and the impact response was scaled specifically to another subject based on either the body mass (body shape) or stature (the overall body size). To identify the valid range of the scaling approach, the scaled response was compared to the experimental results using assessment scores on the peak value, peak timing (the time when the peak value occurred), and the overall curve shape ranging from 0 (extremely poor) to 1 (perfect match). Scores of 0.7 to 0.8 and 0.8 to 1.0 indicate fair and acceptable prediction.

Results: For both ATDs and PMHS, the scaling factor derived from body mass proved an overall good predictor of the peak timing for the shoulder belt (0.868, 0.829) and the lap belt (0.858, 0.774) and for the peak value of the lap belt force (0.796, 0.869). Scaled kinematics based on body stature provided fair or acceptable prediction on the overall head/shoulder kinematics (0.741, 0.822 for the head; 0.817, 0.728 for the shoulder) regardless of the anthropometry. The scaling approach exhibited poor prediction capability on the curve shape for the restraint force (0.494 and 0.546 for the shoulder belt; 0.585 and 0.530 for the lap belt). It also cannot well predict the excursion of the pelvis and the knee.

Conclusions: The results revealed that for the peak lap belt force and the forward motion of the head and shoulder, the underlying linear relationship with body size and shape is valid over a wide anthropometric range. The chaotic nature of the dynamic response cannot be fully recovered by the assumption of the whole-body geometrical similarity, especially for the curve shape. The valid range of the scaling approach established in this study can be reasonably referenced in predicting the impact response of a given specific population with expected deviation. Application of this knowledge also includes proposing strategies for restraint configuration and providing reference for ATD and/or human body model (HBM) development for vulnerable occupants.
\end{abstract}

Introduction

Frontal motor vehicle crashes (MVCs) account for 40% of the occupant fatalities in 2012 in the United States (NHTSA 2013). Motor vehicle–related injuries and fatalities result from the complex human response within the loading environment. To understand the mechanism that links kinetics and kinematics to the trauma outcomes, the response of the occupant can be approximated by the data obtained from human surrogates, such as postmortem human subjects (PMHS) and standard anthropometric test devices (ATDs), more commonly referred to as dummies. Given the limited experimental data for any particular anthropometry, dimensional scaling approaches have been widely used in injury biomechanics research based on the assumption of geometrical similarity (Langhaar 1951).

The most conventional scaling approach uses direction-independent scaling factors from dimensional analysis of each individual subject relative to a baseline subject. The scaling factors are established by the ratios of 3 fundamental properties: the length, mass density, and Young's modulus. Then the impact response of the baseline subject, which usually represents a mid-size (50th percentile) adult male, can be scaled to a different...
body size to account for the biological variance. The mid-size adult male ATD serves as the standard for crash test evaluations and has been scaled to different sizes, including the 5th percentile female and 95th percentile male, to reflect the occupant population (Eppinger 1989; Mertz 1984; Mertz et al. 1989). Scaling approach remains an analysis tool commonly used in the injury biomechanics research (Kent et al. 2001, 2004). It has been recognized that the predictive ability of the scaling approach is limited due to the inability to completely account for the non-linearity in the occupant–restraint interactions (Forman et al. 2006, 2008). However, limited attempts have been made to quantify the application scope.

Compared to the mid-size occupant, published epidemiology reported that occupants with extreme body size and shape, such as the small female or the obese, tended to have a high risk of injury in automobile collisions (Bose et al. 2011; Bhattie et al. 2016). Biomechanical factors of the occupants, including age, sex, body shape, and size, all contribute to the dynamic response and the resultant injury patterns. Specifically, age and sex significantly affected the tolerance level of MVC injuries (Carter et al. 2014; Kent and Patrie 2005; Zhou et al. 1996); body mass and the distribution of body mass (body shape) and stature (the overall body size) exhibited significant influence on the occupant trajectories during the collisions (Forman et al. 2006). Despite the impressive increasing safety level of current vehicle design, the restraint systems were developed and evaluated mainly relying on particular anthropometries that limited the ability of the resultant protection in covering a broader range of occupant properties (Segui-Gomez et al. 2007). To account for the increasing vulnerability associated with particular anthropometries, it is critical to assess the ability of the scaling approach in predicting the dynamic responses of different of occupant groups to reflect the real-world population variance.

With the use of both PMHS and ATDs over a wide range of anthropometry, a series of sled tests have been performed representing frontal impacts of a typical mid-size passenger car (Forman et al. 2008, Forman, Lopez-Valdes, Lessley, Kindig, et al. 2009; Forman, Lopez-Valdes, Lessley, et al. 2009; Lopez-Valdes et al. 2009). An obese ATD, built up via recent efforts to capture the occupant–restraint interaction for obese occupants, was also included (Joodaki et al. 2015). The experimental data set provides a unique opportunity to investigate the prediction capability of scaling approaches under different impact configurations. The objective of this study is to demonstrate the valid range of scaling approaches in predicting the impact kinematics of the occupants with particular focus on the vulnerable populations. Two restraint types were included: a 3-point standard belt (SB) and a force-limiting belt (FLB). Two impact severities were considered with the initial velocities of 48 and 29 km/h.

Materials and methods

For the two categories of human surrogates, the PMHS and the ATDs, a mid-size subject was selected as the baseline. The restraint loads and the forward kinematics of the baseline subject were scaled specifically to another subject based on anthropometric difference as a prediction of the impact responses under the same loading condition. Following this, the scaled responses were compared to the experimental results using a quantitative assessment to establish the valid range (Figure 1).

![Figure 1. Overview of the methodology to evaluation the scaling approach on predicting occupant kinematics.](image)

An overview of the experimental data set, a detailed description of the data scaling, and the evaluation algorithm are presented in the following subsections.

Overview of the experimental data set

The experimental data set consisted of 60 previously reported frontal crash tests in the same sled buck representing a typical mid-size passenger car (Forman et al. 2008; Forman, Lopez-Valdes, Lessley, Kindig, et al. 2009; Forman, Lopez-Valdes, Lessley, et al. 2009; Lopez-Valdes et al. 2009), including 10 tests with PMHS and 50 tests with ATDs (Figure 2, Table 1). The restraint system included either a 3-point SB or a pretensioning, 2-stage progressive FLB. The occupants in the test matrix, referred to either as an ATD or a PMHS, included 5 different ATDs; that is, the Hybrid III 6-year-old (H3 6YO) in a booster seat, the Hybrid III 5th percentile female (H3 AF05), the Hybrid III 50th percentile male (H3 AM50), the THOR-NT, an obese ATD, and 9 PMHS covering a wide anthropometric range (stature 147–189 cm; body mass 27.0–151.0 kg). The obese ATD was developed by adding a flesh jacket representing the superficial tissue to THOR-NT (Joodaki et al. 2015). To facilitate a reasonable comparison, the weighted flesh jacket was designed to replicate both the external geometry and overall body mass specifically on the anthropometry of one of the tested PMHS (subject 404). CT scans confirmed that the internal skeletal dimensions of subject 404 were similar other mid-size PMHS (Forman, Lopez-Valdes, Lessley, Kindig, et al. 2009) and therefore the majority of the difference in body mass can be explained by the increased superficial tissue.
The ATDs were tested in 2 impact severities with the initial velocities of 48 and 29 km/h, except that the obese ATD was tested only with the use of FLB under 48 km/h. All of the PMHS were tested only under 48 km/h and no test at 29 km/h was performed. The small PMHS (subject 437) was tested repeatedly in both belt systems (tests 1388, 1385) to maximize the information obtained from this rare subject. Trapezoidal sled pulses were used for both impact severities, representing a typical vehicle deceleration experienced by a mid-sized sedan (Figure A1, see online supplement). For the tested ATDs, the front seats were installed in the mid-track position, representing a typical position as in a mid-size sedan. One exception was for the obese dummy, in which the front seats were removed to observe the occupant–restraint interactions without the highly potential interactions with front seat belt. For all tested PMHS, the front seats were removed to avoid potential confounding effects of interaction between the knees and the front seatback.

The impact response included the restraint loads and the forward kinematics of multiple body segments. Tension force of the seat belt was measured by belt tension gauges at 2 locations. The upper shoulder belt force, \( F_{\text{US}} \), was measured by belt tension gauges between the occupant shoulder and the anchor location. The lap belt force, \( F_{\text{Lap}} \), was measured in the outboard lap belt between the right side of the occupant and the belt anchor. The overall occupant kinematics were recorded with off-board high-speed video (1,000 fps), providing the trajectories of relevant anatomical targets with respect to the buck reference frame. The markers were externally mounted to the head, the shoulder, the pelvis, and the knee. The thorax was not available due to the blocking view by the arms and the confounding effects of rotation in the acceleration signals. All of the markers were tracked throughout the impact to quantify the excursion, which was defined as the forward displacement relative to the sled buck in the sagittal plane of the occupant.

### Data scaling

Scaling factors of the impact responses were calculated based on 3 fundamental dimensional ratios: the length ratio, \( \lambda_L \), the mass density ratio, \( \lambda_\rho \), and the material property (Young’s modulus) ratio, \( \lambda_E \) (Eppinger et al. 1984). All 3 ratios were recorded as the ratio of the entity on the scaled subject relative to that of the baseline. Equal mass densities were assumed across human subjects (\( \lambda_\rho = 1 \); Melvin 1995; Mertz et al. 1989). The same Young’s modulus was assumed across the adult subjects. This decision was made given the limited range of age for the tested PMHS (59 ± 7.5 years) and existing studies on structural stiffness of the human body, which revealed an approximate constant modulus after 20 years old (Irwin and Mertz 1997; Ivarsson et al. 2004; Kleinberger et al. 1998; Snyder 1977). The scaling ratio of the Young’s modulus for the 6YO was adopted from bending tests of the cranial bone (Irwin and Mertz 1997).

The Hybrid III family of dummies was developed based on the morphology studied corresponding to the normal-weight anthropometry (Melvin 1995; Mertz 1993). Therefore, the H3 AM50 was selected as the baseline and the response in each loading condition or restraint pattern was scaled to the H3 6YO and H3 AF05. Scaling from the H3 AM50 to THOR-NT was also performed to highlight the possible underlying response difference due to the design changes between the 2 different mid-size dummies. Prediction of the responses of the obese ATD was scaled from THOR-NT because THOR-NT was used as the basis to build up the obese dummy. For the PMHS, 2 subjects with approximate 50th percentile stature and mass—that is, subject 367 for the SB and subject 429 for the FLB (Table 1)—were chosen as the baseline for the SB and FLB, respectively. Given the fact that the occupants with great body mass sustained characteristic different sets of body motions, especially for the hip and pelvis (Kent et al. 2010), we divided the tested subjects into 3 groups based on body mass index (BMI) for the screening purpose of weight category: (1) underweight (BMI \( \leq 18.5 \text{ kg/m}^2 \), \( n = 2 \)); (2) normal (BMI between 18.5 and 24.9 kg/m\(^2\), \( n = 5 \)); (3) obese (BMI \( \geq 30 \text{ kg/m}^2 \), \( n = 3 \)); \( n \) indicates the number of tests.

To get the scaled response of the test subjects, let \( f_{\text{scaled}}(t) \) denote the time histories of the kinetic and kinematic data in the baseline subject, where \( t \) represents time and was used as the independent variable. The general form of the scaled impact response specifically to another subject is

\[
f_{\text{scaled}} = \lambda_R \cdot f_{\theta} (\lambda_L \cdot t),
\]

where \( f_{\text{scaled}} \) is the scaled response derived for a given subject; \( \lambda_R \) is the scaling factor that was applied to the kinetic or kinematic
responses, \( f_b \); and \( \lambda_1 \) is the scaling factor that was applied to time, \( t \).

For the restraint force, the scaling factor was based on the body mass (body shape) compared to the baseline subject. This decision was made because greater occupant mass was associated with increasing kinetic energy to be restrained by the seatbelt and led to higher belt force (Nie, Poulard, et al. 2016; Viano et al. 2008). The characteristic ratio based on body mass, \( \lambda_{L,M} \), was determined as

\[
\lambda_{L,M} = \left( \frac{m}{m_b} \right)^{\frac{1}{2}},
\]

where \( m \) and \( m_b \) are the body mass of the target and the baseline subject, respectively.

Occupant motion is reported to be more relevant to the initial body size (Kerrigan et al. 2005). Because multiple body segments were taken into account and the seated head-to-buttock height measured among the subjects exhibited an approximate linear correlation with stature, the kinematic data were scaled by comparison to the stature of the baseline subject. The length ratio based on body size, \( \lambda_{L,S} \), was determined as

\[
\lambda_{L,S} = \frac{L}{L_b},
\]

where \( L \) and \( L_b \) are the overall body size of the target and the baseline subject.

Following the principles of dimensional analysis, scaling factors for the kinetic and kinematic response in magnitude and in time can be determined in terms of \( \lambda_L \) and \( \lambda_E \) (Table 2). One exception is that the shoulder belt force with the use of FLB was dictated by the design of the force limiter; therefore, it was not included in the scaling.

| Table 1. Baseline subject and fundamental ratios for other subjects. |
|---|
| (a) ATDs | Number of tests |
| ATDs | Mass (kg) | Stature (cm) | BMI | \( \lambda_{L,M} \) | \( \lambda_{L,S} \) | \( \lambda_E \) | SB (29 km/h) | SB (48 km/h) | FLB (29 km/h) | FLB (48 km/h) |
| Baseline: H3 | M | 78 | 178 | 25 | | | 3 | 3 | 3 | 3 |
| Baseline: THOR-N | M | 78 | 178 | 25 | 1.000 | 1.000 | 3 | 3 | 3 | 3 |
| H3 6YO | — | 23 | 114 | 18 | 0.669 | 0.641 | 0.667 | 3 | 3 | 3 | 3 |
| H3 AF5 | F | 49 | 150 | 22 | 0.856 | 0.842 | 1.000 | 3 | 3 | 3 | 3 |
| THOR-N | M | 78 | 178 | 25 | 1.000 | 1.000 | 3 | 3 | 3 | 3 |
| Obese ATD | M | 124 | 175 | 40.4 | 1.166 | 1.000 | 1.000 | — | — | — | 2 |
| PMHS (subject no.) | Age/gender | Mass (kg) | Stature (cm) | BMI | Rib fx | Chest | Max AIS | \( \lambda_{L,M} \) | \( \lambda_{L,S} \) | \( \lambda_E \) | Test no. | Test setup |
| Baseline: 367 | 57/M | 59 | 179 | 18 | 13 | 4 | 3 | 0.773 | 0.821 | 1.000 | 1264 | 5B (48 km/h) |
| 437 | 54/F | 27 | 147 | 13 | 12 | 3 | 1 | 0.976 | 0.978 | 1.000 | 1264 | 5B (48 km/h) |
| 394 | 57/F | 109 | 165 | 40 | 29 | 4 | 1 | 1.227 | 0.922 | 1.000 | 1263 | 5B (48 km/h) |
| Baseline: 429 | 67/M | 69 | 175 | 23 | 12 | 4 | 3 | 0.733 | 0.840 | 1.000 | 1838 | |
| 437* | 54/F | 27 | 147 | 13 | 12 | 3 | 1 | 0.990 | 0.977 | 1.000 | 1264 | |
| 444 | 69/M | 67 | 171 | 23 | 12 | 4 | 1 | 1.014 | 1.046 | 1.000 | 1387 | 5B (48 km/h) |
| 457 | 72/M | 72 | 183 | 21 | 17 | 4 | 1 | 1.216 | 1.080 | 1.000 | 1333 | |
| 400 | 53/M | 151 | 182 | 46 | 7 | 3 | 1.298 | 1.040 | 1335 | |

*The H3 6YO was tested in a booster seat.

**Subject 437 was tested in a booster seat.

General: Rib fx indicates the number of rib fractures identified posttest via autopsy.

“Chest Max AIS” was based on the AIS 2005 code. The injury results were previously reported in several separate studies (Forman et al. 2008; Forman, Lopez-Valdes, Lessley, Kindig, et al. 2009; Forman, Lopez-Valdes, Lessley, et al. 2009; Lopez-Valdes et al. 2009).

| Table 2. Calculation of the scaling factors for the kinetic and kinematic response. |
|---|
| Response | Scaling factor |
| Kinetics | Force \( \lambda_K = \lambda_{L,M}^2 \cdot \lambda_E \) |
| Time | \( \lambda_T = \lambda_{L,S}^{-1/2} \) |
| Kinematics | Excursion \( \lambda_K = \lambda_{L,M} \cdot \lambda_E \) |
| Time | \( \lambda_T = \lambda_{L,S}^{-1} \) |

Evaluation of the scaled responses

The scaled responses were quantitatively assessed by comparison with the experimental results. Three criteria were defined as below with considerations on the peak value, peak timing (the time when the peak value occurred), and the overall curve shape; that is, \( R_{\text{peak}} \), \( R_{\text{time}} \), and the weighted integrated factor (WIFac), \( R_{\text{WIFac}} \) (Hovenga et al. 2005; Jacob et al. 2000).

\[
R_{\text{peak}} = 1 - \frac{|f_{\text{scaled}}(t)| - |f_{\text{test}}(t)|}{|f_{\text{test}}(t)|} \quad (4)
\]

\[
R_{\text{time}} = 1 - \frac{t_{\text{scaled}} - t_{\text{test}}}{0.4 \cdot t_{\text{eval}}} \quad (5)
\]

\[
R_{\text{WIFac}} = 1 - \frac{\int \max \left( |f_{\text{test}}(t)|, |f_{\text{scaled}}(t)| \right) dt}{\int \max \left( |f_{\text{test}}(t)|, |f_{\text{scaled}}(t)| \right) dt} \quad (6)
\]
where $f_{test}(t)$ and $f_{scaled}(t)$ are the time histories of the test results and the scaled results; the subscript $pt$ indicates the peak value and $t_{eval}$ indicates the evaluation period (160 ms was used for the frontal impact). In the calculation of $R_{time}$, the reference period was considered as $0.4 \cdot t_{eval}$ as an estimation corresponding to an intuitive judgment in frontal crashes (Jacob et al. 2000). The overall evaluation score, $R_{tot}$, was calculated as the root mean square addition with equal weights, $w_i$ of the 3 criteria (Hovenga et al. 2005):

$$R_{tot} = 1 - \sqrt{\frac{\sum (w_i \cdot (1 - R^2_i))}{\sum w_i}},$$

where $R$ represents $R_{peak}$, $R_{time}$, and $R_{WFac}$.

Each criterion returned a score ranging between 0 (extremely poor) and 1 (perfect match). The scores of 0.0 to 0.5, 0.5 to 0.7, 0.7 to 0.8, and 0.8 to 1.0 are deemed to indicate unacceptable, poor, fair, and acceptable prediction of the scaled response, respectively (Jacob et al. 2000).

**Results**

**Evaluation of the scaled responses**

Anthropometric information and fundamental dimensional ratios of different subjects are provided in Table 1. The tested subjects showed a general positive correlation between body stature and body mass, which was in accordance with the statistical population trend as reported in existing biological studies (Benn 1971; Burton 2007). The scaling factors were calculated using Equations (1)–(3) and then applied to the impact responses of the baseline subject. The evaluation score for each test was provided in Table A1 (see online supplement). Based on the evaluation score, prediction capability of the scaled impact responses for different occupant groups was plotted in Figure 3. For each occupant group, the average scores of the tested subject were provided. The green area shown in the background corresponds to the valid range satisfying an evaluation score over 0.8. For ATDs, the scaled kinetic and kinematic responses of the H3 AF05 and THOR-NT exhibited generally acceptable prediction level, except for the curve shape of the restraint force of H3 AF05. For the obese ATD, the scaled responses are acceptable for peak value and timing of the lap belt and excursion of the head and shoulder. For the PMHS, the prediction was acceptable for most responses of the normal subjects, except for knee excursion. The prediction level on the underweight subject was only acceptable for the excursion of the head and knee. In the obese group, the scaled response on the excursion of the head and knee agreed with the experiments, though most other responses cannot be inferred from the mid-size subject.

Representative comparison of the scaled responses and the experimental results are provided in Figures A2 and A3 (see online supplement). The scaling factor applied to time, $\lambda_t$, led to a phase shift from the baseline response. For the small subjects, the phase difference in the kinematic curves was overpredicted in the scaled response (Figure A2). Most responses of the H3 6YO cannot be well inferred from the mid-size ATD in use.
of FLB. In contrast, the characteristic ratio based on either body mass ($\lambda_{L,M}$) or body stature ($\lambda_{L,S}$) was close to 1 among the normal subjects (Table 1). As a result, the scaling approach led to very minor change on the baseline responses (Figure A3). Excursion time histories of multiple body segments were close for the normal PMHS subjects. For the mid-size dummies, the H3 AM50 and the THOR-NT, the structural difference led to different kinematics of the head, the pelvis, and the knee.

**Summary of the evaluation score**

Overall, for both ATDs and PMHS, the scaling approach provided fair or acceptable prediction capability on timing of kinetic response (0.868, 0.829 for shoulder belt; 0.858 and 0.774 for lap belt) and the peak value of the lap belt force (0.796, 0.869). The overall evaluation score was 0.741, 0.822 for the head and 0.817, 0.728 for the shoulder, indicating a fair or acceptable prediction of the scaled responses. The scaling approach exhibited poor prediction capability on the curve shape for the restraint force (0.494, 0.546 for shoulder belt; 0.585 and 0.530 for lap belt) and for the pelvis excursion (0.573, 0.663; Table 3).

**Discussion**

**Valid range of the scaled responses**

The present study demonstrated the valid range of scaling approaches in predicting the impact kinematics of the occupants with a focus on the vulnerable populations in MVCs. Dimensional scaling approaches are widely used in injury biomechanics research based on the assumption of geometrical similarity. Though it has been recognized that the predictive ability of the scaling approach is limited due to the lack of considerations on the specific structural change (Nie, Kim, et al. 2016), it is necessary to establish the application scope of such a straightforward and easy-to-perform approach in predicting the impact responses of any particular anthropometry. All 60 frontal sled tests included in the experimental data set were set up with the same sled buck and deck-mounted retractor. This eliminated potential complexities from the simulated vehicle environment and provided the primary data source for evaluating the scaled response and establishing the valid range of the scaling approach.

The fundamental assumption in a scaling approach is the global geometric similarity between the baseline and the scaled subject. The consequent valid range is limited when the assumption does not hold; that is, when the target subject is “dissimilar” to the baseline in anthropometry, as in the small or obese occupant. It is therefore reasonable to infer that the application scope of scaling will follow the same trend in other crash modes, such as oblique or lateral impacts. The present results revealed that for the peak force of the lap belt and the head/shoulder kinematics, the underlying similarity of the impact responses was valid over a wide anthropometric range (Figure 3). Based on a normal subject, the body mass for a given anthropometry can be used as a good predictor on the peak value and the occurrence of the lap belt force for ATDs of difference sizes or PMHS in a range of 27 to 151 kg. Previous field and computational studies reported that occupant stature exhibited a significant influence on the risk of head and thoracic injury (Bose et al. 2010; Miller 1995; Segui-Gomez et al. 2009). From a kinematic point of view, the stature proved a good predictor on the forward motion of the head and the shoulder in well-controlled laboratory impacts.

It should be noted that although the Hybrid III family of dummies was developed based on the morphology corresponding to the normal-weight anthropometry, the THOR-NT held some different internal structures. Therefore, the scaling from the H3 AM50 to THOR-NT did not imply an anthropometry-based response change but suggested a possible underlying response difference due to the design changes (Figure 3). Being developed as a successor to the HIII AM50 dummy, the THOR-NT has a similar body stature representing a mid-size male. This explained the fact that the peak belt forces sustained by the 2 dummies were close, resulting in a high evaluation score of $R_{\text{peak}}$ and $R_{\text{time}}$ for the shoulder and the lap belt (Figure 3; Table A1). On the other hand, with a less coupled thoracic structure and more anteriorly protruding clavicle (Forman et al. 2008; Xu et al. 2000), the THOR-NT sustained a different curve shape of the shoulder belt force (Figure A3). Additionally, the higher hip-to-knee length of the THOR-NT led to a closer position of the knee to the front seats in the experimental setup. As a result, the THOR-NT knees experienced interaction with the front seats during the tests, whereas the HIII dummies did not. This led to relatively low value of $R_{\text{WIFac}}$ and $R_{\text{peak}}$ when comparing the kinematics of the pelvis and the knee (Figure A3).

Overall, given the chaotic nature of the anatomical and physiological variability, it was observed that the whole curve shape cannot be fully recovered by the assumption of geometrical similarity in the crash test, for either body mass or body stature. One possible confounding factor is the injury occurrence, which was influenced highly by subject characteristics. The most common injury patterns were rib fractures during the PMHS tests, with the number of rib fractures varied greatly among subjects (Table 1). Although rib fractures were believed to have a limited effect on the whole-body kinematics (Duma et al. 2006; Kemper et al. 2011; Motozawa et al. 2015), it could potentially contribute to the nonlinearity in the kinetic response. For example, the obese subject in test 1263 (subject 394) sustained 29 rib fractures, which may have limited the buildup of the restraint

---

**Table 3. Summary of the evaluation score on the scaled responses.** The value was calculated as the root mean square addition according to Equation (6) for the ATDs and the PMHS, respectively.

| Response   | ATD            | PMHS           |
|------------|----------------|----------------|
|            | $R_{\text{WIFac}}$ | $R_{\text{peak}}$ | $R_{\text{time}}$ | $R_{\text{tot}}$ | $R_{\text{WIFac}}$ | $R_{\text{peak}}$ | $R_{\text{time}}$ | $R_{\text{tot}}$ |
| Shoulder   | 0.494          | 0.703          | 0.868          | 0.649          | 0.546          | 0.683          | 0.829          | 0.665          |
| Lap belt   | 0.585          | 0.796          | 0.858          | 0.716          | 0.530          | 0.869          | 0.774          | 0.683          |
| Head       | 0.678          | 0.803          | 0.768          | 0.741          | 0.847          | 0.892          | 0.765          | 0.822          |
| Shoulder   | 0.775          | 0.815          | 0.898          | 0.817          | 0.800          | 0.772          | 0.698          | 0.728          |
| Pelvis     | 0.573          | 0.680          | 0.690          | 0.640          | 0.663          | 0.668          | 0.646          | 0.656          |
| Knee       | 0.562          | 0.681          | 0.760          | 0.653          | 0.707          | 0.801          | 0.819          | 0.762          |
force form the upper shoulder belt. This led to the unacceptable results when using linear scaling factors to predict the belt forces based on anthropometric measures (Figure 3). For the small subjects, as described above, the overall phase shift resulting from time scaling was likely to overpredict the response change in the head and the shoulder and resulted in a low $R_{\text{time}}$ and the associated $R_{\text{opt}}$. Given the scarcity of small-size subjects, much more awaits discovery before making an acceptable conclusion. Additionally, existing studies have also demonstrated that higher body mass led to significantly increasing forward pelvis motion before the sufficient belt restraint can be achieved (Kent et al. 2010; Viano et al. 2008). Likewise, as observed in our evaluation, the small and obese subjects sustained an altered restraint response and body kinematics during a crash that cannot be deduced by scaling that of a normal subject.

Limitations

Anthropometric variability across the population and the characteristic dynamic response cannot be fully recovered by the overall body size and shape. The confounding factors, including but not limited to body mass distribution, interaction among body segments, variabilities in the structural and material properties, were not taken into account in the present scaling approach. For example, obesity is usually associated with more adipose tissue over the anterior pelvis instead of a shape change at the whole-body level. This necessitates further considerations on the geometry or mass distribution of specific body regions when drawing analogies between different human surrogates.

Secondly, the characteristic length ratios were calculated based on either body mass or stature; however, this also incorporated the coupled effects of the 2 anthropometric measurements, because taller occupants tend to have higher body mass. As preliminary efforts, we divided the occupants into different groups using BMI as an adjusted measure. While presenting the results, to which level the influence on the prediction from both body size and shape was mixed remains unknown.

Thirdly, due to the sample sizes, the conclusions drawn here are associated with the specific loading condition and selected responses. The obese ATD was tested only in 48 km/h frontal impacts with a FLB; no Hybrid III 95th percentile male dummy were tested. For the PMHS tests, only one small subject was available so far and all subjects covered a limited age scope (53–69 years old). The frontal impact is representative of a typical and common impact configuration but more complex crash conditions were not considered. In case of more realistic vehicle environments, more complicated impact pulse, other restraint parts—for example, airbags, knee bolster, or the front seats—will probably contribute to more complex interaction with the occupants. For example, airbags will change the distribution of restraint force over the thorax, limiting the head and shoulder excursion to some level (Forman et al. 2006). With the absence of front seats, the potential contact with the knee will also decrease the pelvis motion and affect the loads sustained from the lap belt (Forman et al. 2008). Although it remains unknown to which level the application of scaling will be affected when multiple restraint parts are involved, a further reduced valid range can be foreseen due to the increasing nonlinearity. Subsequent studies are recommended to investigate the possible effects from more complex interaction of occupants and contemporary restraint design, either experimentally or computationally.

Future work and applications

Future work involves more comprehensive examination of the formulation of scaling approaches. Although the overall body mass and stature were believed to be 2 indicators on occupant anthropometry at the global level, it should be noted that there can be better scaling laws than those of the current study. For example, instead of establishing scaling factors under the same assumption across different subjects, it is possible to investigate the set of optimal scaling factors for the individual subject via optimization of the baseline response. The subsequent comparison of the present scaled response to the best possible match can provide deeper insight into the assessment of the scaling approach. Along with advanced measurement techniques (Gayzik et al. 2011), it would be beneficial to use the segmented geometry of multiple body regions or skeletal structures in deducing scaling factors. Subsequence studies include extending the methods to cover the risk of injury and identifying the causal mechanics that link the occupant kinematics to the trauma outcomes.

To the best of the authors’ knowledge, this study presents the first preliminary quantitative assessment of a scaling approach in predicting occupant kinematics in frontal crash mode. The valid range of the scaling approach, as established here, can be reasonably referenced in predicting the impact response of a given specific population with expected deviation. As a result, priorities in the research efforts on occupant safety can be identified to make full use of available resources. For example, the dynamic response of the small female ATDs can be well predicted via scaling, whereas more investigations on the kinematics of child dummies or obese subjects need to be taken into account given their dissimilarity to the mid-size. The application scope of scaling provides implications on promoting effective restraint configurations for covering a broader range of populations. To reduce vulnerability associated with particular anthropometries, such as the small female and obese occupants, the vehicular countermeasure design should reply on experimental and/or computational analysis to investigate their specific interactions with the restraint system and interior parts. Application of this knowledge also includes contributing data that can be used to refine and evaluate ATDs and/or human body models to replicate kinematics and assess injury risk of occupants.

Funding

The experiments reviewed in this study were supported by the NHTSA, Autoliv Research, and Humanetics Innovative Solutions Corporation. This work was sponsored by the State Key Laboratory of Automotive Safety and Energy, Tsinghua University under Project No. KF16172. The opinions expressed here are solely those of the authors.

References

Benn RT. Some mathematical properties of weight-for-height indices used as measures of adiposity. Br J Prev Soc Med. 1971;25:42–50.
Bhatti JA, Nathens AB, Redelmeier DA. Driver’s obesity and road crashes in the United States. Traffic Inj Prev. 2016 in press. doi:10.1080/15389588.2015.1134793

Burton RF. Why is the body mass index calculated as mass/height², not as mass/height²? Ann Hum Biol. 2007;34:656–663.

Bose D, Crandall JR, Untaroiu CD, Maslen EH. Influence of pre-collision occupant parameters on injury outcome in a frontal collision. Accid Anal Prev. 2010;42:1398–1407.

Bose D, Segui-Gomez M, Crandall JR. Vulnerability of female drivers involved in motor vehicle crashes: an analysis of US population at risk. Am J Public Health. 2011;101:2368–2373.

Carter PM, Flannagan CA, Reed MP, Cunningham RM, Rupp JD. Comparing the effects of age, BMI and gender on severe injury (AIS 3+) in motor-vehicle crashes. Accid Anal Prev. 2014;72:146–160.

Duma S, Stitzel J, Kemper A, et al. Acquiring non-censored rib fracture data during dynamic belt loading. Biomed Sci Instrum. 2006;42:148–153.

Eppinger RH. On the development of a deformation measurement system and its application toward developing mechanically based injury indices. Paper presented at: Stapp Car Crash Conference, 1989.

Eppinger RH, Marcus JH, Morgan RM. Development of Dummy and Injury Index for NHTSA’s Thoracic Side Impact Protection Research Program. Warrendale, PA: Society of Automotive Engineers; 1984; SAE No. 840885.

Forman J, Lessley D, Kent R, Bostrom O, Pipkorn B. Whole-body kinematic and dynamic response of restrained PMHS in frontal sled tests. Stapp Car Crash J. 2006;50:299–336.

Forman JL, Lopez-Valdes FJ, Lessley D, Kindig M, Kent RW, Bostrom O. The effect of obesity on the restraint of automobile occupants. Ann Adv Automot Med. 2009;53:25–40.

Forman JL, Lopez-Valdes FJ, Lessley D, et al. Rear seat occupant safety: an investigation of a progressive force-limiting, pretensioning 3-point belt system using adult PMHS in frontal sled tests. Stapp Car Crash J. 2009;53:49–74.

Forman JL, Michaelson J, Kent RW, Kuppa S, Bostrom O. Occupant restraint in the rear seat: ATD responses to standard and pretensioning, force-limiting belt restraints. Ann Adv Automot Med. 2008;52:141–154.

Gayzik FS, Moreno DP, Geer CP, Wuerther SD, Martin RS, Stitzel JD. Development of a full body CAD dataset for computational modeling: a multi-modality approach. Ann Biomed Eng. 2011;39:2568–2583.

Hovenga PE, Spitz HH, Uijldert M, Dalenoort AM. Improved Prediction of Hybrid-III Injury Values using Advanced Multibody Techniques and Objective Rating. Warrendale, PA: Society of Automotive Engineers; 2005; SAE No. 2005-01-1307.

Irwin A, Mertz HJ. Biomechanical basis for the CRABI and Hybrid III child dummies. In: Proceedings of the 41st Stapp Car Crash Conference. Warrendale, PA: Society of Automotive Engineers; 1997:261–272.

Ivarsson B, Crandall J, Longhitanio D, Okamoto M. Lateral Injury Criteria for the 6-Year-Old Pedestrian—Part II Criteria for the Upper and Lower Extremities. Warrendale, PA: Society of Automotive Engineers; 2004. SAE No. 2001-01-1755.

Jacobs C, Charras F, Trosseille X, Hamon J, Pajon M, Lecoq JY. Mathematical models integral rating. Int J Crashworthiness. 2000;5:417–432.

Joodaki H, Forman J, Forghani A, et al. Comparison of kinematic behaviour of a first generation obese dummy and obese PMHS in frontal sled tests. Paper presented at: International Research Council on the Biomechanics of Injury conference (IRCOBI); September 9–11, 2015; Lyon, France.

Kepler AR, Kennedy EA, Mcnally C, Manooqian SJ, Stitzel JD, Duma SM. Reducing chest injuries in automobile collisions: rib fracture timing and implications for thoracic injury criteria. Ann Biomed Eng. 2011;39:2141–2151.

Kent RW, Crandall JR, Bolton J, Prasad P, Nusholtz G, Mertz H. The influence of superficial soft tissues and restraint condition on thoracic skeletal injury prediction. Stapp Car Crash J. 2001;45:183–204.

Kent RW, Forman JL, Bostrom O. Is there really a "cushion effect"? A biomechanical investigation of crash injury mechanisms in the obese. Obesity (Silver Spring). 2010;18:749–753.

Kent RW, Lessley D, Shrewed C. Thoracic response to dynamic, non-impact loading from a hub, distributed belt, diagonal belt, and double diagonal belts. Stapp Car Crash J. 2004;48:1–25.

Kent RW, Patrie J. Chest deflection tolerance to blunt anterior loading is sensitive to age but not load distribution. Forensic Sci Int. 2005;149:121–128.

Kerns D, Kam C, Drinkwater C, et al. Kinematic comparison of the Polar-II and PMHS in pedestrian impact tests with a sport-utility vehicle. In: Proceedings of the International Research Council on the Biomechanics of Injury Conference. Prague, Czech Republic: International Research Council on Biomechanics of Injury; 2005:159–174.

Kleinerber M, Yoganandan N, Kumaresan S. Biomechanical considerations for child occupant protection. Annu Proc Adv Automot Med. 1998;42:115–136.

Langhaar HL. Dimensional Analysis and Theory of Models. New York, NY: Wiley; 1951.

Lopez-Valdes FJ, Forman JL, Kent RW, Bostrom O, Segui-Gomez M. A comparison between a child-size PMHS and the Hybrid III 6 YO in a sled frontal impact. Ann Adv Automot Med. 2009;53:237–246.

Melvin JW. Injury Assessment Reference Values for the CRABI 6-Month Infant Dummy in a Rear-Facing Infant Restraint with Air Bag Deployment. Warrendale, PA: Society of Automotive Engineers; 1995; SAE No. 950872.

Mertz HJ. A procedure for normalizing impact response data. In: Proceedings of the 28th Stapp Car Crash Conference. Warrendale, PA: Society of Automotive Engineers; 1984:1159–1166.

Mertz HJ. Anthropometric test devices. In: Nahum AM, Melvin JW, eds. Accidental Injury: Biomechanics and Prevention. New York, NY: Springer-Verlag. 1993:85–87.

Mertz HJ, Irwin A, Melvin J, Stanaker R, Beebe M. Size, Weight and Biomechanical Impact Response Requirements for Adult Size Small Female and Large Male Dummy. Warrendale, PA: Society of Automotive Engineers; 1989. SAE No. 890756.

Miller H. Injury reduction with smart restraint systems. Ann Adv Automot Med. 1995;39:527–541.

Motozawa Y, Okamoto M, Mori F. Comparison of whole body kinematics between fracture and non-fracture finite element human body models during side impact. Paper presented at: International Research Council on the Biomechanics of Injury Conference (IRCOBI); 2015; Lyon, France.

NHTSA. Traffic Safety Facts 2012—A Compilation of Motor Vehicle Crash Data from the Fatality Analysis Reporting System and the General Estimates System. Washington, DC: US Department of Transportation, Author; 2013.

Nie B, Kim T, Wang Y, Bollapragada V, Daniel T, Crandall JR. Comparison of two scaling approaches for the development of biomechanical multi-body human models. Multibody Syst Dyn. 2016 in press. doi:10.1007/s11044-016-9502-2

Nie B, Poulard D, Subit D, Donlon JP, Forman JL, Kent RW. Experimental investigation of the effect of occupant characteristics on contemporary seatbelt payout behavior in frontal impacts. Traffic Inj Prev. 2016;17:374–380.

Segui-Gomez M, Lopez-Valdes FJ, Crandall JR. Characterizing the distribution of injury and injury severity for belted front-seat occupants involved in frontal crashes. In: Proceedings of the International Research Council on the Biomechanics of Injury Conference (IRCOBI). York, UK; 2009:155–168.

Segui-Gomez M, Lopez-Valdes FJ, Frampton R. An evaluation of the EuroNCAP crash test safety ratings in the real world. Ann Adv Automot Med. 2007;51:281–298.

Snyder RG. Anthropometry of Infants, Children, and Youths to Age 18 for Product Safety Design. Ann Arbor, MI: University of Michigan; 1977.

Viano DC, Parenteau CS, Edwards ML. Crash injury risks for obese occupants using a matched-pair analysis. Traffic Inj Prev. 2008;9:59–64.

Xu L, Jensen J, Byrnes K, et al. Comparative Performance Evaluation of THOR and Hybrid III. Warrendale, PA: Society of Automotive Engineers; 2000. SAE No. 2000-01-0161.

Zhou Q, Rouhana SW, Melvin JW. Age effects on thoracic injury tolerance. Proc Stapp Car Crash Conf. 1996;40:137–148.