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Frontal impact response of a virtual low percentile six years old human thorax developed by automatic down-scaling

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Abstract

Traffic accidents cause one of the highest numbers of severe injuries in the whole population spectrum. The numbers of deaths and seriously injured citizens prove that traffic accidents and their consequences are still a serious problem to be solved. The paper contributes to the field of vehicle safety technology with a virtual approach. Exploitation of the previously developed scaling algorithm enables the creation of a specific anthropometric model based on a validated reference model. The aim of the paper is to prove the biofidelity of the small percentile six years old virtual human model developed by automatic down-scaling in a frontal impact. For the automatically developed six years old virtual specific anthropometric model, the Kroell impact test is simulated and the results are compared to the experimental data. The chosen approach shows good correspondence of the scaled model performance to the experimental corridors.

Keywords: virtual human model, scaling, the Kroell test, frontal response

1. Introduction

More than 80 people die on European roads each day and many others get seriously injured. The estimated economic loss from road traffic injuries accounts for about 2 % of gross domestic product in the EU, while the human suffering in terms of physical and psychological consequences of death, injury and disability is hard to quantify [5, 13]. This situation is unacceptable both from an ethical perspective and from an economic point of view.

To reduce the effects of traffic accidents, well optimized active and passive safety systems are developed. The tests define hardware dummies used for particular impact cases and injury criteria to be assessed. However, the dummies are not multi-purpose and they usually represent only a few anthropometric percentiles [7, 8, 11, 13]. Generally the current state of children models are very low. The hardware dummies are used for particular cases and injury criteria to be assessed, e.g. Q-Series Child Dummy [10]. Besides the physical dummies, there also exists a group of numerical (virtual) models of the dummies [17]. On the other hand, the virtual small specific anthropometric human model (children) lays back. Such contribution is not currently the main topic of research. One can find some children model for example in MADYMO model family [3, 6, 16].

Besides the dummies, human body models are starting to play an important role in safety system design [9]. Compared to physical dummies, the virtual human models can usually be scaled for representing a broader spectrum of human anthropometry. The scaling process usually

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defines a reference model that is validated for various impact loading [18]. However, the scaled models should be validated as well, since their anthropometry and dynamical properties may change considerably.

2. Methodology

The goal of the paper is to contribute to the development of biofidelic human models useful for a virtual approach in vehicle safety technology taking into account different age and body stature. The previously developed human body model VIRTHUMAN [18], including scaling algorithm [11], is used for developing a six years old human body model by automatic scaling.

The methodology concerns the validation of the frontal impact response of an automatically scaled small percentile six years old (hereafter referred to as 6YO) human body thorax. Previous papers show that scaling up to the higher mass and height do not cause serious problems, however down-scaling can influence the external shape and the dynamical properties in the negative way concerning both calculation stability and quality of results [4]. Hence the small percentile group $Q_{5/100}$ was developed for 6YO. Both geometrical and stiffness scaling were applied on the thorax [11].

The scaled model was prepared for a frontal impact test on the thorax defined by Kroell (see Fig. 1 on the left side). The standard frontal Kroell test concerns a rigid impactor of a mass equal to 23.4 kg and a diameter equal to 150 mm hitting the thorax frontally between the 4th rib and the 5th rib. Comparing the mass and the diameter to the body mass and size for particular specimens, both parameters were lowered for tests with small specimens at a lower age, giving a smaller impactor mass equal to 3.8 kg [14]. Three levels of impact speeds, 4.9 m/s, 6.7 m/s and 9.9 m/s, were used.

2.1. Reference model

The VIRTHUMAN model is based on the reference average (percentile $Q_{50/100}$) model and it has been validated within the wide spectrum of impact configurations [18].

The skeleton of the whole model is created as a multi-body system (MBS), which consists of rigid bodies. The skin surface is segmented into rigid surface parts (so-called superelements) interconnected by strips of elements without any mechanical response. The superelements are fixed to the basic MBS structure by springs and dampers (see Fig. 1 on the left right side).

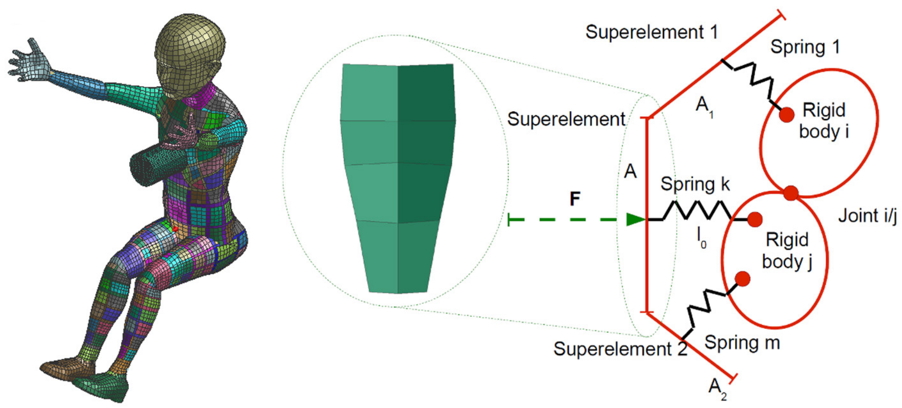

Fig. 1. VIRTHUMAN model in the Kroell test setup (left) with the deformable thorax (right)

2.2. *Anthropometric scaling*

The previously developed scaling algorithm able to automatically develop virtual human body models based on the reference model VIRTHUMAN [11] was exploited. The scaling is based on age and height percentiles. For the given age, the user inputs the desired percentiles and adapts total mass to specify the particular subject.

The whole scaling process draws data from a huge anthropometric database [2]. When the percentile is chosen, dimensions of particular segments of the model are adapted for height, width and depth [11]. The appropriate volume change updates also the mass of any articular segments, assuming uniform mass distribution and moments of inertia.

2.3. *Stiffness scaling*

The age influences the stiffness of the body. The overall body stiffness is based on the so-called flexindex [1]. The flexindex is a number representing the stiffness of 20 major human joints, giving the flexibility scale from 0 (rigid) to 4 (hypermobile) per joint. The flexindex is dependent on a subject and it is a unique number from 0 to 80. The scaling process takes the age-dependent flexindex [1] as a modifier for each joint's range of motion.

Scaling dimensions, mass, moments of inertia and joint stiffness complete the scaling process for any MBS model.

Preliminary tests proved correct response of the scaled models around the average percentile $Q_{50/100}$, and scaling up to higher age and percentiles, the response is still acceptable. As mentioned above, the problem appears with down-scaling, especially for models of 6YO human.

Hence, the authors adapted the scaling algorithm for thorax local stiffness scaling based on elasticity of the bone. Due to the available data [14], the local stiffness update was applied only for 6YO models. Considering the Hooke's law the force F acting on the superelement having area A is

$$F = \sigma A = E\epsilon A = \frac{EA}{l_0}\Delta l, \tag{1}$$

where σ is the strength of the spring element, E is the Young's mudulus, ϵ is the element deformation, l_0 is initial length of the spring and Δl is deflection (see Fig. 1 on the right side). Both geometrical parameter A and Δl are countable as the area of particular elements and spring length, so the appropriate scaling coefficients can be derived as ratios of the actual (6YO) and reference (adult) values as

$$\lambda_A = \frac{A_{6\mathrm{YO}}}{A_{\mathrm{adult}}}, \qquad \lambda_{l_0} = \frac{l_{0_{6\mathrm{YO}}}}{l_{0_{\mathrm{adult}}}}. \tag{2}$$

This geometrical scaling procedure was applied on all thorax superelements. Such scaling method might be adopted to scale the initial length of all springs l_0, which link particular superelements to the basic MBS structure. However, in this case, the change of the initial length of the spring does not significantly influence the results, so it is not considered here.

Irwin [12] considers the parietal bone stiffness as a basic parameter for the scaling of age-dependent bone stiffness. The age-dependent stiffness was measured by [14] (see Table 1).

Table 1. Age-dependent parietal bone stiffness [14]

Age [year]	E [GPa]
0	2.5
3	4.7
6	6.6
> 20	9.9

The scaling coefficient for elastic modulus is further calculated as a ratio of actual and reference Young's modulus as

$$\lambda_E = \frac{E_{\text{bone}_{6\text{YO}}}}{E_{\text{bone}_{\text{adult}}}} = \frac{6.6\ [\text{GPa}]}{9.9\ [\text{GPa}]} = 0.667. \tag{3}$$

Stiffness of particular springs linking superelements to the basic MBS structure is defined as partly linear force F dependent on deflection Δl. The scaling coefficients for force and deflection are derived from the limit values of maximum bending strength and maximum bending deflection of femur defined by [14] and stated in Table 2.

Table 2. Maximum bending strength and deflection [14]

Age [year]	$\sigma_{\max}$ [MPa]	$\epsilon_{\max}$ [–]
< 5	150–180	1.7–1.9
$[6, 40]$	180–210	1.1–1.3

Considering average values from Table 2, the scaling coefficients for stress and deflection are derived as

$$\lambda_\sigma = \frac{\sigma_{\max_{6\text{YO}}}}{\sigma_{\max_{\text{adult}}}} = \frac{165\ [\text{MPa}]}{195\ [\text{MPa}]} = 0.85, \qquad \lambda_\epsilon = \frac{\epsilon_{\max_{6\text{YO}}}}{\epsilon_{\max_{\text{adult}}}} = \frac{1.8\ [\text{mm}]}{1.2\ [\text{mm}]} = 1.5. \tag{4}$$

Since the spring stiffness in the VIRTHUMAN model [11] is defined within the curve (force vs. deflection), force F and deflection Δl are finally scaled separately based on coefficients given by (2), (3) and (4). Equation (1) is rewritten to

$$F_{6\text{YO}} = \epsilon_{6\text{YO}} E_{6\text{YO}} A_{6\text{YO}} = \underbrace{\epsilon_{\text{adult}} E_{\text{adult}} A_{\text{adult}}}_{F_{\text{adult}}} \underbrace{\lambda_\epsilon \lambda_E \lambda_A}_{\lambda_F}. \tag{5}$$

Analogously, deflection Δl is derived as

$$\Delta l_{6\text{YO}} = l_{0_{6\text{YO}}} \frac{\sigma_{6\text{YO}}}{E_{6\text{YO}}} = \underbrace{l_{0_{\text{adult}}} \frac{\sigma_{\text{adult}}}{E_{\text{adult}}}}_{\Delta l_{\text{adult}}} \underbrace{\lambda_{l_0} \frac{\lambda_\sigma}{\lambda_E}}_{\lambda_{\Delta l}}. \tag{6}$$

2.4. *Validation*

For frontal impact validation of the scaled model, the Kroell test [15] was used. The Kroell test concerns a blunt impact of a cylinder to the thorax. The thorax stiffness defined as the force dependent on deflection is assessed. The human sits on a rigid plate and the cylinder impactor (the mass equals 23.4 kg and the diameter equals 150 mm) impacts the thorax between the 4$^{\text{th}}$ rib and the 5$^{\text{th}}$ rib (see Fig. 1 on the left side). For children, an impactor of lower mass of 3.8 kg is used to be comparable to the thorax mass [14].

Impact velocities 4.9 m/s, 6.7 m/s and 9.9 m/s were chosen, which means that low, medium and high energy loading is evaluated. However, experimental data developed by Kroell [15] are normalized to the percentile $Q_{50/100}$. Hence, the Mertz method for corridor scaling was implemented [12]. The method exploits the corridors for percentile $Q_{50/100}$ and defines scaling coefficients based on geometrical and mass parameters of the particular model. The basic coefficients according to [12] for scaling the Kroell test are:

- Total body mass coefficient

$$\lambda_{mt} = \frac{m_{\mathrm{body}_{\mathrm{scal}}}}{m_{\mathrm{body}_{50\%}}}$$

- Impactor mass coefficient

$$\lambda_{\mathrm{imp}} = \frac{m_{\mathrm{imp}_{\mathrm{scal}}}}{m_{\mathrm{imp}_{50\%}}}$$

- Impactor initial velocity coefficient

$$\lambda_v = \frac{v_{\mathrm{scal}}}{v_{50\%}}$$

- Thorax and impactor mass coefficient

$$\lambda_{ms} = \frac{(m_{\mathrm{thor}} + m_{\mathrm{imp}})_{\mathrm{scal}}}{(m_{\mathrm{thor}} + m_{\mathrm{imp}})_{50\%}}$$

- Equivalent body mass coefficient

$$\lambda_{me} = \frac{\lambda_{mt}\lambda_{\mathrm{imp}}}{\lambda_{ms}}$$

- Sitting height coefficient

$$\lambda_z = \frac{h_{\mathrm{sitting}_{\mathrm{scal}}}}{h_{\mathrm{sitting}_{50\%}}}$$

- Bone elastic modulus coefficient

$$\lambda_E = \frac{E_{\mathrm{bone}_{\mathrm{scal}}}}{E_{\mathrm{bone}_{50\%}}}$$

- Thorax stiffness coefficient

$$\lambda_k = \lambda_E \lambda_z$$

The defined subscripts are named in order to keep the physical meaning of the quantities. Consequently, the *scal* subscript represents scaled unit, 50 % is a reference median value, *imp* is impactor, *thor* is thorax. Meaning of the other subscripts is evident from their name or from the name of the scaling coefficients. Those coefficients define scaling factors for experimental force, deflection and time to be multiplied:

- Force coefficient

$$R_F = \lambda_v \sqrt{\lambda_{me}\lambda_k}$$

- Deflection coefficient

$$R_D = \lambda_v \sqrt{\frac{\lambda_{me}}{\lambda_k}}$$

- Time coefficient

$$R_T = \sqrt{\frac{\lambda_{me}}{\lambda_k}}$$

3. Results

The values of the scaling coefficients for the 6YO model are stated in Table 3.

Table 3. 6YO scaling coefficients

Coefficient	λ_ϵ	λ_E	λ_A	λ_σ	λ_{l_0}	$\boldsymbol{\lambda_F}$	$\boldsymbol{\lambda_{\Delta l}}$
6YO	1.5	0.67	0.44	0.85	0.52	**0.44**	**0.66**

Application of the Mertz method [12] to create the scaled experimental corridors develops the scaling coefficients shown in Table 4. The calculated coefficient was applied on scaling both the experimental corridors and the mechanical properties (deformation curves of the spring) of the virtual human body model.

Table 4. Experimental scaling coefficients

Coefficient	λ_{mt}	λ_{imp}	λ_v	λ_{ms}	λ_{me}	λ_z	λ_E	λ_k	$\boldsymbol{R_F}$	$\boldsymbol{R_D}$	$\boldsymbol{R_T}$
6YO	0.23	0.16	1	0.22	0.18	0.71	0.67	0.47	**0.28**	**0.6**	**0.6**

Scaled force and deflection corridors for 6YO of percentile $Q_{5/100}$ for three impact velocities 4.9 m/s, 6.7 m/s and 9.9 m/s are shown in Figs. 2–4. Fig. 2 shows the time dependent 6YO thorax deflection (left) and impact force dependent on deflection for percentiles $Q_{5/100}$ at 4.9 m/s impact velocity. Both signals fit in the improved experimental corridors well.

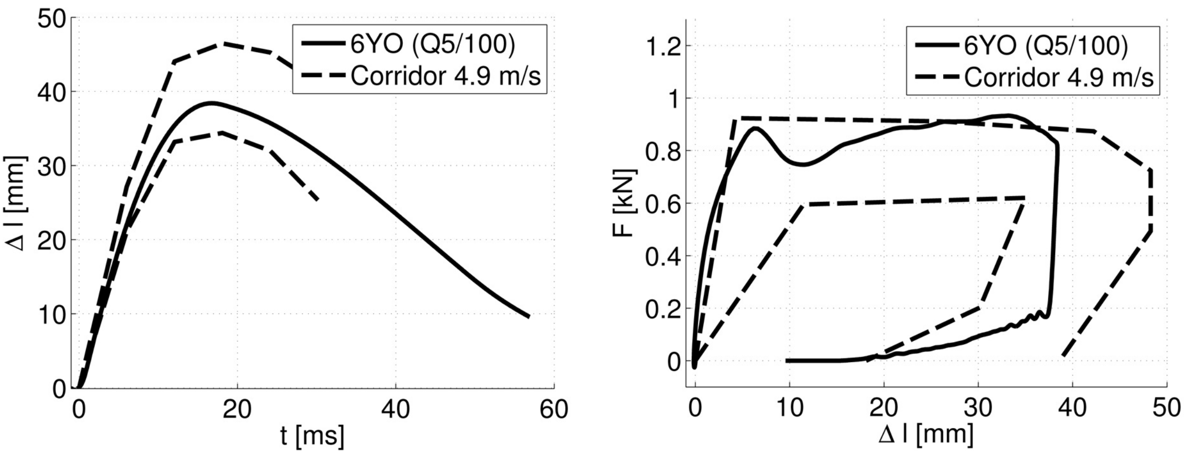

Fig. 2. Improved response for $Q_{5/100}$ at 4.9 m/s

Fig. 3 shows the time dependent 6YO thorax deflection (left) and impact force dependent on deflection for percentiles $Q_{5/100}$ at 6.7 m/s impact velocity. Both signals fit in the experimental corridors well. However, the force response exceeded the corridor at the region of the maximum values.

Fig. 4 shows the time dependent 6YO thorax deflection (left) and impact force dependent on deflection for percentiles $Q_{5/100}$ at 9.9 m/s impact velocity. Both signals fit in the experimental corridors well.

The model behaves well and the response is not perfectly within the corridors and there is some additional effort to be done. The paper proves that the model performance is suitable taking into account fully automatic development scaling process. The obtained scaling method shows

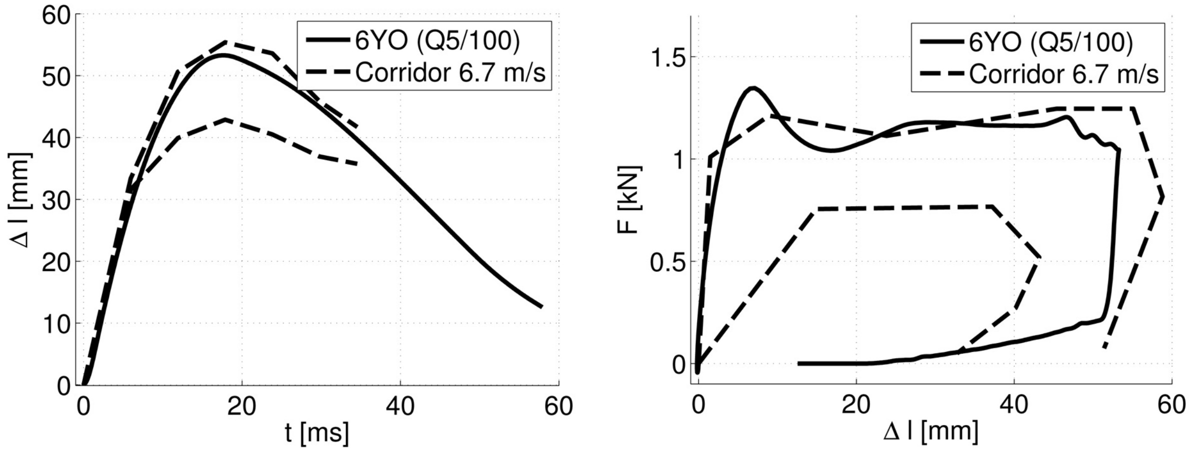

Fig. 3. Improved response for $Q_{5/100}$ at 6.7 m/s

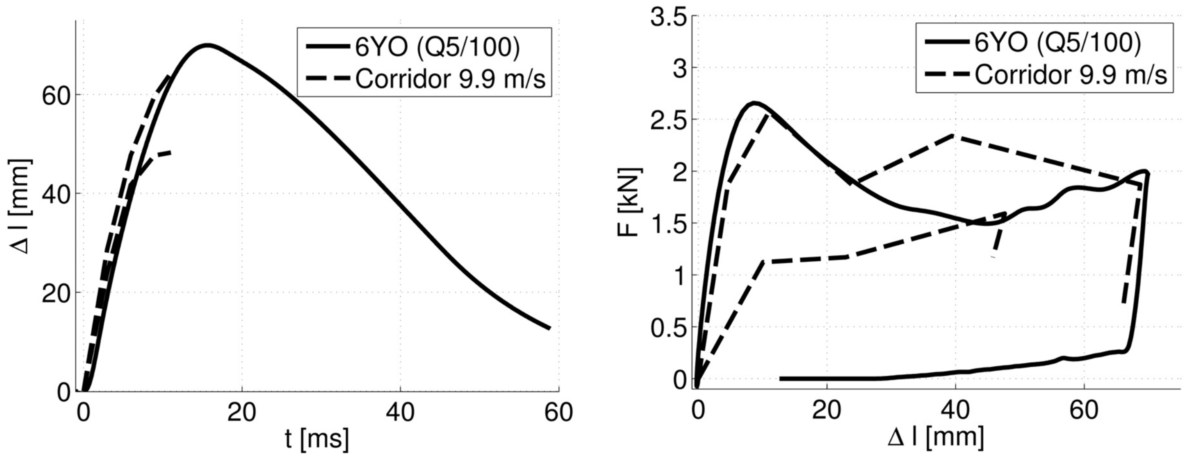

Fig. 4. Improved response for $Q_{5/100}$ at 9.9 m/s

the good capability for a down-scaling process. Since the up-scaling does not usually bring any problem, the downward process does [4]. The presented paper introduces the approach of scaling not only the mechanical properties of the model but also of the loading items (impactor mass and dimension) and even more, the mechanical response corridors improvement. Within this method, the simulations performed on small children models are comparable to the adult models.

4. Conclusion

The scaled models are useful for virtual assessment of human body behaviour under different loading types related to a wide spectrum of the population.

The presented work exploited the existing virtual human body model VIRTHUMAN and applied a previously developed scaling algorithm to develop age and percentile group dependent dataset to be evaluated in the Kroell impact test. The numerical analyses show that for a local impact test it is also necessary to scale the local stiffness of the area.

Based on the stress/strain theory, the local stiffness was updated with the stiffness scaling using the reference bone stiffness data in order to fit the response into the scaled experimental corridors.

The model behaves well for small percentile $Q_{5/100}$ but the response is not perfectly within the corridors and there is some additional research to be done. For the future improvements, the application of thorax bone stiffness can be considered as a reference one, instead of parietal bone. However it is not easy to find or estimate such values. Maximum bending strength and deflection values might be improved especially for the case of 6 years old human.

Acknowledgements

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