Extended Study of Human Trunk Model Including Inter-Individual Variability

Ferdaous Moalla[1*], *Fehmi* Najar[2,1], and *Sadok* Mehrez[3,1]

[1]Applied Mechanics and Systems Research Laboratory, Tunisia Polytechnic School, University of Carthage, 2078 Tunis, Tunisia
[2]Department of Mechanical Engineering, College of Engineering at Al Kharj, Prince Sattam bin Abdulaziz University, Al-Kharj, 11942, Saudi Arabia
[3]Department of Mechanical Engineering, National Engineering School of Tunis, University of Tunis El Manar,1002 Tunis, Tunisia

Abstract. Personalized musculoskeletal models of the human trunk are essential for advancements in diagnosis, medical devices development, and ergonomics. In a previous study, a finite element model (FEM) of the human trunk was developed, and its effective mechanical properties such as stiffness and damping were calibrated using experimental data from a single participant when subjected to time-dependent excitations at the thoracic level. The objective of the present work is to extend this modeling approach by evaluating its ability to incorporate inter-individual variability. To this end, an identical experimental campaign was conducted on a second participant. The model parameters were then identified for this new subject using the same calibration methodology. A strong correlation was obtained between the simulated results and the experimental data from the second participant. This successful outcome confirms the model adaptability to other individual variation. The validated approach constitutes a reliable tool for generating realistic and subject-specific databases intended for advanced applications in machine learning and biomechanical simulation.

1 Introduction

Human trunk behavior has been the subject of numerous studies aimed to characterize its mechanical properties. These parameters play a crucial role in lumbar stability, injury prevention, and understanding neuromuscular mechanisms. Modeling approaches in literature extend from simple lumped parameter systems [1] to complex multi-segment musculoskeletal models [2,3]. Lumped models are more used for system identification due to their mathematical simplicity and robustness, while advanced finite element and musculoskeletal models offer deep understanding into spinal load sharing and muscle coordination. Several studies have shown that trunk mechanical properties vary significantly depending on muscle activation, posture, and individual conditions [4]. Recent works highlight the need for subject-specific modelling [5] presenting major issue in trunk biomechanics, with the increasing evidence of inter-individual variability in trunk stiffness,

---

* Corresponding author: moallaferdaous@hotmail.com

mass distribution and neuromuscular strategies [6]. Other studies focus on the sensitivity analysis of material properties variability in models of the human spine [7]. These observations illustrate the importance of adaptable models for describing diverse behaviors for both healthy and pathological subjects [8]. In this context, a multi body musculoskeletal model based on finite element technique was developed in a previous work [9]. This model includes the spine, the rib cage and 174 trunk muscles. The effective mechanical properties of the model such as stiffness and damping coefficients were successfully calibrated using experimental data collected from one participant subjected to time-varying mechanical excitations. This pilot study confirmed the modelling approach. The main objective of this study is to test the feasibility of the developed modelling approach on a second participant having differences from the first one. This extended study aims to develop generate subject-specific models to be used for advanced applications such as machine learning model training and complex biomechanical simulation.

2 Methodology

2.1 Experimental protocol

Experimental tests were performed using a set-up developed to study the dynamic behavior of the human trunk under flexion-extension movements. It includes an adjustable metallic support, an excitation system generating an external force perturbation applied to the T8 thoracic level, a load cell sensor to measure the applied forces and a laser sensor recording trunk displacements.

Signals are collected and recorded through a data acquisition system with a sampling frequency of 2kHz and a fourth-order Butterworth low-pass filter was implemented.

A pilot study was developed in the previous work [9], and a series of tests was conducted on one healthy participant to calibrate a 3D finite element model of the human trunk. The extracted mechanical parameters from experimental displacement curves are compared to those found numerically. To test the robustness of the numerical model, series of tests was conducted on a second participant. The experimental protocol was reproduced as depicted in Figure 1 taking into account participant size adjustment in the set up.

Fig. 1. Experimental set up

The selected participant is in good health and has no history of spinal pathology or low back pain, his physical appearance is chosen to be different from the first one. Anthropometric characteristics of the two participants are detailed in table 1.

Table 1. Anthropometric characteristics.

characteristics	Participants	
	Participant 1	Participant 2
Age (year)	28	38
Weight Kg)	68	75
Size (cm)	173	178
BMI (Kg/m2)	22,6	23,7

2.2 Numerical model

A 3D biomechanical model of the human trunk was developed to reproduce the global dynamic response observed experimentally. The trunk model is subdivided into 335 beam finite elements, including the spine, the rib cage, discs and 174 trunk muscles. It takes into account linear stiffness, inertial characteristics, and linear viscous damping. Equation formulas and numerical resolution approach are detailed in the previous work [9] considering the system is linear for small displacements.
Newmark method is used as a time integration method to solve the derived equations in time after applying a modal projection to extract the dynamic properties of the trunk.

3 Results and discussions

3.1 *Experimental results*

Table 2 summarizes the estimated dynamic parameters for the two participants under different applied forces. The applied force and trunk response are observed at the T8 level of the thoracic vertebra, directed towards the sternum. Effective trunk parameters are calculated from the maximum measured displacement d0 and the associated maximum force Fmax. An increase in force level leads to an increase in maximum displacement Dmax, demonstrating that the trunk response is amplified under the applied excitation.

Table 2. Dynamic parameters.

Participants	Parameters						
	Force max (N)	Overshoot d0 (mm)	K (N/m)	M (Kg)	wn (Hz)	wd (Hz)	ε
P1	90	20.5	4604	15.5	2.73	2.02	0.16

	100	**24.1**	**4270**	**14.8**	**2.7**	**2.58**	**0.21**
	120	**29.7**	**4140**	**12.2**	**2.93**	**2.76**	**0.31**
P2	**90**	**18**	**4918**	**10.17**	**3.64**	**3.5**	**0.28**
	100	**20.62**	**4849**	**9.5**	**3.77**	**3.6**	**0.3**
	120	**26.5**	**4528**	**8.6**	**3.8**	**3.65**	**0.34**

A comparison between the two participants, subjected to the same force levels shows significant differences. The first participant P1 presents greater peak displacements values D_{max} and lower effective stiffness values K then P2, the effective mass M is significantly lower for P1 due to morphological differences. Natural frequencies w_n and damping frequencies w_d are higher for P2, reflecting a faster dynamic response. Damping values ε are higher in P2, indicating greater energy dissipation for P2.

These results indicate that the morphology strongly influences mechanical parameters of the trunk and confirm the internal variability of the model. Low stiffness of P1 leads to a decrease in natural frequency and an increase in displacement amplitudes, which can be explained by increased postural rigidity for the second participant.

3.2 Numerical results

Numerical results were performed using MATLAB software. Dynamic response of the system was simulated by solving the differential equation of the mass-spring-damper model considering the experimental damping ratio from the temporal response. Minor adjustment of global mass and stiffness parameters was performed in order to have the natural frequency ωd close to the one obtained experimentally. No structural modification of the model was introduced.

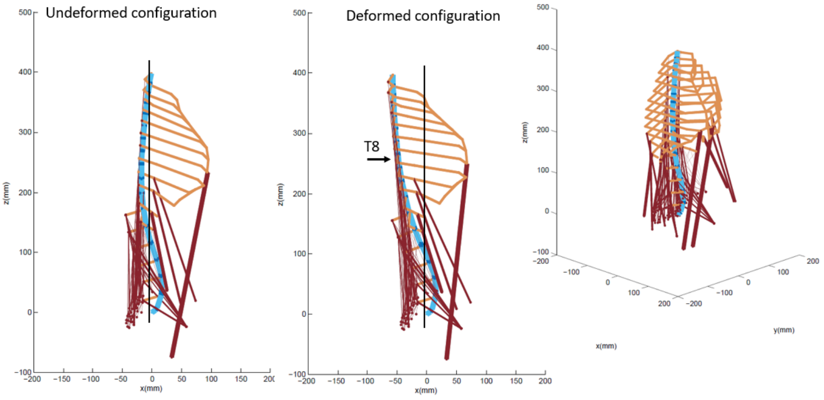

Fig. 2. Trunk musculoskeletal model

Fig.2. presents the initial trunk configuration and the deformed one at a time step close to the overshoot obtained from the multibody musculoskeletal model. The deformed configuration plotted under a load force vector presenting a maximum of 120 N, flexural deformation is observed in refer to the vertical black line in the figure. Dynamic displacements are obtained

when the force is applied at the T8 thoracic level and could be obtained for all model nodes.

Relative errors remained below 6% for peak displacement and below 3% for natural frequency, confirming good agreement between experimental and numerical responses.

Results confirm that the proposed modelling approach can successfully reproduce subject-specific mechanical characteristics, it has also the possibility to predict responses at any tissue-level, disc-level and for entire spine elements.

3.3 Discussion

The proposed study shows that the variability of parameters at the individual level has a significant impact on simulation results. Calibrating the numerical model using experimental data provides an accurate estimation of trunk stiffness and damping, allowing for reliable prediction of its dynamic behavior under various loading conditions.

A trunk with good rigidity and effective damping offers better stability under dynamic perturbations. Low rigidity can lead to instability under load, while excessive rigidity can cause harmful muscle overuse. Damping reflects the neuromuscular system ability to absorb shocks, underestimating damping can lead to overestimating peak stresses during oscillations while insufficient damping increases the risk of imbalance or back pain.

This study represents a robustness assessment of the linear FE trunk model on a second participant. However, only two healthy volunteers on the experimental data present a limitation to capture all variabilities. Access to open datasets [10,11] for normal and deformed thoracolumbar spine models now allows for a more systematic exploration of the effects of geometry and mechanical properties on simulated spinal behavior, providing a solid basis for quantifying variability in future studies.

4 Conclusions

By applying an identical calibration protocol to a second participant, we demonstrated that the model able to adapt inter-individual variability in mechanical properties.

Expanding this study for different participants would enhance the generalizability of the results to further developments across multiple domains.

References

1. J. Cholewicki, A.P. Simons, A. Radebold, Effects of external trunk loads on lumbar spine stability. *J. Biomech.* **33**, 1377–1385 (2000). https://doi.org/10.1016/S0021-9290(00)00118-4
2. M. Christophy, N.A.F. Senan, J.C. Lotz, O.M. O'Reilly, A musculoskeletal model for the lumbar spine. *Biomech. Model. Mechanobiol.* **11**, 19–34 (2012). https://doi.org/10.1007/s10237-011-0290-6
3. M. de Zee, L. Hansen, C. Wong, J. Rasmussen, E.B. Simonsen, A detailed lumbosacral spine model with muscle forces. *J. Biomech.* **40**, 1219–1227 (2007). https://doi.org/10.1016/j.jbiomech.2006.05.030
4. S.H. Brown, J.R. Potvin, Constraining spine stability calculations using in vivo data: examination of trunk stiffness and damping. *J. Biomech.* **38**, 880–890 (2005). https://doi.org/10.1016/j.jbiomech.2004.05.011

5. N. Arjmand, A. Shirazi-Adl, Sensitivity of spinal loads and muscle forces to optimization criteria in static lifting tasks. *Med. Eng. Phys.* **27**, 571–579 (2005).
6. Larivière, H. Mecheri, D. Gagnon, A.B. Arsenault, Effect of load and trunk orientation on trunk stiffness and damping. *Clin. Biomech.* **25**, 1031–1036 (2010).
7. J. Noailly, J. Wilke, H. Planell, Sensitivity analysis of a lumbar spine finite element model to material properties. Med. Biol. Eng. Comput. 45, 527–538 (2007). https://doi.org/10.1007/s11517-007-0197-5
8. A. Ahmadi, M. Mohseni, N. Arjmand, AI-based human whole-body posture-prediction in continuous load reaching/leaving activities. J. Biomech. 185, 112681 (2025). https://doi.org/10.1016/j.jbiomech.2025.112681
9. F. Moalla, S. Mehrez, F. Najar, Multi-body musculoskeletal dynamic model of the human trunk based on an experimental approach. *Arch. Appl. Mech.* **93**, 1201–1215 (2023).

   https://doi.org/10.1007/s00419-022-02323-x
10. M. Rasouligandomani, A. del Arco, F.K. Chemorion, et al., Dataset of finite element models of normal and deformed thoracolumbar spine. Sci Data 11, 549 (2024). https://doi.org/10.1038/s41597-024-03351-8
11. A. Jamshidian, A. Rouyin, H. Nazemi, et al., The effect of walking speed on spinal loads and trunk muscle forces using subject-specific musculoskeletal modeling: a database for clinical and modeling applications. J. Orthop. Surg. Res. 20, 951 (2025). https://doi.org/10.1186/s13018-025-06408-5