International Journal of Occupational Safety and Ergonomics (*JOSE*) 2012, Vol. 18, No. 4, 509–519

A Whole Body Postural Loading Simulation and Assessment Model for Workplace Analysis and Design

Francisco Rebelo
K. Correia da Silva

Ergonomics Laboratory, CIPER, Technical University of Lisbon, Cruz-Quebrada, Portugal

Waldemar Karwowski

Department of Industrial Engineering and Management System, University of Central Florida, Orlando, USA

This study reports on the development and validation of a new computer model for simulating human postures at work, and assessing the reaction forces and bending moments in 43 human articulation joints. The proposed model estimates the intradiscal pressure in the vertebral column in response to external loading forces encountered during human interactions with work objects or processes. The model was implemented in a self-contained interactive software package. The simulation results compare favorably with the reported experimental data, and provide reasonable confidence in the quality of the model. Its characteristics and its applications in evaluating physical task performance are also discussed.

simulation models   spinal loading   ergonomics   manual materials handling
work design   safety

1. INTRODUCTION

A review of the literature reveals a wealth of information on human capabilities and limitations, both in physical aspects (e.g., body size, tolerances, strength [1]) and mental aspects (e.g., perception and information processing [2]). However, human factors professionals often find it difficult to meaningfully incorporate this knowledge into their projects. A possible reason is that ergonomics literature currently contains studies on people's capabilities regarding single design variables such as biomechanical predictions; however, this basic research does not lead to applied design solutions [3, 4]. Another reason, as pointed out by Butters and Dixon [5], is that the available data and recommendations in ergonomic guidelines and heuristics are often incomplete and out-of-date; they rarely incorporate the developments of contemporary findings.

A solution to this problem has been to use tools that incorporate ergonomics knowledge that aids the evaluation of human interaction during the earliest stages of product, service or task design. Normally, these tools focus on single aspects of the performance of a human operator and require the designer to conduct separate analyses with several different tools [6, 7]. This difficulty stimulated the development of integrated tools that allow designers to incorporate features, at the earliest stages of design, which not only optimize the users' safety and well-being, but also improve the efficiency of the system.

Correspondence and requests for offprints should be sent to Francisco Rebelo, Ergonomics Laboratory, Technical University of Lisbon, Estrada da Costa, 1495-688 Cruz-Quebrada, Portugal. E-mail: fsrebelo@gmail.com.

The most successful approaches use commercially available computer-aided design (CAD) systems with relevant analysis software. These software packages use parametric digital mannequins and ergonomics knowledge to optimize variables like reach, clearance and posture. Examples of these include 3DSSPP/AutoCAD [8], Safework [9], Jack [10] and Ramsis [11]. To use these tools effectively, the user must be an expert in CAD systems and modelling techniques. To use the mannequins effectively and optimize a new design, the user must have relevant ergonomics knowledge of the scenario. Typically, human factors professionals do not have expertise in CAD, while design engineers lack ergonomics and human factors knowledge.

Many of these software packages have serious shortcomings and limitations. For example, an investigation of the most popular program for calculating low-back compression force predictive capabilities, 3DSSPP [8], has the following limitations:

- the calculation of the lumbar disc compression force is only at the L5/S1 level;
- the spine is represented by a beam and does not consider the normal configurations during work activity;
- it is only possible to put loads in the hands;
- it is impossible to simulate support of the upper limbs or other parts of the body in the physical environment.

These constraints can limit the human factors professional in optimizing designs, due to the inability to simulate or obtain information about the users' possible strategies for interacting with a new product or work system. Furthermore, previous integrated solutions are part of software packages deemed too expensive for most human factors consultants.

Chaffin described a need to improve existing digital human models so they are better able to serve as tools for effective ergonomics analysis and design [12]. Particularly, the need to include valid posture and motion prediction models for various populations was listed as a requirement.

These considerations led to the development of a three-dimensional computer model named Humanoid Articulation Reaction Simulation (HARSIM) for optimizing product, workspace and task procedures, with the end goal being minimizing musculoskeletal stress and strain. This paper discusses the development and validation of this model,

2. MODEL CHARACTERISTICS

The HARSIM model consists of a humanoid computer representation with 38 segments, a full spine with 24 vertebrae, and upper and lower limbs with 8 and 6 segments, respectively. The developed model has 100 degrees of freedom, 72 for the spine, 12 for the lower limbs and 16 for the upper limbs. For each joint articulation, the model calculates three reaction forces (one axial and two shear) and three bending moments around each axis of the orthonormal reference frame and the maximal compression force in the intervertebral space. To simplify interaction with the model, only data related to the vertebral column were used.

The HARSIM model has four operational features.

- Human model generation: this feature enables the user to generate any anthropometrical profile based on population percentiles, for different age groups and genders. With this feature, the user can select an anthropometrical profile from the database or tailor-build a human model by specifying the exact human dimensions.
- Simulation of postures and movements: the model allows the simulation of the main postures in an interaction with a product or workplace situation. Particularly, the model uses a fully articulated spine, capable of simulating every posture or movement. The upper and lower limbs can simulate the main postures or movements, including the scapulahumeral rhythm. To prevent the simulation of impossible movements, the user is warned when the articulation limits are reached.
- Creation of geometrical objects: to optimize a product or workplace situation, the user can

simulate the physical environment, employing simple three-dimensional graphics such as parallelograms, pipes, spheres, boxes and simple objects.

- Calculation of forces, strain and stress in each articulation joint: to calculate forces, stress and strain, the HARSIM model requires a full geometrical and mechanical characteristics for each segment. It is possible to introduce loads that can be internal, due to the weight of the body segments, and external, due to loads applied in any of the 38 segments of the model. Depending on the situation, different kinds of support can be introduced in all articular joints. To introduce data into the program, the user can select available data from the subject literature or specify the data to be collected.

3. SIMULATING POSTURE IN HARSIM

To simulate a posture in HARSIM, we employed the Denavit-Hartenberg formulation [13] to implement a forward kinematics algorithm for all bodies of the model. It is easy to create a posture for the upper and lower limbs, but for the vertebral column with 24 vertebrae corresponding to 72 rotation angles and values, it could be impractical to generate a vertebral column posture given the great number of variables to consider. To overcome these difficulties, we developed a model that allowed the capability to generate the spinal angular values from sagittal, frontal and axial torsion.

A configuration of the vertebral column is controlled independently for each vertebral column zone (lumbar, dorsal and cervical), and the same value of the intervertebral angle is applied to all vertebrae of that zone. For example, to simulate trunk bending with axial rotation, an intercalated angular value for saggital and axial rotation is introduced to all lumbar vertebrae until the posture is reached. The model also incorporates the physiological limits of the vertebral column according to Kapandji [14] for the maximum values of the angles of flexion–extension, lateral bending and axial rotation.

Another aspect of creating a posture is the possibility to fit the HARSIM model over an image of a human interacting with a product or a load. In this case, the model is scaled according to the subject's anthropometry. The user interactively fits the HARSIM model over the image, until posture is reproduced.

4. CALCULATING REACTION FORCES IN EACH INTERVERTEBRAL JOINT

The mechanical response of the HARSIM model to loading was approximated with a finite-element direct-stiffness method of structural analysis. The vertebral column is represented as a continuous beam-like structure with parametric geometry and elastic properties divided into 24 parametric three-dimensional space frame elements, each representing a vertebra. The HARSIM upper and lower limbs are modelled with the same methodology.

The vertebral column is simulated with a three-dimensional finite element model (FEM) beam connected between two of its joints and a system of actions, i.e., forces and torques that are applied at each joint. This FEM quantifies the static equilibrium condition reached by the structure under the effects of the actions applied to its joints; static equilibrium is expressed with a relationship between the actions and the linear and angular displacements [15].

Table 1 displays the default geometrical parameters that characterize the vertebral column. The vertebrae are represented as elliptical cylinders whose cross-sectional area corresponds to that of the vertebral body plus the projection, on the plane of the body, of the articulating facets. This data were obtained by measuring the geometry of a model skeleton.

In the HARSIM model, the vertebral column is considered as a homogeneous linear elastic material whose mechanical characteristics, Young's modulus (E) and Poisson's ratio (ν), have to be specified. The data available in the literature [16, 17, 18] specify for the nucleus and the annulus matrix a value of E of the order of 4 and 2 MPa, respectively, and of ~0.46 for ν. This value of E is

TABLE 1. Geometrical Parameters of the Vertebral Column

Vertebra	Cross-Section Area (m^2)	Cross-Section Major Axis r_y (m)	Cross-Section Minor Axis r_z (m)	Inertial Moment I_y	Inertial Moment I_z
L5	0.0015634	0.05584	0.03565	4.88E-06	1.99E-06
L4	0.0014833	0.05454	0.03463	4.41E-06	1.78E-06
L3	0.0013978	0.05313	0.03350	3.95E-06	1.57E-06
L2	0.0013136	0.05159	0.03242	3.5E-06	1.38E-06
L1	0.0012325	0.04974	0.03155	3.05E-06	1.23E-06
T12	0.0011525	0.04737	0.03098	2.59E-06	1.11E-06
T11	0.0010699	0.04439	0.03069	2.11E-06	1.01E-06
T10	0.0009851	0.04099	0.03060	1.66E-06	9.22E-07
T9	0.0009016	0.03758	0.03055	1.27E-06	8.42E-07
T8	0.0008259	0.03465	0.03035	9.92E-07	7.61E-07
T7	0.0007635	0.03258	0.02984	8.1E-07	6.8E-07
T6	0.0007185	0.03158	0.02897	7.17E-07	6.03E-07
T5	0.0006885	0.03158	0.02776	6.87E-07	5.31E-07
T4	0.0006652	0.03222	0.02629	6.91E-07	4.6E-07
T3	0.0006373	0.03288	0.02468	6.89E-07	3.88E-07
T2	0.0005937	0.03284	0.02302	6.4E-07	3.15E-07
T1	0.0005317	0.03162	0.02141	5.32E-07	2.44E-07
C7	0.0004571	0.02922	0.01992	3.9E-07	1.81E-07
C6	0.0003815	0.02616	0.01857	2.61E-07	1.32E-07
C5	0.0003160	0.02321	0.01734	1.7E-07	9.5E-08
C4	0.0002671	0.02101	0.01619	1.18E-07	7E-08
C3	0.0002350	0.01983	0.01509	9.24E-08	5.35E-08
C2	0.0002158	0.01948	0.01411	8.19E-08	4.3E-08

Notes. r_y = major axis of the vertebrae cross section in *y* axis, r_z = minor axis of the vertebrae cross section in *z* axis, I_y = moment of inertia in *y* axis, I_z = moment of inertia in *z* axis.

quite insufficient to prevent the bending of the vertebral column even when loaded with body weight. Thus, the equilibrium of the vertebral column achieved through the action of the associated musculature can be interpreted as a fictitious increase in the value of E attributed to the vertebral column. This conformity was used for each configuration, where a given vertebral column movement was decomposed and an effective value of E which was obtained. This was done through a relaxation procedure, as illustrated in Figure 1. This procedure gradually reduced the magnitude of E from an initial value of 1 GPa to that for which a maximum linear deformation of 0.0005 m or a maximum angular deformation of the 0.0085 rad is reached at any of the column nodes.

5. LOADING HARSIM

The vertebral column is shown to be the resistant shaft of the trunk; its performance under load is obtained by transferring, to the corresponding vertebrae, the loads applied to the different regions of the body: the thorax, the abdomen, the head and the upper limbs. The weight of the head is directly applied to C1. The external load applied to the upper limbs is transferred to the first three thoracic vertebrae [19], or to the last five cervical vertebrae, according to whether the direction of the applied forces is toward the head or away from it. This load is divided into equal fractions by the vertebrae of the selected region; each fractional load is transferred from the limb to the corresponding vertebra.

The load corresponding to the weight of the internal organs of the trunk is taken into account

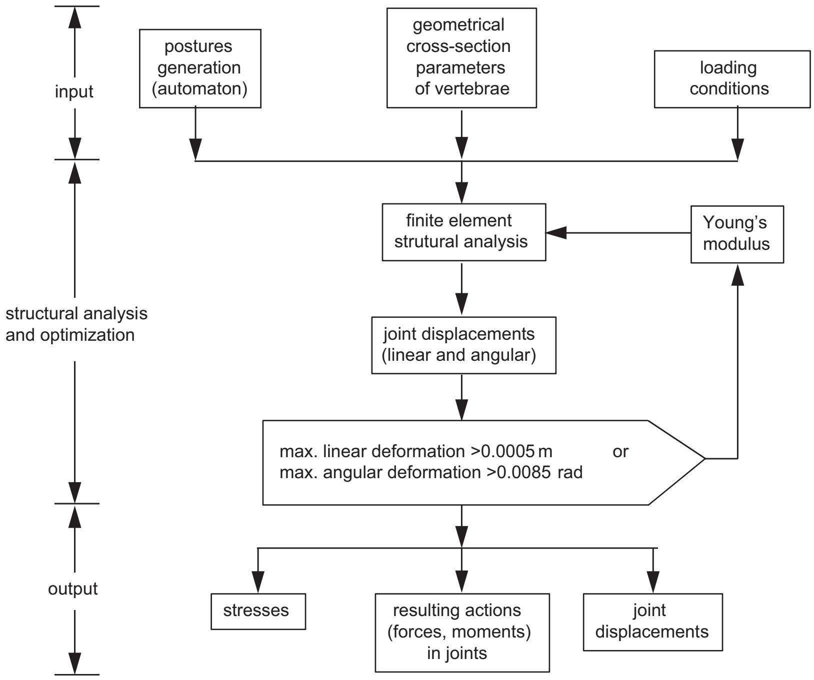

Figure 1. Flowchart for a computer program for the structural analysis of the response of the vertebral column to loading, including the relaxation procedure used for adjusting the value of Young's modulus.

by considering this body compartment divided into parallel slices, one for each vertebra [19, 20]. The corresponding fractional weight, supposedly acting vertically on the respective center of mass, is then transferred to the associated vertebra by the same expedient of adding a torque. This replicates the action of that load in its original point of application. The position of the center of mass of each body slice and weight is evaluated along the lines described by Clauser, McConville and Young [21]. Each body slice is treated as homogeneous; therefore, its center of mass is located at the corresponding centroid. The center of mass (and its associated vertebra slice) moves so that it alters the lever arm to be used for calculating the torque. This replicates the eccentric location of the slice weight.

The rib cage is an integrated part of the upper thoracic region of the vertebral column and performs mechanically as an articulated, thick-walled shell structure whose structural stability is the combined result of the activity of the thoracic muscles inserted in it [22]. The rib cage transfers loads resulting from the weight of the thorax, upper abdominal viscera and upper limbs to the thoracic region of the vertebral column.

6. VALIDATING HARSIM AT L4/L5

HARSIM is validated with in-vivo intradiscal pressure measurements as reported by Wilke, Neef, Hinz, et al. [23]. We used Wilke et al.'s anthropometrical data along with the following 10 activities (cases) formatted to the HARSIM model and calculated the intradiscal pressure in the vertebral column:

1. relaxed standing (Figure 2a);
2. holding a full crate of beer 60 cm away from the chest (Figure 2b);
3. holding a full crate of beer close to the body (Figure 2c);
4. trying to place a fingertip on the floor (Figure 2d);
5. lifting a full crate of beer (Figure 2e);
6. walking with two crates of beer (Figure 2f);
7. unsupported, relaxed erect sitting (Figure 2g);
8. erect sitting bent forward (Figure 2h);
9. relaxed erect sitting with backrest (Figure 2i);
10. body lifting by arm support (Figure 2j).

Table 2 shows the intradiscal pressure results on L4/L5, estimated by the HARSIM model, and the values for 10 daily activities [23]. The esti-

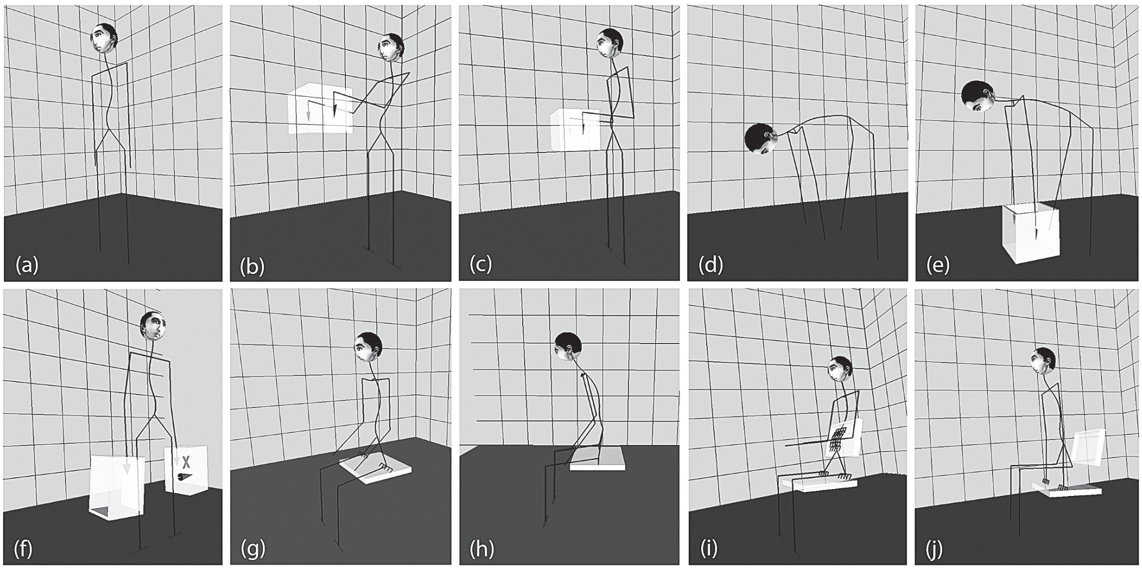

Figure 2. Simulated postures in the HARSIM model: (a) relaxed standing; (b) holding a full crate of beer 60 cm away from the chest; (c) holding a full crate of beer close to the body; (d) trying to place a fingertip on the floor; (e) lifting a full crate of beer bent over with a round back; (f) walking with two crates of beer; (g) unsupported, relaxed erect sitting; (h) erect sitting bent forward; (i) relaxed erect sitting with backrest; (j) body lifting with arm support. *Notes.* HARSIM = Humanoid Articulation Reaction Simulation.

mated intradiscal pressures reveal a good match. It is important to point out that Wilke et al.'s results were average values [23], and relevant information about data dispersion is unavailable.

TABLE 2. Calculated Values at L4/L5 in the HARSIM Model and Measured Values for 10 Daily Activities [24]

Case	HARSIM Calculated (MPa)	Measured (MPa) [24]
1	0.46	0.48
2	1.88	1.80
3	1.09	1.00
4	1.66	1.60
5	2.84	2.30
6	0.86	0.90
7	0.44	0.45
8	0.77	0.63
9	0.38	0.45
10	0.12	0.10

Notes. HARSIM = Humanoid Articulation Reaction Simulation. For a list of the activities, see section 6 and Figure 2.

The detailed results for case 2, holding a full crate of beer 60 cm away from the chest, demonstrate the benefits of the HARSIM model. Table 3 shows the estimated axial and shear forces, bending moments and intradiscal pressures at all levels of the spine. Experimental data in the literature that could be used to validate these results are unavailable; therefore, these results are explained according to the mechanical functioning of the spine. The axial forces decrease from the lumbar to thoracic zones, which is normal behavior for this structure. Relative to the bending moments, they increase in the thoracic level. This can be explained with the curvature of the spine.

Figure 3 shows a screen display of the HARSIM software; the figure illustrates the model's structure interacting with a virtual box (representing the crate of beer) and the graphs of the intradiscal pressure, bending moments, and axial and shear reaction forces, developed at each intradiscal disk in response to the loading system.

In the case of the intradiscal pressures, an increase in the values in the thoracic intradiscal spaces compared to the values at the lumbar zone is verified. These findings are in accordance with those of Cramer, King Liu and von Rosenberg [20], who demonstrated a relative increase in the stresses and injuries at the thoracic level, associated with accelerations in aircraft ejections. Serpil

TABLE 3. Calculated Axial and Shear Forces, Bending Moments and Intradiscal Pressure in All Levels of the Spine, Holding a Full Crate of Beer 60 cm Away From the Chest (Figure 2b)

Intervertebral Space	Axial Force (N)	Shear Force (N)	Bending Moment (N/m)	Intradiscal Pressure (MPa)
L5/S1	646.51	235.31	114.58	1.385
L5/L4	675.49	26.345	107.40	1.882
L4/L3	653.70	116.49	–106.6	1.859
L3/L2	591.53	274.22	–110.15	1.874
L2/L1	508.16	389.07	–118.52	1.945
L1/T12	478.49	406.74	–130.38	2.095
T12/T11	497.51	363.24	–142.42	2.275
T11/T10	508.56	325.86	–153.5	2.436
T10/T9	517.79	286.99	–163.41	2.580
T9/T8	537.26	218.51	–170.73	2.694
T7	542.43	168.52	–176.3	2.775
T6	545.11	109.52	–180.31	2.833
T5	539.84	67.12	–182.85	2.865
T4	531.77	15.56	–184.47	2.884
T3	519.93	8.46	184.85	2.882
T2	506.76	35.45	184.66	2.872
T1	492.69	57.18	183.89	2.854
C7	122.65	18.26	1.65	0.111
C6	110.47	18.44	1.29	0.097
C5	98.27	18.48	0.98	0.084
C4	86.36	16.90	0.64	0.071
C3	75.12	11.50	0.34	0.058
C2	63.81	4.82	0.15	0.048
C1	51.85	3.92	0.07	0.038

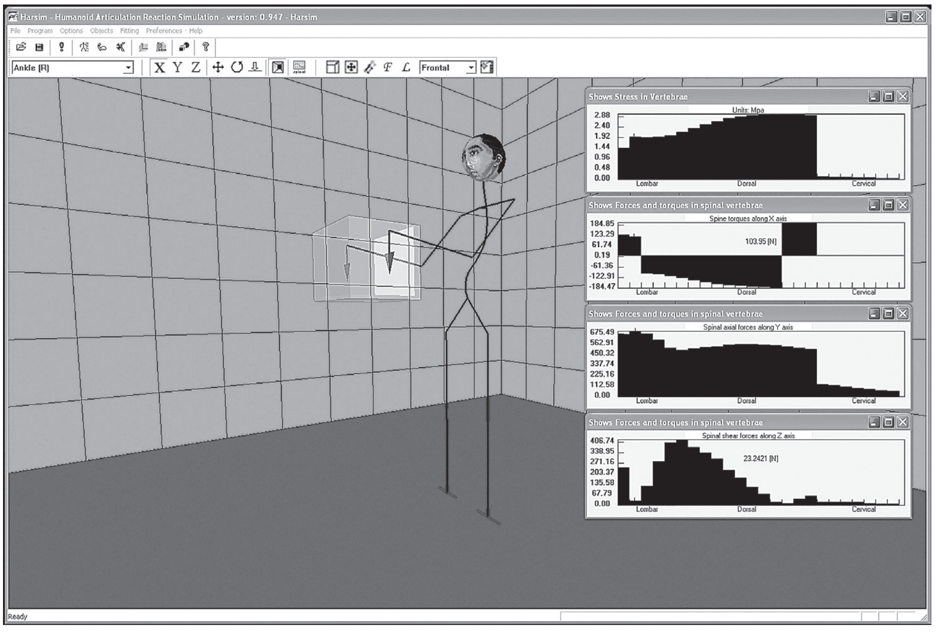

Figure 3. Print screen of the HARSIM model: from top to bottom, graphs of intradiscal pressures (MPa), bending moments (N/m) and axial and shear forces (N), at all levels of the spine. *Notes.* HARSIM = Humanoid Articulation Reaction Simulation.

TABLE 4. Estimated Intradiscal Pressure (IP) at Lumbar and Thoracic Levels of the Spine, in a Sitting Subject Holding 20 kg With Flexed Arms

Joint	S1–L5	L5–L4	L4–L3	L3–L2	L2–L1	L1–T12	T12–T11	T11–T10	T10–T9
IP (MPa)	1.60	2.1	2.1	2.1	2.2	2.3	2.3	2.4	2.5

Joint	T9–T8	T8–T7	T7–T6	T6–T5	T5–T4	T4–T3	T3–T2	T2–T1
IP (MPa)	2.5	2.5	2.4	2.4	2.3	2.3	2.1	2.1

Acar and Grilli obtained the same results by using a distributed loading pattern for the whole spine for various postures [24].

Polga, Beaubien, Kallemeier, et al. conducted an experimental study that measured the intradiscal pressure in the thoracic spine [25]. They measured a value of 2.77 MPa in a situation were the subject was sitting and holding 20 kg with flexed arms.

Table 4 shows estimated results for the same situation. The maximum estimated results for the thoracic spine (2.4 MPa) are lower than those measured (2.77 MPa). It should be noted that no detailed information about the anthropometric data of the subject and this posture, particularly data on the extent of the spinal curve, is available.

7. DISCUSSION

Regarding modelling realism, the results obtained with this software package for the situations described compare very closely with those Wilke et al. reported recently [23], and, thus, support a reasonable confidence in the correctness of the model. This validation was performed for the L4/L5 level in a predefined simple load manipulation only. Experimental data for other spine zones or data with complex loading situations, namely asymmetrical loads applied in the body or simultaneous body segments loaded with possible supports, were unavailable. Therefore, it is advisable to use the HARSIM model only to compare different simulation interaction strategies and optimize certain characteristics of a product, task procedure or physical workplace design. In this context, the model of the possibility of building simple virtual objects representing equipment or furniture that could be considered for handling difficulties is introduced. The ability to create animations allows the possibility to optimize tasks in light of the calculated data.

However, the model is limited in this particular aspect. The nature of the structural analysis finite-element approach employed was applied to static situations. It did not consider the effects of the body inertia due to acceleration. Therefore, for rapid movements, this approach may not be appropriate. Nevertheless, it is possible to validate the model's suitability for dynamic situations using heuristics. McGill and Norman calculated that the effects of inertia due to acceleration could be accounted for by increasing the calculated static data values by ~20% [26]. This correction can be used when comparing different simulation interaction strategies to select the best one for stress minimization.

This model can also be used as a first look in evaluating a particular interaction situation. The most accurate way to calculate stresses in a particular interaction allows the user insight into its severity in terms of loading stresses. This information can provide the opportunity to define intervention priorities and to use other methodologies to study a potential problem.

However, the present model differs from other models, which adopt the same approach [27, 28, 29]. The HARSIM model treats the vertebral column as a single beam of variable cross-sections with an effective Young's modulus value that is capable of guaranteeing its mechanical stability and integrity. This structural parameter is the combined result of intrinsic structural properties, such as the elasticity of the anatomical elements of the vertebral column, and extrinsic factors embodied in the activity of the associated musculature, which stabilize the action of the rib cage.

In relation to the kinematic proprieties of the HARSIM model, the adopted approach considers a movement as a succession of still postures.

Jäger and Luttman described a similar approach based on a model that used 15 segments to represent the vertebral column between the S1/L5 and the T3/T4 joints, plus another 15 segments to simulate the upper and lower limbs, the pelvis and the head-neck segment [30]. The present work differs from Jäger and Luttman's as each configuration of the vertebral column is generated by polynomial interpolation in the space of the intervertebral angles of each segment. This procedure takes place independently of the visual feedback provided by the structure which is generated in the three-dimensional Cartesian space and displayed on the screen. This approach offers the user greater flexibility for movement interpretation and postural design as well as for a more interactive and controlled process for configuration synthesis. The presented numerical model of the vertebral column kinematics is to be understood primarily as an instrument with which it becomes possible to reproduce or create complete movements of the torso in the form of sequences of frames or posture configurations. These postures afford evaluation of the mechanical response to loading. These frames may be used individually or "animated" collectively by the sequential projection of stored images. The main purpose of creating such a projection is to provide pertinent geometrical data for the evaluation of stresses on the vertebral column by loading during the performance of specific tasks. Furthermore, the information that the model provides would be quite difficult to obtain by other means. Other methods suffer from imprecision and limitations severely reducing the scope and validity of their use. If it is desired to know the details of a precise response of a given individual, the anthropometric profile can be easily introduced in the model once it is made available.

Typically, traditional models that calculate the probability of problems related to the musculoskeletal system have been developed to analyze manual materials handling tasks. However, these tools do not consider particular task conditions related to body support and loads applied in all body segments. This can be a limitation for the design of products, work situations or task procedures, which the HARSIM model should help to overcome. As noted by Chaffin [31], the efficient use of this knowledge requires an understanding of the variance in people's physiological and psychological capacities to cope with the physical requirements of various jobs. This means that the HARSIM model could be used during both an ergonomics analysis and intervention efforts, to best simulate the task conditions and interpret the study results. Since many workplace and product designers are often not ergonomists [32], the application of HARSIM should help to improve the quality of ergonomic job evaluations conducted in the field.

8. CONCLUSIONS

This paper reports the development and validation of a computer-based tool for estimating the reaction forces and bending moments in 43 human articulation joints. The model can also be used to assess the intradiscal pressure in the vertebral column in response to loading forces encountered during interactions with work objects or processes. The developed model was implemented in a self-contained interactive software package. The interactive model characteristics are achieved through intuitive interface menus and immediate visual presentation of the results on screen. The model simulation results compare favorably with the reported experimental data recoding data indicating its high quality. Compared to other models, HARSIM offers many advantages, particularly the ability to simulate more realistic human interactions with the workplace. Especially useful is the option of using either simple or complex loads, or considering different types of support in defining human body segments. The expected time to train human factors students in HARSIM is 12 h, including theoretical background and related exercises.

REFERENCES

1. Pheasant S, Haslegrave CM. Bodyspace: anthropometry, ergonomics, and the design of work. 3rd ed. Boca Raton, FL: Taylor & Francis; 2005.

2. Wickens CD. Engineering psychology and human performance. 2nd ed. New York, NY, USA: Harper Collins; 1992.
3. Suri JF, Marsh M Scenario building as an ergonomics method in consumer product design. Appl Ergon. 2000;31(2):151–7.
4. Feyen R, Liu Y, Chaffin D, Jimmerson G, Joseph B. Computer-aided ergonomics: a case study of incorporating ergonomics analyses into workplace design. Appl Ergon. 2000;31(3):291–300.
5. Butters LM, Dixon RT. Ergonomics in consumer product evaluation: an evolving process. Appl Ergon. 1998;29(1):55–8.
6. Porter JM, Freer M, Case K, Bonney MC. Computer aided ergonomics and workspace design. In: Wilson JR, Corlett EN, editors. Evaluation of human work: a practical ergonomics methodology. 2nd ed. London, UK: Taylor & Francis; 1995. p. 574–620.
7. Karwowski W, Genaidy AM, Asfour SS, editors. Computer-aided ergonomics. London, UK: Taylor & Francis; 1990.
8. Three Dimensional Static Strength Model Prediction Program (3DSSPP). User's manual. Version 6.0.2. Ann Arbor, MI, USA: University of Michigan; 2009.
9. Human Posture Analysis user's guide. Catia version 5 release 16. Vélizy-Villacoublay, France: Dassault Systèmes; 2005.
10. Jack user manual. Version 5.1. Unigraphics Solutions. 2006.
11. RAMSIS user guide version 3.7. Kaiserslautern, Germany; Tecmath; 2007.
12. Chaffin DB. Improving digital human modelling for proactive ergonomics in design. Ergonomics. 2005;48(5):478–91.
13. Spong MV, Vidyasagar M. Robot dynamics and control. New York, NY, USA: Wiley; 1989.
14. Kapandji IA. Physiologie articulaire, tronc et rachis [The physiology of the joints: the trunk and the vertebral column]. Paris, France: Maloine; 1975.
15. Weaver W Jr. Computer programs for structural analysis. Princeton, NJ, USA: Van Nostrand; 1967.
16. Goel VK, Kim YE, Lim TH, Weinstein JN. An analytical investigation of the mechanics of spinal instrumentation. Spine (Phila PA 1976). 1988;13(9):1003–11.
17. Shirazi-Adl A, Shrivastava SC, Ahmed AM. Stress analysis of the lumbar disc-body unit in compression: a three-dimensional non-linear finite element study. Spine (Phila PA 1976). 1984;9(2): 120–34.
18. Lavaste F, Skalli W, Robin S, Roy-Camille R, Mazel C. Three-dimensional geometrical and mechanical modelling of the lumbar spine. J Biomech. 1992;25(10):1153–64.
19. Takashima ST, Singh SP, Haderspeck KA, Schultz AB. A model for semi-quantitative studies of muscle actions. J Biomech. 1979; 12(12):929–39.
20. Cramer HJ, King Liu Y, von Rosenberg DU. A distributed parameter model of the inertially loaded human spine. J Biomech. 1976;9(3):115–30.
21. Clauser CE, McConville JT, Young JW. Weight, volume, and center of mass of segments of the human body (Report No. AMRL-TR-69-70). Dayton, OH, USA: Wright-Patterson Air Force Base; 1969. Retrieved October 2, 2012, from: http://www.dtic.mil/dtic/tr/fulltext/u2/710622.pdf
22. Da Silva KMC, Sayers BMcA, Sears TA, Stagg DT. The changes of configuration of the rib cage and abdomen during breathing in the anaesthetised cat. J Physiol (Lond). 1977;266(2):499–521. Retrieved October 2, 2012, from: http://www.ncbi.nlm.nih.gov/pmc/articles/PMC1283577/pdf/jphysiol00817-0285.pdf
23. Wilke H, Neef P, Hinz B, Seidel H, Claes L. Intradiscal pressure together with anthropometric data—a data set for the validation of models. Clin Biomech (Bristol, Avon). 2001;16 Suppl 1:S111–26.
24. Serpil Acar B, Grilli SL. Distributed body weight over the whole spine for improved inference in spine modelling. Comput Methods Biomech Biomed Engin. 2002; 5(1):81–9.
25. Polga DJ, Beaubien BP, Kallemeier PM, Schellhas KP, Lew WD, Buttermann GR, et al. Measurement of in vivo intradiscal pressure in healthy thoracic intervertebral discs. Spine (Phila PA 1976). 2004;29(12): 1320–4.

26. McGill SM, Norman RW. Dynamically and statically determined low back moments during lifting. J Biomech. 1985;18(12): 877–85.
27. Belytschko TB, Andriacchi TP, Schultz AB, Galante JO. Analog studies of forces in the human spine: computational techniques. J Biomech. 1973;6(4):361–71.
28. Stokes IA, Gardner-Morse M. Analysis of the interaction between vertebral lateral deviation and axial rotation in scoliosis. J Biomech. 1991;24(8):753–9.
29. Schultz AB, Galante JO. A mathematical model for the study of the mechanics of the human vertebral column. J Biomech. 1970; 3(4):405–16.
30. Jäger M, Luttmann A. The load on the lumbar spine during asymmetrical bi-manual materials handling. Ergonomics. 1992;35(7–8):783–805.
31. Chaffin DB. The evolving role of biomechanics in prevention of overexertion injuries. Ergonomics. 2009;52(1):3–14.
32. Marras WS. The future of research in understanding and controlling work-related low back disorders. Ergonomics. 2005; 48(5):464–77.