Aalborg Universitet

Musculoskeletal modelling of low-frequency whole-body vibrations

Rasmussen, John; Andersen, Michael Skipper

Published in:
Proceedings of BNAM 2012

Publication date:
2012

Document Version
Early version, also known as pre-print

BNAM2012
Joint Baltic-Nordic Acoustics Meeting
June 18th - 20th 2012 Odense, Denmark

Musculoskeletal modelling of low-frequency whole-body vibrations

John Rasmussen and Michael Skipper Andersen
M-Tech, Aalborg University, Fibigerstræde 16, DK-9220, Aalborg Ø, Denmark, jr@m-tech.aau.dk

This paper presents a musculoskeletal model for assessment of the effect of low-frequency whole-body vibrations on the human body. It is a basic assumption behind the model that the vibrations are slow enough to allow the central nervous system to respond to them in terms of muscle activations for stabilization and posture adjustments. It is concluded that the approach extends the availability of models for assessment of workspaces and other environments in which the human body is subjected to slow vibrations and relies on support points to maintain posture and stability.

1 Introduction

The human body responds differently to different vibration frequencies and working requirements. For sufficiently high frequencies, the central nervous system and muscle activation dynamics are too slow to respond to individual oscillations. The body copes with such vibrations by modulating joint stiffness and compliance to allow the body to take up the vibrations while maintaining the required posture in an overall sense. The modelling of responses, whether by hardware dummies or mathematics, is therefore typically addressed with passive models [1].

However, at lower frequencies, and in the presence of a working task, such as driving a vehicle over rough terrain, muscles are forced to respond to each oscillation to maintain the necessary working posture. This brings dynamic equilibrium of the human body and the active function of the sensory-motor system into the simulation loop and complicates matters significantly compared to passive models. Ideally, the formulation of a model would comprise (i) a goal function or quest modelling the desire of the body to accomplish the working task, (ii) sensory feedback, (iii) processing delays and (iv) muscular actions to cope with the working task. However, such a model would be overwhelmingly complicated and many of the basic parameters for modelling the sensory-motor system in a forward dynamics paradigm are unknown or difficult to obtain.

Instead, this paper presents a musculoskeletal model that covers the domain directly opposite the passive models, i.e. vibrations that require active muscle response and are slow enough for the latency of the sensory-motor system not to influence the solution significantly. In such a case, dynamic equilibrium of the redundant muscle system is involved and muscle actions must be computed, but it can be presumed that the muscle actions arise as instant responses to changes of directions and magnitudes of gravity and inertia forces. This model involves active muscle recruitment and the muscle actions are dynamic, but the model can rely on the assumption that muscle actions are optimal and instant, and this makes it mathematically and numerically feasible.

Similarly to the benefit of passive models, this enables computational investigation of the effect of the environment we are placing the body in, for instance answering the question of which are the best locations of support points?

2 Methods

A model of a seated human was developed using the AnyBody Modeling System ver. 5.2 (AnyBody Technology, Aalborg, Denmark). The model (Figure 1) is a detailed representation of the human musculoskeletal physiognomy comprising approximately 700 individually activated muscles. In the following, the inner workings of the model are briefly explained.

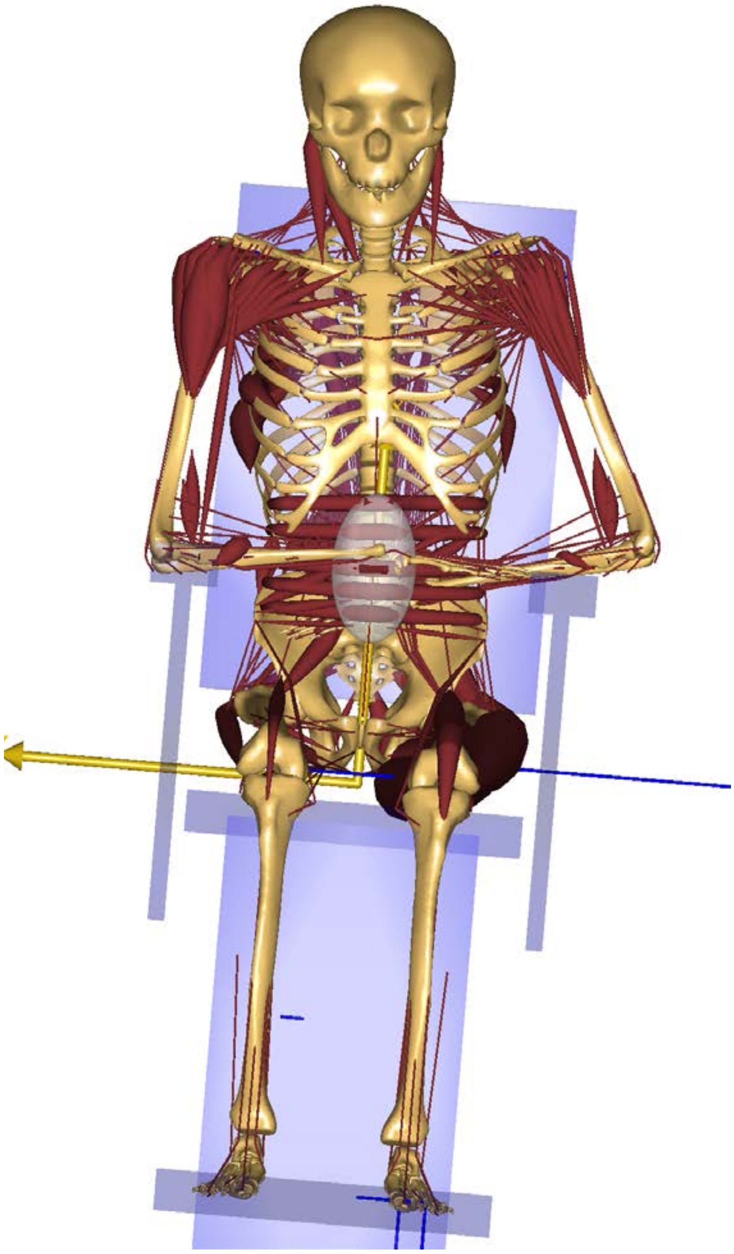

Figure 1: Musculoskeletal model of a human body in lateral forced harmonic oscillation of 1 Hz. Bulging of muscles is a graphical indication of the current muscle force.

2.1 Equations of motion

The model assumes rigid segments and perfect articulation at the joints and is therefore in essence a multibody mechanics model. It turns out to be beneficial to work with segment-fixed reference frames because it places bony landmarks and other anatomical features at fixed local coordinates, and this choice of formulation leads to the Newton-Euler equations, which are valid for body-fixed reference frames originating in the inertial centres of the segments. A slightly generalized version of these will allow more freedom in the choice of local reference frames for the segments, and we may arrive at the following formulation:

$$\mathbf{M}\dot{\mathbf{v}} = \mathbf{f}^{(applied)} + \mathbf{f} - \mathbf{f}^{(v)} \quad (1)$$

where $\mathbf{M}$ is the mass matrix, $\mathbf{v}$ is the velocity vector comprising linear as well as angular velocities with the dot designating differentiation with respect to time, $\mathbf{f}^{(applied)}$ is the applied forces including gravity, $\mathbf{f}$ are the internal reaction forces, and $\mathbf{f}^{(v)}$ are the velocity-dependent forces, i.e. centrifugal forces, gyroscopic forces, Coriolis forces and damping. Please notice that equilibrium equations (1) are fully dynamic and make no assumptions about static or quasi-static conditions.

2.2 Kinematics

For any mechanism that is not very simple, computation of the individual terms of (1) is a complex matter requiring extensive coordinate transformations of vectors and matrices in three-dimensional space. The analysis of kinematics is particularly challenging because closed chains are inherent in the human body and in the situations we may wish to analyse, i.e. human bodies supported on multiple points by an environment.

The closed chain kinematics complicates the solution significantly because some amount of implicit equations must be used. In the AnyBody Modeling System, a so-called Cartesian method [2] relying on a general set of nonlinear, implicit equations, is used to solve the kinematics:

$$\mathbf{\Phi}(\mathbf{q}, t) = \mathbf{0} \tag{2}$$

where $\mathbf{q}$ are the system coordinates comprised of locations and rotations of all the segments in the system, t is time, and $\mathbf{\Phi}=\mathbf{0}$ is the system of all kinematic constraints in the system, typically joints and motion drivers, regardless of the system topology. The consequence of this approach is that there is no distinction between open and closed chain kinematics. This implicit formulation provides the maximum amount of generality allowing for any topological configuration of the system at the cost of computational efficiency, owing to the fact that equations (2) are generally nonlinear. It turns out, however, that the kinematic analysis is rarely the bottleneck of the computations.

Nonlinear equations may have a single solution, multiple solutions or no solution at all, even in the case where the number of equations matches the number of system coordinates. Cases of no solution arise when incompatible kinematic constraints are imposed, for instance requiring the hand to reach a point beyond the length of the arm. However, a more common cause for empty solution sets is the presence of more constraint equations than system coordinates, and this is a frequent occurrence when driving the model with motion capture data. The remedy is to make use of solvers that minimize the right hand side of (2) without requiring complete fulfilment of the equations, and this actually leads to opportunities to exploit the redundancy of equations to determine unknown system parameters, such as functional joint locations or joint axis orientations and described by Andersen et al. [3].

2.3 Kinetics

Any realistic musculoskeletal model as the one depicted in Figure 1 has more actuators in the form of muscles than it has degrees of freedom. This leads to an indeterminacy or redundancy in the muscle system that disallows muscle forces to be determined from the equilibrium equations alone. This means that any form of solution method for the Newton-Euler equations (1) must include a way of selecting among infinitely many possible solutions to the equations.

We observe, as Erdemir et al. [4], that it is very difficult or impossible to measure muscle forces. However, knowing the movement including the kinematic boundary conditions allows for determination of $\mathbf{f}^{(v)}$. Subsequently, the unknown muscle and reaction forces are collected in $\mathbf{f}$. The unknown forces can now be resolved in an inverse dynamics computation. This approach is sometimes referred to as 'static optimization', which is somewhat misleading given that equation (1) contains all dynamic terms.

For the purpose of inverse dynamics, (1) is transferred to this form:

$$\mathbf{Cf} = \mathbf{r} \tag{3}$$

where

$$\mathbf{f} = \left[\mathbf{f}^{(\mathbf{M})}\mathbf{f}^{(\mathbf{R})}\right]^T \tag{4}$$

and $\mathbf{r}$ is a vector of externally applied forces and body forces as defined in (2). The first part, $\mathbf{f}^{(\mathbf{M})}$, contains muscle forces, while the latter, $\mathbf{f}^{(\mathbf{R})}$, contains the joint reactions. This division reflects that muscle forces require metabolism, and therefore resources, while the joint reactions are not perceived as requiring effort. The system of linear equations (3) is usually under-determinate in the sense that it has more unknowns than equations and, hence, has infinitely many solutions. This follows from the fact the human body has significantly more muscles than degrees of freedom. We, therefore, presume that the solution to (3), i.e. $\mathbf{f}$, can be found by solving the following optimization problem:

Minimize

$$G(\mathbf{f}^{(\mathbf{M})}) \tag{5}$$

subject to

$$\mathbf{Cf} = \mathbf{r} \tag{6}$$

$$f_i^{(M)} \geq 0, \quad i = 1..n^{(M)} \tag{7}$$

The solution to this mathematical program is a set of muscle forces, $\mathbf{f}^{(\mathbf{M})}$, and joint reaction forces, $\mathbf{f}^{(\mathbf{R})}$, that minimize the objective function *G*, while honouring the constraints. Notice that (6) is the set of equilibrium equations (3), i.e. feasible solutions must satisfy equilibrium.

It is, therefore, plausible that the cost function, *G*, depends on the muscle forces but not necessarily on the joint reactions. Finally, please notice the additional non-negativity condition for each element of $\mathbf{f}^{(\mathbf{M})}$. This maintains the physiological limitation that muscles cannot push.

One of the obvious and popular choices of cost function *G* is a polynomial form:

$$G = \sum_{i=1}^{n^{(M)}} \left(\frac{f_i^{(M)}}{N_i} \right)^p \tag{8}$$

where N_i, are normalization factors. It is a reasonable assumption that large muscles pull more load than small muscles, and this indicates that N_i should somehow express the strength of the *i*'th muscle, for instance the cross sectional area of the muscle. There is good experimental evidence [5] for the notion that unrealistic predictions of muscle forces will result from failure to consider the mutual strengths of the muscles in the formulation of the problem. Thus, N_i represents the strength or maximum voluntary contraction (MVC) of the *i*'th muscle, and each term in (8) becomes the relative load of muscle *i*.

The degree, *p*, of the polynomial is more disputable except for the general agreement that $p=1$ is physiologically unrealistic because it does not produce the synergism between the muscles that can be observed experimentally. Any value of $p > 1$ will result in synergy between the muscles and also some extent of antagonistic muscle forces in complex systems. It is possible to show that this antagonism arises as a result of either bi-articular muscles or of three-dimensional joints. It is also possible to show mathematically that the amount of synergism between the muscles increases with *p*. A special case occurs for $p \to \infty$. In this case, the muscle recruitment problem becomes a minimum fatigue problem. In the following we shall use $p = 3$.

2.4 The model

The model comprises a human part and an environment, i.e. the seat. The human part stems from the AnyScript Managed Model Repository (AMMR) version 1.4.1, which is available from www.anyscript.org. AnyScript is the name of the scripting language in which the model is developed. The model is copyrighted but developed in an open source setting that allows users to read and modify and contribute to any part of the source code of the model. Model parameters are derived from cadaver studies reported in the literature, for instance [6-9], and validated in a number of different studies [10-13].

The seating part comprises the seat pan, backrest, foot rest and leg rest, where the latter is vertical in this model and, therefore, has little influence. The footrest may be thought of as mimicking the floor. Contact points are created between the human body and the chair on bony prominences, such as the ischial tuberosities, and on other places that are likely to come into contact with the chair. On these points, contact conditions allowing only for compression forces and Coulomb friction enable computation of the contact forces and fulfilment of the equilibrium conditions that balance the body. The seat is connected to ground via a joint placed 2 m below the head of the human occupant, and the inverse dynamics nature of the analysis allows the seat to be rocked about this point in random patterns, but for the purpose of the subsequent example, a sinusoidal lateral oscillation with a frequency of 1 Hz and amplitude of 5.7 degrees (0.1 radian) is used. The human body follows the rocking of the seat except for the upper body, which flexes laterally against the movement as shown in Figure 1 to maintain the angle of the head stable.

The chair is also equipped with armrests supporting the elbows of the occupant, but only in the vertical direction. In the interest of simplicity, this support is implemented as a reaction force enabling the elbows to retain support as they move with the upper body posture adapting to the change of chair position. The model has the forearms arms crossed in front of the abdomen, and to avoid collision between the arms, the posture is slightly asymmetrical as will be evident from the results section.

The model is analysed for a number of properties investigating the consequence of the rocking as shown in the subsequent section.

3 Results

The complexity of the model means that the amount of results from each analysis is very comprehensive. The model computes all kinematics and all forces in muscles and joints, including the connections between the body and the environment, for each time step in the analysis. A model without arm support was initially attempted, but it turns out that this model cannot be balanced under the influence of the vibrations. It will inevitably tumble sideways because the contact conditions with the seat are insufficient to keep it stable as the lateral acceleration grows in phase with the displacement. In the inverse dynamics paradigm, this behaviour comes out as an inability to solve the equilibrium conditions. The arm rests were subsequently added and the arms were placed in the occupant's lab such that the forearms can make contact with the armrests and contribute to the stabilization of the body. Other ergonomically relevant positions are holding on to a steering wheel or a safety bar like in an amusement park ride.

3.1 Kinematics

The head movement, more precisely the acceleration, is of ergonomic concern and not easily assessed in the absence of a detailed model because the upper body moves to compensate to some extent for the rocking motion (Figure 2). The analysis reveals head accelerations slightly below 0.6 g.

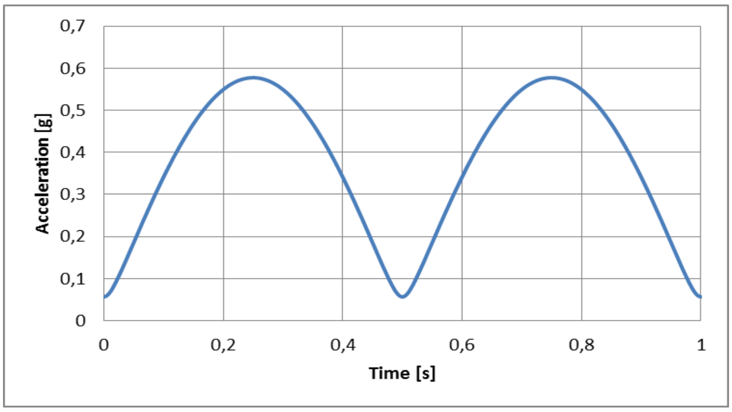

Figure 2. Absolute acceleration in g of the occupant's head.

3.2 Spinal loads

Concerns have earlier been raised about the influence of speed bumps on the spinal loads of bus drivers and other vehicle occupants [14, 15]. Experimental assessment of the influence of vibrations on the spinal disk forces is technically difficult and ethically questionable due to its invasive nature. On the other hand, computational investigation is difficult because the actions of the muscles contribute significantly to the disk forces. In particular, we can notice that these muscle forces may increase the spinal disk forces significantly even in the absence of a vertical acceleration. The detailed and validated spine model of AnyBody allows for a computation of the spinal disk loads under the influence of the lateral vibrations imposed on this model. The result for the L4/L5 disk is shown in Figure 3.

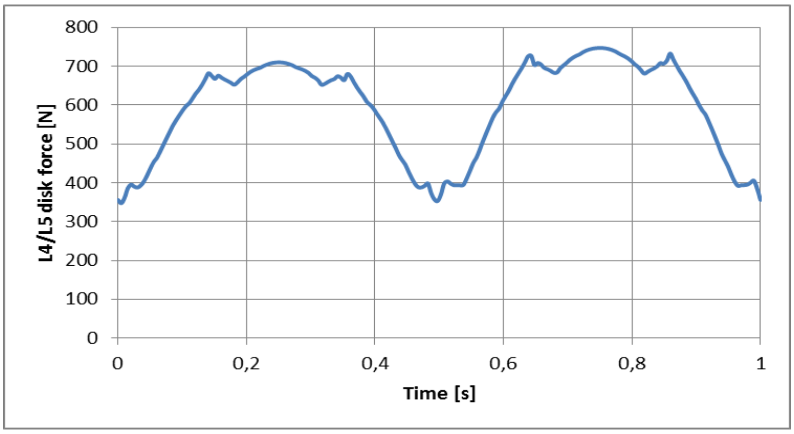

Figure 3. Spinal disk forces between L4 and L5.

Figure 3 reveals that, despite the movement inducing very little vertical acceleration, the compression of the L4/L5 disk is approximately doubled in the lateral extremes of the posture compared with the value in the upright posture. This effect is caused by the necessity of the muscles to contract to maintain stability of the body. The slight asymmetry between the two peaks is caused by differences in postures between the right and left arms. Comparison with published data for other tasks [13, 16] reveals that this disk compression is significant and roughly corresponds to the load of lifting a load of 20 kg in a standing posture.

3.3 Muscle Effort

Muscle actions require effort and high or sustained muscle actions will be perceived as discomfortable due to development of fatigue or ultimately injury. It is a reasonable assumption that perception of discomfort increases with the muscle activity. In a muscle system, we can, therefore, use the relative load of the highest loaded muscle as a measure of discomfort:

$$\max_i \left(\frac{f_i^{(M)}}{N_i} \right) \tag{9}$$

i.e. the envelope of muscle activity in fractions of the muscle's strength. For the whole-body vibration of the present example, this measure is shown in Figure 4.

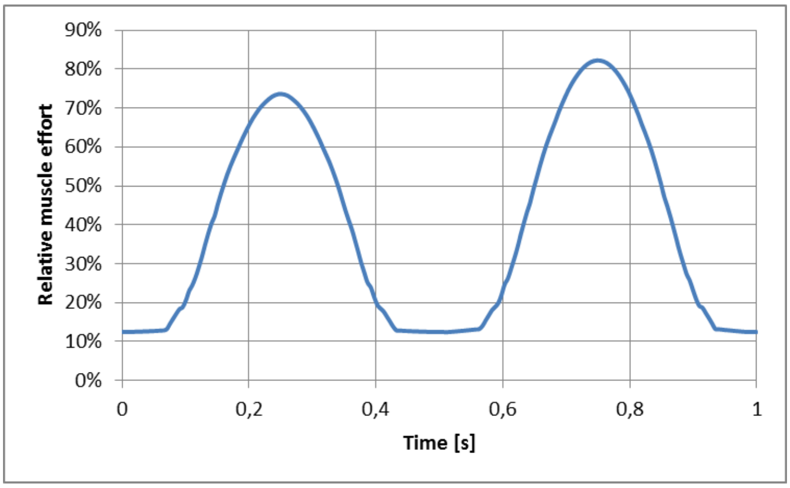

Figure 4. Relative muscle effort in per cent of maximum voluntary contraction over the cycle.

4 Discussion

The method presented in this paper extends the frequency domain in which assessment of the effect of slow body vibrations can be performed with computational models. The extension covers the low-frequency domain and includes the activation of the living muscles. Unfortunately, the method does not bridge the entire gap to the fast vibration domain, where passive models can be used for the assessment. In an intermediate range of frequencies, the passive-elastic properties as well as the active control of posture will play significant roles and there is currently no method available to handle this combination.

The example treated in this paper clearly demonstrates that the computational treatment of the problem allows for assessment of body loads that are infeasible and unsafe to investigate by experimental means. We find that for a 1 Hz lateral oscillation with 5.7 degrees amplitude, head accelerations, despite posture correction, approaches 0.6 g, spinal loads approach the load of lifting a 20 kg box, and the muscle effort required to maintain the posture exceeds 80% of the body strength. All of these numbers indicate that vibrations of this nature create an unsafe working environment if sustained for an extended period of time.

More realistic assessments can be obtained by replacing the presumed lateral vibration by recorded time-histories of motions of actual seats in heavy machinery, fairground rides or sea and air vessels where motions such as these are likely to occur.

References

[1] M.J. Griffin, The validation of biodynamic models, *Clin. Biomech*. 16, 2001, S81-S92.

[2] P.E. Nikravesh, *Computer-aided analysis of mechanical systems*, Prentice Hall, Englewood Cliffs, New Jersey, 1987.

[3] M.S. Andersen, M. Damsgaard, B. Mac-Williams, J. Rasmussen, M.S. Andersen, M. Damsgaard, B. Mac-Williams, J. Rasmussen, A computationally efficient optimisation-based method for parameter identification of kinematically determinate and over-determinate biomechanical systems, *Comput. Methods Biomech. Biomed. Engin*. 13, 2010, 171-183.

[4] A. Erdemir, S. McLean, W. Herzog, A.J. van den Bogert, Model-based estimation of muscle forces exerted during movements, *Clin. Biomech*. 22, 2007, 131-154.

[5] B.I. Prilutsky, R.J. Gregor, Analysis of muscle coordination strategies in cycling, *IEEE Transactions on Rehabilitation Engineering* 8, 2000, 362-370.

[6] F.C.T. Van der Helm, H.E.J. Veeger, G.M. Pronk, L.H.V. Van der Woude, R.H. Rozendal, Geometry parameters for musculoskeletal modelling of the shoulder system, *J. Biomech*. 25, 1992, 129-144.

[7] L. Hansen, M.D. Zee, J. Rasmussen, T.B. Andersen, C. Wong, E.B. Simonsen, L. Hansen, M.D. Zee, J. Rasmussen, T.B. Andersen, C. Wong, E.B. Simonsen, Anatomy and biomechanics of the back muscles in the lumbar spine with reference to biomechanical modeling, *Spine* 31, 2006, 1888-1899.

[8] M.D. Zee, L. Hansen, C. Wong, J. Rasmussen, E.B. Simonsen, M.D. Zee, L. Hansen, C. Wong, J. Rasmussen, E.B. Simonsen, A Generic Detailed Rigid-Body Lumbar Spine Model, *J. Biomech*. 40, 2007, 1219-1227.

[9] M.D. Klein Horsman, H.F.J.M. Koopman, F.C.T. van der Helm, L.P. Prosé, H.E.J. Veeger, Morphological muscle and joint parameters for musculoskeletal modelling of the lower extremity, *Clinical Biomechanics* 22, 2007, 239-247.

[10] M.S. Andersen, M. de Zee, S. Dendorfer, B. MacWilliams, J. Rasmussen, M.S. Andersen, M. de Zee, S. Dendorfer, B. MacWilliams, J. Rasmussen, Validation of a detailed lower extremity model based on the Klein Horsman data set, *Proceedings of the 12th International Symposiumon Computer Simulation in Biomechanics, Cape Town, South Africa* 2009 .

[11] C. Schwartz, M.E. Lund, M. de Zee, J. Rasmussen, C. Schwartz, M.E. Lund, M. de Zee, J. Rasmussen, Prediction of knee loads using a lower extremity model based on the Klein Horsman data set, *Proceedings of the ASME [American Society of Mechanical Engineers] 2010 Summer Bioengineering Conference* , 2010, 269-270.

[12] A. Nolte, P. Augat, J. Rasmussen, A. Nolte, P. Augat, J. Rasmussen, Analysis of the muscle and joint forces in the shoulder joint using the anybody simulation model, *J. Biomech*. 41, 2008, 492.

[13] J. Rasmussen, M. de Zee, S. Carbes, J. Rasmussen, M. de Zee, S. Carbes, Validation of a biomechanical model of the lumbar spine, *Proceedings of the XXII Congress of the International Society of Biomechanics*, Cape Town, South Africa, 2009 .

[14] S.J. Triantafyllou, S.D. Gertzbein, Flexion distraction injuries of the thoracolumbar spine: A review, *Orthopedics* 15, 1992, 357-364.

[15] J. Giacomin, Some observations regarding the vibrational environment in child safety seats, *Appl. Ergon*. 31, 2000, 207-215.

[16] H. Wilke, P. Neef, M. Caimi, T. Hoogland, L.E. Claes, New in vivo measurements of pressures in the intervertebral disc in daily life, *Spine* 24, 1999, 755-762.