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Dynamic Modelling of Paraplegic Person's Standing-Up

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Abstract. The simulation results of the standing-up process in paraplegic persons rising from the sitting to the standing position by the help of functional electrical stimulation (FES) and arm support are presented. A six segments planar rigid body dynamic model was designed and simulated in the Matlab-Simulink environment. The aim of the model was to investigate different rising strategies and determine how much effort they require. The model behaviour was compared to experimental results.

Key words: standing-up, FES, modelling, simulation

Dinamični model vstajanja paraplegične osebe

Povzetek. Razvit je bil dinamični model vstajanja paraplegične osebe, ki ima stimulirane ekstenzorje kolena in se opira na roke. Telo je bilo modelirano s šestimi togimi segmenti. Namen modela je primerjava strategij vstajanja glede na zahtevane sile in navore v sklepih. Obnašanje modela je bilo preverjeno s primerjavo z meritvami vstajanja skupine paraplegičnih oseb.

Ključne besede: vstajanje, FES, modeliranje, simulacija

1 Introduction

Rising from a sitting to a standing position is one of the most frequently performed human activities. Because of the inability to stand up, handicapped, elderly, and paralyzed subjects are confined to wheel chairs and beds. Restoration of the activity of rising from a chair is essential for independent living. Studies of biomechanics during a sit-to-stand process were mostly performed in healthy subjects. However, only a few studies on standing-up of paraplegic subjects who use functional electric stimulation (FES) of their paralyzed extremities and arm support were performed due to the difficulty of investigating with impaired subjects [1]. It is for this reason that new methods of FES supported standing-up are tested mainly by simulations [2,3].

A dynamic model of a paraplegic standing-up is presented. FES of knee extensors and use of arms were modelled. The development of the model was based on measurements of rising from a chair of a group of paraplegic subjects [4]. The model was designed for the analysis and comparison of different standing-up strategies by comparing the maximal required joint moments and arm support forces. The Matlab-Simulink* and SD/FAST† software packages were used to implement the model.

2 Software

The Matlab-Simulink environment was used to simulate the sit-to-stand process. Deriving the dynamic equations of motion takes considerable time because of the complexity of the system being simulated. For this reason the SD/FAST program was used. The SD/FAST allows a dynamic analysis of mechanical systems which are represented as a chain of rigid body segments interconnected by joints. Inverse dynamics, direct dynamics or combination of both problems can be computed.

Figure 1 illustrates the way SD/FAST was included in the Matlab-Simulink environment. The user creates a file (*body.sd*) that contains the parameters of each segment and the description of the system structure. This file is used by SD/FAST to derive the equations of motion which describe the dynamic behaviour of the system. They are given in the form of C rutines (*body_s.c*, *body_d.c*). The next user's task is to write a procedure in the programming language C which calls the SD/FAST rutines in an appropriate sequence (*model.c*). The C procedure is compiled (*model.mex*) and included in Simulink. The main program is a Simulink file (*sim.m*). Two additional Matlab programs were written: *prepare.m* computes the data necessary to run *sim.m*, while *display.m* analyses the *sim.m* results.

The development of dynamic equations of motion is

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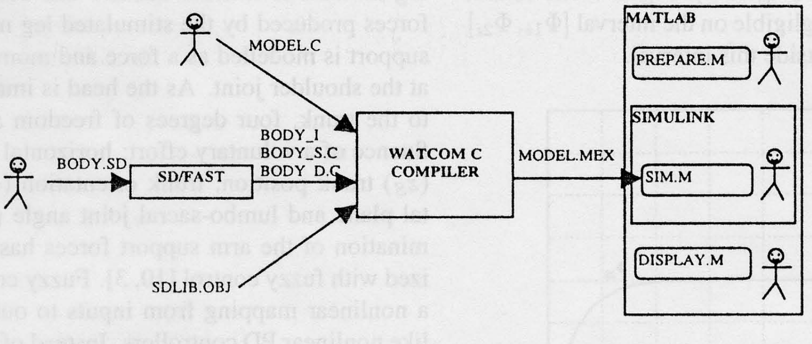

Figure 1. Simulation environment of the standing-up process. Names of programs are written in bold. Human symbol denotes user created files

based on the Kane's formulation [5].

The functions are distributed among progams as follows:

1. SD/FAST: computation of derivatives and outputs.
2. Simulink: integration of states, display and animation.

3 A model of the sit-to-stand process

The human body is described with a 2-dimensional model in the sagittal plane. The model consists of six rigid segments: foot, shank, thigh, pelvis, trunk and head (figure 2). The segments are interconnected by rotational pin joints.

In order to derive the dynamic equations of motion, four parameters must be given for each segment: length, mass, center of mass and inertia matrix. The characteristics of the model are similar to the characteristics of patients who participated in the investigation [4]. The trajectories of approximate joint centers were measured in [4]. Segment lengths are set to the mean value of the distance between two centers of the adjacent joints. The segment center of the mass is located on the straight line interconnecting two adjacent joint centers. The distance from the proximal joint to the center of the mass is given in [6]. The whole body mass was measured with a force plate. The masses and inertia matrices are calculated from the body mass according to the proportional anthropometric model [6].

The equations of motion were automatically generated by the SD/FAST program. These equations must be completed with proper forces and moments acting on the kinematic chain. A program in the programming language C was written. It calculates forces and moments applied in each integration step. Thus, models of passive joint moments, sitting reaction force and arm force were programmed in the C language. The number of studies describing loads acting on the human body is rather small.

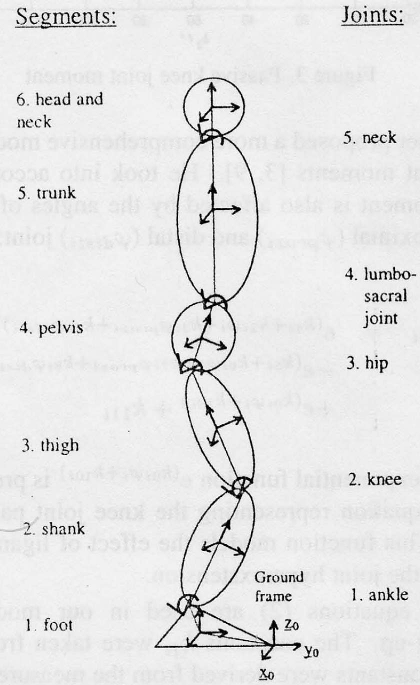

Figure 2. Segmental model of human body

As the paraplegic subject is not able to voluntarily control the muscles of his lower extremities, only passive joint moments act around these joints. A passive joint moment occurs when all muscles crossing the joint are relaxed. A passive joint moment is caused by deformation of the tissues surrounding the joint, such as relaxed muscles, ligaments, tendons, skin, etc. Passive moments are often modelled with a nonlinear spring and a linear damper. The elastic component is described by the sum of two exponential functions [7, 8]:

$$M_{ei} = k_{1i}e^{-k_{2i}(\varphi_i - \Phi_{1i})} - k_{3i}e^{-k_{4i}(\Phi_{2i} - \varphi_i)} \qquad (1)$$

In Eq. (1), φ_i is the joint angle while k_{1i}, k_{2i}, k_{3i}, k_{4i}, Φ_{1i}, Φ_{2i} are constants. Each exponential function describes an increase of the passive moment at one side of the joint range of motion (Figure 3). Figure 3 shows that the passive moment is negligible on the interval $[\Phi_{1i}, \Phi_{2i}]$ and increases quickly outside this interval.

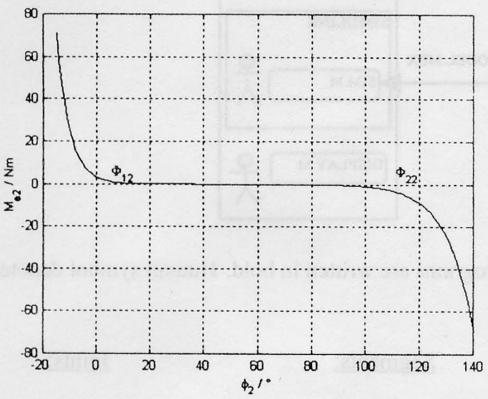

Figure 3. Passive knee joint moment

Riener proposed a more comprehensive model of passive joint moments [3, 9]. He took into account that a joint moment is also affected by the angles of the adjacent, proximal (φ_{proxi}) and distal (φ_{disti}) joint:

$$\begin{aligned} M_{ei} \;=\; & e^{(k_{1i}+k_{2i}\varphi_i+k_{3i}\varphi_{proxi}+k_{4i}\varphi_{disti})} \\ & -e^{(k_{5i}+k_{6i}\varphi_i+k_{7i}\varphi_{proxi}+k_{8i}\varphi_{disti})} \\ & +e^{(k_{9i}\varphi_i+k_{10i})} + k_{11i} \end{aligned} \tag{2}$$

The exponential function $e^{(k_{9i}\varphi_i+k_{10i})}$ is present only in the equation representing the knee joint passive moment. This function models the effect of ligaments preventing the joint hyperexstension.

The equations (2) are used in our modelling of standing-up. The constants k_{ji} were taken from [3, 9]. These constants were derived from the measurement of a healthy subject.

An improved way to derive a model of a passive joint moments model is to find such constants that the equations (1) or (2) would match the inverse dynamics resulting from the experimental data [4]. The equations (1) are more suitable for this purpose as they include fewer constants. However, this is not a straightforward procedure because of the spasticity and spasms present in spinal cord injuried persons.

The damping component around the joint is described by the equation:

$$M_{di} = -k_{12i}\dot{\varphi}_i \tag{3}$$

The value of the constant k_{12i} can be found in [3, 7, 8]. The total passive joint moment is:

$$M_i = M_{ei} + M_{di} \tag{4}$$

Paraplegic subjects make use of their arms during rising in order to achieve balance and compensate for low forces produced by the stimulated leg muscles. The arm support is modelled as a force and moment vector acting at the shoulder joint. As the head is immovably attached to the trunk, four degrees of freedom are under the influence of a voluntary effort: horizontal (y_S) and vertical (z_S) trunk position, trunk orientation (φ_T) in the sagittal plane and lumbo-sacral joint angle (φ_L). The determination of the arm support forces has often been realized with fuzzy control [10, 3]. Fuzzy controllers provide a nonlinear mapping from inputs to outputs and behave like nonlinear PD controllers. Instead of fuzzy controllers four independent PID controllers were applied (Figure 4). As the behaviour of a PID controller is more predictable, it is easier to adjust its parameters properly. The controller outputs are: horizontal arm force (F_{Ay}), vertical arm force (F_{Az}), arm moment (M_{Ax}) and lumbo-sacral joint moment (M_L).

It was assumed that the patient tries to follow a trajectory he had learned. The reference trajectories used for the model verification were obtained from the experimental data [4]. In order to make the experimenting with the model more flexible, the reference trajectories were approximated with a trapezoidal velocity profile (Figure 5). Thus, only four parameters were needed to completely describe a trajectory: initial value, final value, transition time and acceleration.

Modelling of the interaction of the human body with a seat is more complex than the models developed by others [3, 11]. During the sit-to-stand process two types of a contact of the body with the seat occur. In the beginning, there is a surface contact (Figure 6). We assume that the thigh flattens equally along its length.

The measurements showed that the component of the seat reaction force which is perpendicular to the seat surface can be represented with an exponential function:

$$F_N = Ae^{-\frac{l-l_0}{B}} - Cv_N \tag{5}$$

where F_N is the normal component of the seat reaction force and l is the distance from the hip joint to the seat surface. Constants A, B and l_0 are chosen for each patient individually in such a way that the function F_N approximates the measured force. In order to get a realistic response of the seat model, a damping element is added. Here, v_N is the normal thigh velocity. The constant C is assigned arbitrarily.

The friction force F_T is paralell to the seat surface. It is assumed that during quiet sitting a paraplegic subject does not use an arm support. As a consequence, the seat force F_S acts at such a point of the thigh that it passes through the center of the mass of the upper body. When the projection of the center of the mass moves outside the seat supporting area, the thigh turns around the seat edge. A linear contact of the body with the seat arises. The

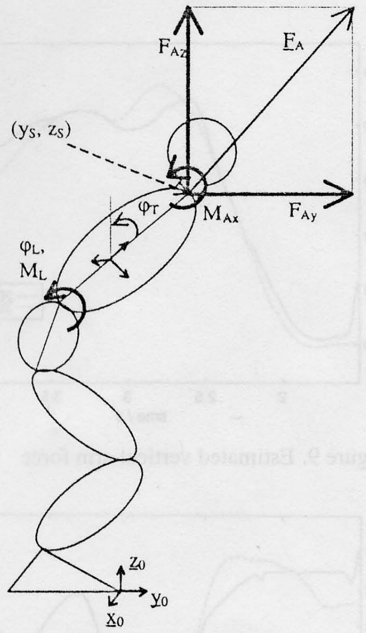

Figure 4. Paraplegic person's voluntary effort

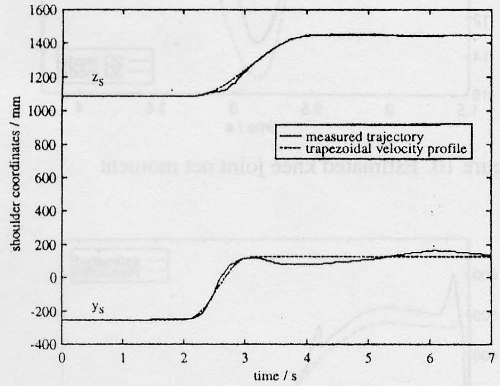

Figure 5. Approximation of the measured shoulder trajectory with the trapezoidal velocity profile

reaction force is now made up of two components; one is perpendicular to the thigh, the other is the friction force which is parallel to the thigh.

Since paraplegic subjects are not capable of voluntary control over the leg muscles, the knee joint moment supporting the standing-up is achieved by means of FES. Models of stimulated lower extremities, taking into account the dynamics of each particular leg muscle, include a considerable number of parameters [3]. It is difficult to adapt such parameters to the subjects which participated in the investigation [4]. The amount of the muscle force depends on the muscle length, rate of contraction and fatigue. The moment at the knee joint is the sum of the moments produced by each muscle. A moment arm of a muscle is a function of the joint angle.

This problem was solved with the assumption, that the

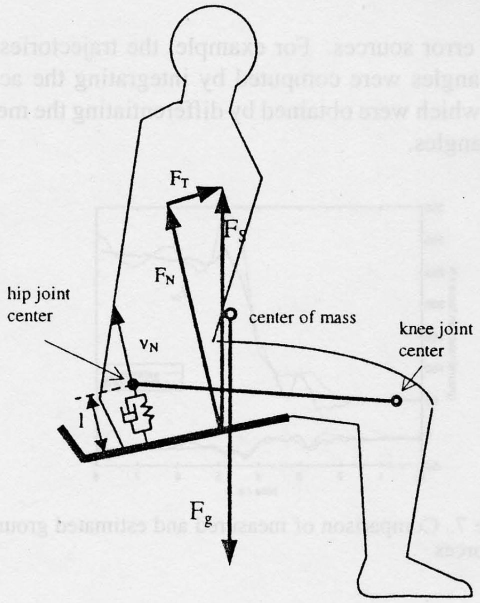

Figure 6. The patient's body in the surface contact with the seat

knee joint moment produced by the electrical stimulation depends mainly on the joint angle and its rate of change. The shape of the knee moment increase after the start of stimulation was taken from the measurement results [4].

4 Verification of the model

The model was compared to the measurements of paraplegic person's standing-up [4]. The investigation [4] included measurements of ground reaction forces, seat supporting force, handle force and joint positions. A 3- dimensional model is presented in [4]. It was used to compute the net joint moments and reaction forces from the aquired data. This model was developed using the Newton - Euler formulation and was only appropriate to compute the inverse dynamics.

The verification of our model was divided into two parts. In the first part, the inverse dynamics was studied. The influences of the seat, the passive joint moments, the moment due to stimulation and the voluntary effort were not included in the model. The measured values were used instead. The body moved along the measured trajectories. Since all the external forces and moments acting on the body were known, the system of dynamic equations of motion was overdetermined. Thus, the moment and force in each joint were calculated in two ways. A bottom-up calculation was performed taking into account the measured ground reaction forces, while a top-down calculation included the measured arm forces. The more the two computed forces are similar, the better is the model. In this way, the accuracy of the anthropometric parameters used was evaluated (Figures 7 and 8). However, this type of the model evaluation is affected by other error sources. For example, the trajectories of the joint angles were computed by integrating the accelerations which were obtained by differentiating the measured joint angles.

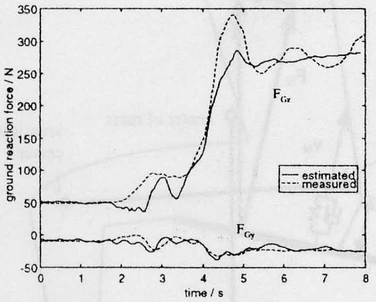

Figure 7. Comparison of measured and estimated ground reaction forces

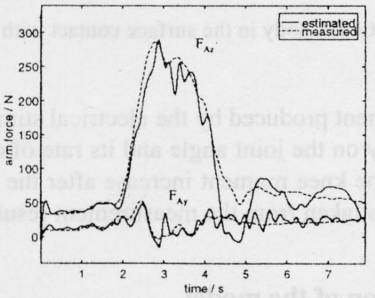

Figure 8. Comparison of measured and estimated arm forces

From the comparison of the inverse dynamics results of our model with the 3-dimensional model [4] it can be concluded that the stand-up-process approximation with a 2-dimensional model is acceptable.

In the second part of the verification, the complete model was tested. The reference trajectories of the PID controllers were taken from the experimental data [4]. The net moments and reaction forces acting at the joints were not measured. For this reason our model results were compared to the model [4] results. The results are shown in Figures 9 to 13. All the figures in this work refer to a single rising of a paraplegic patient MT (29 years old, 171 cm height, 75 kg of weight, T 4-5 lesion, injured 7 years ago, FES user for 5 years).

The aim of the model is to study different stand- up strategies by analysing the required forces and moments. The vertical arm force acting at the shoulder joint and the knee net moment are particularly important. The vertical arm force shows how much additional effort of the upper extremities a particular strategy requires.

The estimated knee moment has to be as close to the measured time course as possible to accurately represent the contribution of the electrical stimulation to the minimization of the effort.

The initial difference between the arm forces obtained

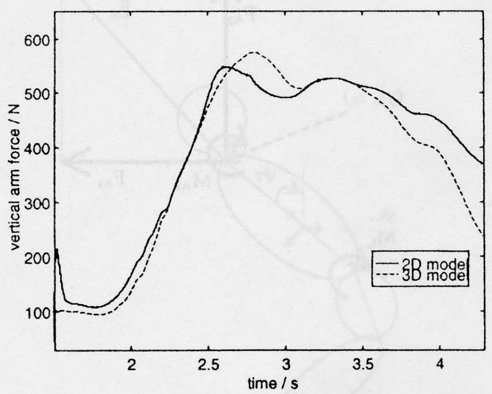

Figure 9. Estimated vertical arm force

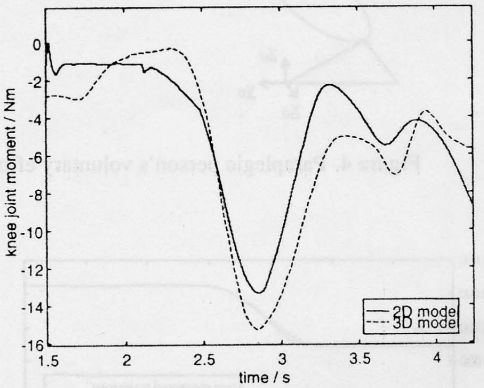

Figure 10. Estimated knee joint net moment

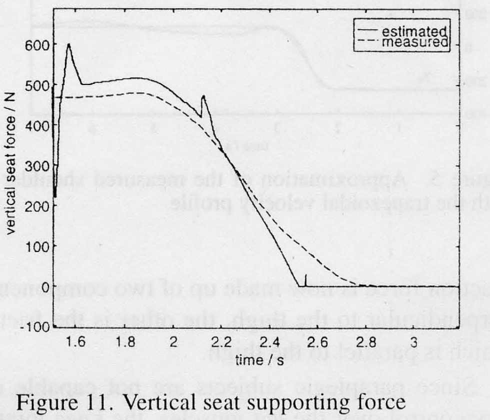

Figure 11. Vertical seat supporting force

from 2D and 3D model arises due to stabilization of the body on the seat (Figure 9). As the difference occurs when the patient is sitting still, it does not affect the model reliability. It can also be seen from Figure 9 that when the patient is already standing the estimated vertical arm force does not decrease appropriately. This occurs mainly due to improper values of passive joint moments. Passive joint moments were measured in a healthy subject [9]. The range of the motion of the lower extremity joints in a paraplegic subject is smaller. Thus the efficiency of

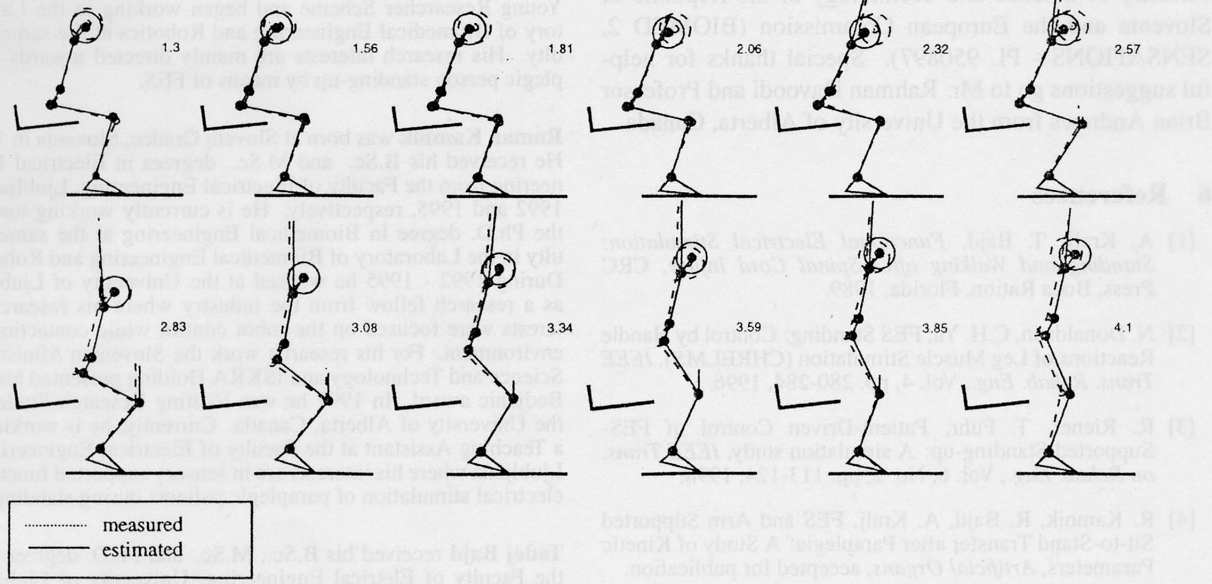

Figure 13. Measured and computed body movement during rising from a sitting to standing position. Arm force vector and knee joint moment vector are also shown

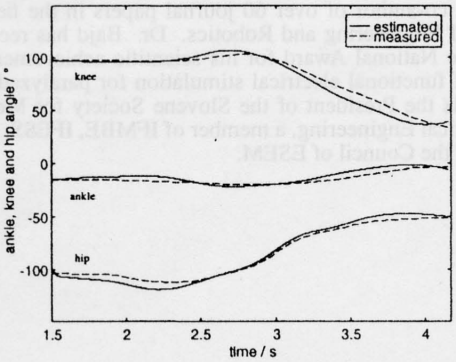

Figure 12. Joint angles of lower extremities

the leg support during standing is decreased. It was also noticed that the MT patient could not step with the whole foot surface on the force plate bacause of rather strong ankle joint contractures.

When the surface with the seat changes into a linear contact, a peak in the seat reaction force occurs. The model of the seat is likely to be simple. The value of the seat reaction force during quiet sitting is estimated too high. This can be due to the assumption that the seat force passes through the center of the mass.

5 Conclusion

A model of paraplegic person's standing-up was developed and implemented in the Matlab-Simulink simulation environment. The model was designed to study different standing-up strategies in completly paralyzed persons using FES of lower extremities by comparing the joint moments and reaction forces. The inputs to the model are desired trajectories of the joints that are controlled by a voluntary effort and the electrical stimulation. The most significant model output is the arm supporting force. The other outputs are ankle, knee, and hip angle trajectories and passive joint moments. The verification showed that the model behaves reliably.

The model is being used to assess whether a dynamic standing-up strategy can be adopted by a paraplegic subject. In the beginning of the standing-up process healthy subjects lean forward quickly in order to gain horizontal trunk momentum. In the next phase, the trunk and leg muscles act to transform the horizontal momentum to the whole body vertical momentum. When the body looses the contact with the seat, the projection of the body center of the mass is outside the supporting area of the feet. The body is only dynamically stable until the projection reaches the feet surface. On the other side, the body of a paraplegic patient is always statically stable during rising. Paraplegic subjects lean forward to transfer the centre of the mass over the feet supporting area. After this action they start to stand up. It is sensible to evaluate if it requires less effort for paraplegic subjects to stand up in a similar way as healthy subjects.

Another possible application of the model is to predict the efficency of rising while taking into account different initial foot positions and sitting postures.

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