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Human energy – optimal control of disturbance rejection during constrained standing

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An optimal control system that enables a subject to stand without hand support in the sagittal plane was designed. The subject was considered as a double inverted pendulum structure with a voluntarily controlled degree of freedom in the upper trunk and artificially controlled degree of freedom in the ankle joints. The control system design was based on a minimization of cost function that estimated the effort of the ankle joint muscles through observation of the ground reaction force position relative to the ankle joint axis. By maintaining the centre of pressure close to the ankle joint axis the objective of the upright stance is fulfilled with minimal ankle muscle energy cost. The performance of the developed controller was evaluated in a simulation-based study. The results were compared with the responses of an unimpaired subject to different disturbances in the sagittal plane. The proposed cost function was shown to produce a reasonable approximation of human natural behaviour.

Introduction

Arm-supported standing can be restored in patients with spinal cord injuries through functional electrical stimulation or mechanical bracing of paralysed lower extremities [1]. The standing posture is a prerequisite for accomplishing everyday activities. For example, a paraplegic patient would be able to reach some objects while standing that could not otherwise be reached from the confines of a wheelchair. These increased functional abilities may enhance personal self-esteem, while they provide a level of independence. Considerable effort has therefore been invested in finding an efficient method that offers a paraplegic person the possibility to stand without an arm support.

In order to achieve unsupported standing the body's centre of gravity has to be controlled over a relatively small support surface. To achieve this, body segments need to be positioned in a precise way relative to each other. In particular the ankle joint position needs to be precisely controlled, since only small deviations in the ankle joint angle result in large shifts of the total body centre of gravity (*CoG*). On the contrary knees and hips joints are mostly in a hyperextended position during standing, therefore their role is less significant. Since energy-efficient postures for stable standing are very limited, the selected balance control method needs to be efficient and robust.

Understanding an unimpaired subject's control over the constrained balancing provides useful information for selecting an efficient control method for unsupported standing of paraplegic persons. Analysis of unimpaired subject stance indicates a combined ankle – hip strategy as the most used sequence of responses to anterior/posterior disturbances [2]. Anterior or posterior leaning of the subject's body around the ankles prior to the disturbance also affects the overall postural strategy of disturbance rejection [3, 4]. When subjects are perturbed while standing near to their forward or backward limit of stability, they use a different postural strategy. An important feature of postural dynamics is the effect of the forward lean, which results in a significant increase of the tonic component of the ankle torque. Postural stability is here improved by simplifying the response to the perturbation [4], since the risk of falling backward is reduced by increasing the stability margin between the centre of gravity and the posterior limits of the base of support. Following that, the postural ankle dynamics can be based on a single muscle group—ankle plantarflexors.

In one of our recent studies we showed that unimpaired subjects responded to different anterior and posterior perturbations by increasing the ankle joint torque almost proportionally to the ankle joint angle, which resulted in a constant ankle joint stiffness [5]. In a latter study performed by Matjačić *et al.* the results showed that the same findings also apply for the lateral plane or any combination of the sagittal and lateral plane motion due to the disturbances [6]. However, as noted by Mihelj *et al.* [5], although the average stiffness is used to describe the postural control during perturbed stance, the choice of stiffness as a characteristic measure does not imply that the nervous system primarily regulates the joint stiffness.

A control strategy, which is an implementation of the constant stiffness control for unsupported paraplegic standing and utilizes the residual sensory and the motor abilities of a thoracic spinal cord injured subject, was proposed by Matjačić and Bajd [7, 8]. That strategy is based on voluntary and reflex activity of the patient's upper body and artificially controlled stiffness in the ankles. The knees and hips are maintained in the

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extended position by long leg braces. In this way, the subject is constrained in a double linked inverted pendulum structure. The artificial stiffness is generated using a hydraulic motor mounted in the rotation axis of the mechanical brace. When assisted by an artificial ankle joint stiffness value of 8 Nm/$^\circ$ or more, the paraplegic subject was capable of the proposed balancing. Both healthy and paraplegic subjects were also able to control balance, when standing was perturbed with different anterior/posterior disturbances, which indicates that the constant artificial ankle joint stiffness enables unsupported standing. With the upper trunk free to move, the subject retains voluntary control over the upright posture, which is an important step towards functional standing. The presented control methods allow a good insight into the control of unsupported standing. However, in our view they are not directly applicable in the restoration of unsupported standing of a paraplegic subject. When applying control of unsupported standing through the use of functional electrical stimulation, a major issue is the fatiguing of the stimulated muscles. Rapid muscle fatigue during standing can result in a loss of balance, which could lead to new traumas. During a stiffness-supported standing the subject controls muscle fatigue by using the upper trunk to balance the body in the most suitable posture; therefore the functionality of the upper trunk is reduced. In order to remove the burden of controlling fatiguing from the subject, the artificial control system needs to be designed in such a way that sustained functionality of the upper trunk is allowed and fatiguing is minimized by control rules. The greatest source of ankle muscle fatigue is a compensation of gravity-generated torque around the ankle joints. In order to minimize this torque, the vertical projection of the total body's centre of mass needs to be located within close proximity of the ankle joint axis. Another major contribution to muscle fatigue is the control of body sway in the anterior/posterior direction and the associated torque required to sustain the vertical body equilibrium.

In order to restore functional unsupported standing, a robust control system must be synthesized, so that it can provide support for upright balance, minimize muscle fatigue and at the same time allow the user to retain full control over the acquired posture. One of the prime goals of artificial posture control is the minimization of ankle muscle effort. This suggests the use of optimal control theory for controller synthesis. Implementation of this theory requires selection of optimization criteria which comprise the proposed control objectives. Through analytical and empirical analysis we assessed the centre of pressure (*CoP*) or ground reaction force position relative to the ankle joint as the biomechanical variable that denotes ankle muscle effort caused by gravity-generated torque and body acceleration when swaying. This was used as part of the cost function for the optimal control system design.

The present work involves upgrading the issues just highlighted with the main focus on understanding unimpaired balancing strategy during perturbed standing. We designed an optimal control system to sustain arm-free standing of a paraplegic subject and compared its performance to unimpaired subject responses. A statistically significant resemblance between the responses was observed, leading to an assumption about what may be the primary control goal of the central nervous system of the intact subject, when responses to the perturbations are controlled. The paper is organized as follows: stance dynamics are analysed and the control method is proposed in §2, comparison between artificial and natural controllers is presented in §3 and results are discussed in §4.

Methods

Stance dynamics

In order to design a robust artificial posture control system, body dynamics are considered as well as delays derived from information processing, command issuing and activation dynamics inherent to different muscle groups. Figure 1 depicts a schematic view of the unsupported standing dynamics, consisting of body (pendulum) dynamics, actuating muscles and inherent local control loops dynamics.

Each joint of the double inverted pendulum structure including trunk and ankle is activated by a set of muscles divided into agonist and antagonist groups. The muscle groups are functionally linked to the central nervous system (CNS). The processing delay inherent to the CNS was considered as a pure time delay and for further analysis approximated with the fourth order Pade functions. Signals from CNS define muscle activation. The muscle activation dynamics were approximated with the first order transfer functions. Further analysis is based on the assumption that the standing posture can be simplified as a double inverted pendulum structure with one degree of freedom (*DOF*) in ankle joint axis and the other in the lumbo-sacral joint (figure 2) [9].

Details on the system model can be found in the report by Mihelj and Munih [10]. We will use the preposition that the double inverted pendulum dynamics are linearized around the vertical equilibrium and the entire system is written in a state space form as:

$$\begin{aligned} \dot{\boldsymbol{x}}_\mu(t) &= A_\mu \boldsymbol{x}_\mu(t) + B_\mu \boldsymbol{u}_\mu(t) \\ \mu(t) &= C_\mu \boldsymbol{x}_\mu(t) + D_\mu \boldsymbol{u}_\mu(t)\,, \end{aligned} \tag{1}$$

where $\boldsymbol{x}_\mu(t)$ indicates the plant state vector consisting of pendulum dynamics, muscle dynamics and processing delays, $\boldsymbol{u}_\mu(t)$ the plant input vector, consisting of the ankle and trunk torque reference signals, and $\mu(t)$ the plant output vector, which includes body angles and angular velocities as well as the ankle torque [10].

The double inverted pendulum itself is an inherently unstable system. For a subject with a mass of 64 kg, a height of 1.74 m, an ankle/hip distance of 0.81 m, a hip/shoulder distance of 0.59 m, processing delays equal to 0.1 s and a muscle dynamics time constant of 0.1 s, the linearized system poles are shown in figure 3.

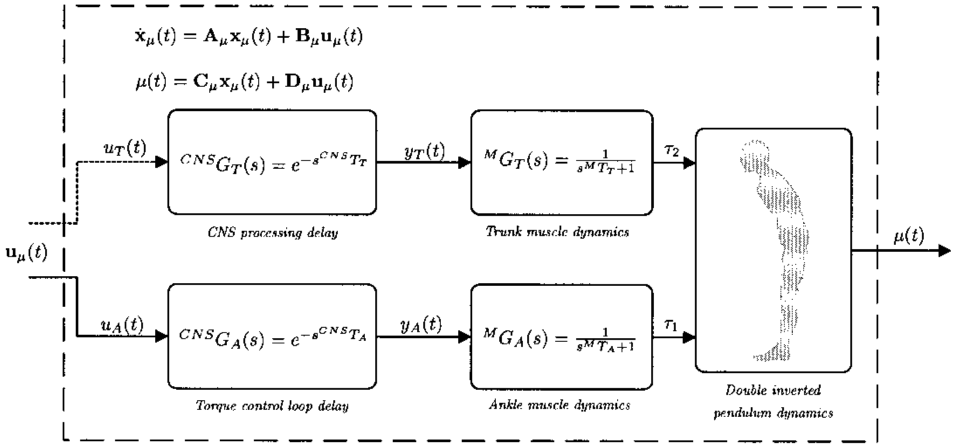

Figure 1. A schematic view of the unsupported standing dynamics, consisting of body dynamics, actuating muscles and inherent local control loops dynamics.

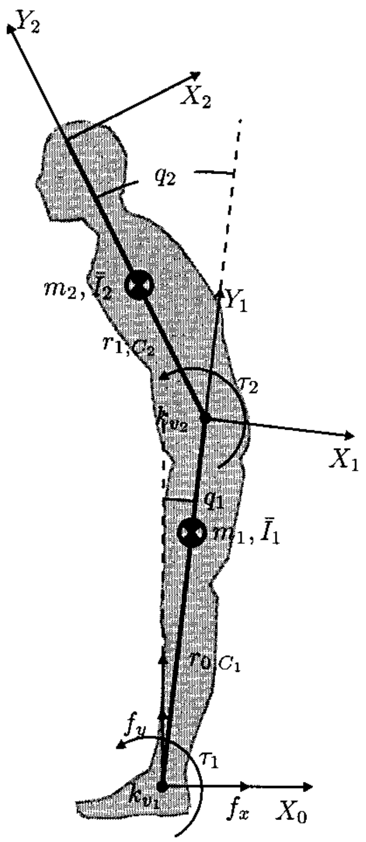

Figure 2. Double inverted pendulum structure: feet—support surface; lower extremities—first segment; trunk, head and arms—second segment.

The poles pertaining to the Pade functions and muscle activation dynamics are all stable and mostly far on the left half plane. However, two of the double inverted pendulum poles are located in the right half plane, thus resulting in an unstable system which needs to be stabilized. In intact subjects the stabilization is done by the central nervous system using an unknown control criteria. However, if restoration of the unsupported standing of a paraplegic subject is considered, then an artificial control system needs to be designed and implemented.

In order to design the artificial control system, we will first focus on kinematic and dynamic relations in the feet during standing (figure 4).

Since foot linear and angular acceleration equals zero during quiet standing, the position of *CoP* relative to the ankle joint axis can be determined only as a function of forces and torque acting about the ankle joint:

$$CoP(t) = -d - \frac{-\tau_1(t) - d f_y(t) + h f_x(t)}{f_y(t) - m_f\, g}\,. \qquad (2)$$

Further, considering the plant state space model (equation 1), the centre of pressure position $CoP(t)$ can be rewritten as a function of system states $\boldsymbol{x}_\mu(t)$ and inputs $\boldsymbol{u}_\mu(t)$ (see [10] for details):

$$CoP(t) = F\big(\boldsymbol{x}_\mu(t), \boldsymbol{u}_\mu(t)\big)\,. \qquad (3)$$

Control algorithm

Based on the redefined centre of pressure position (equation 3), a cost function for an optimal control system design was selected by:

$$\begin{aligned} J\big(\boldsymbol{x}_\mu(t), \boldsymbol{u}_\mu(t)\big) &= \int_0^\infty \Big(CoP^T(t)\, CoP(t) + \boldsymbol{u}_\mu^T(t) R \boldsymbol{u}_\mu(t)\Big)\, dt \\ &= \int_0^\infty \Big(\boldsymbol{x}_\mu^T(t)\, R_{xx}\, \boldsymbol{x}_\mu(t) + 2\boldsymbol{x}_\mu^T(t)\, R_{xu}\, \boldsymbol{u}_\mu(t) \\ &\quad + \boldsymbol{u}_\mu^T(t)\, R_{uu}\, \boldsymbol{u}_\mu(t)\Big)\, dt\,. \end{aligned} \qquad (4)$$

The cost function relates the cost value to the plant states through weight matrix R_{xx}, to the plant inputs through weight matrix R_{uu}, and to states and inputs through cross weighting matrix R_{xu}. All matrices are analytically determined from the subject's anthropometric data.

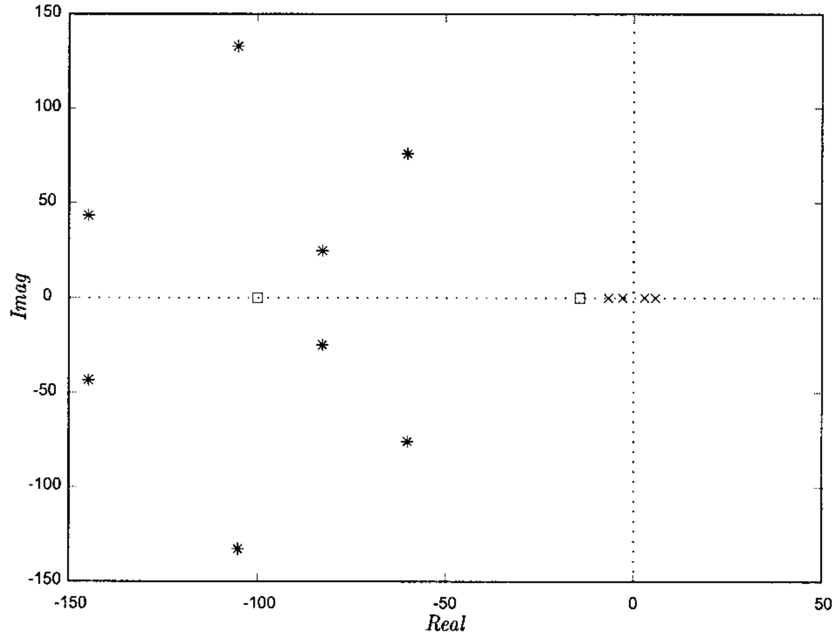

Figure 3. Open loop linearized system poles: double inverted pendulum (×), muscle dynamics (□), processing delay (*).

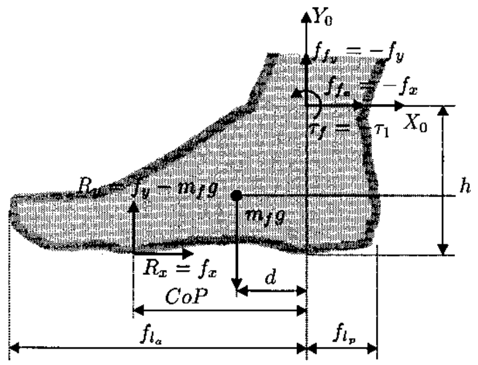

Figure 4. Kinematic and dynamic relations in foot during standing–estimation of ground reaction force distance form ankle joint axis.

The linear quadratic controller can be found as a unique, optimal full-state feedback control law:

$$u_\mu(t) = -{}^{LQ}K\boldsymbol{x}_\mu(t) \quad \textit{with} \quad {}^{LQ}K = R_{uu}^{-1}\left(R_{xu}^T + B_\mu^T S\right), \tag{5}$$

which minimizes the cost $J\,(\boldsymbol{x}_\mu(t),\, \boldsymbol{u}_\mu(t))$, subject to the dynamic constrains in equation 1. By defining $A_r = \left(A_\mu - B_\mu R_{uu}^{-1} R_{xu}^T\right)$, S is the unique, symmetric, positive semi-definite solution to the algebraic Riccati equation:

$$SA_r + A_r^T S + \left(R_{xx} - R_{xu} R_{uu}^{-1} R_{xu}^T\right) - SB_\mu R_{uu}^{-1} B_\mu^T S = 0\,. \tag{6}$$

The closed-loop dynamics derived by substitution of equation 5 into equation 1 are guaranteed to be asymptotically stable [11].

Since only the system states pertaining to the output vector can directly be measured, a Kalman full state observer was designed to estimate the full state vector $\hat{\boldsymbol{x}}_\mu(t)$ as:

$$\dot{\hat{\boldsymbol{x}}}_\mu(t) = F_K(t)\hat{\boldsymbol{x}}_\mu(t) + B_\mu \boldsymbol{u}_\mu(t) + G_K(t)\mu(t)\,, \tag{7}$$

where $F_K(t) = A_\mu - G_K(t)\,C_\mu$ [12]. Equation 7 has a form of time variable Luenberger observer, with Kalman gain matrix defined as $G_K = \Sigma C_\mu^T S_v^{-1}$. The S_v is the symmetric and positive definite sensor noise intensity matrix and Σ denotes the unique, symmetric, and at least positive semi-definite, $\Sigma = \Sigma^T \geqslant 0$, solution matrix of the filter algebraic Riccati equation

$$A_\mu \Sigma + \Sigma A_\mu^T + B_\mu S_w B_\mu^T - \Sigma C_\mu^T S_v^{-1} C_\mu \Sigma = 0\,, \tag{8}$$

with S_w symmetric and positive definite process noise intensity matrix.

The designed linear-quadratic-Gaussian (LQG) control system stabilizes the double inverted pendulum, providing a basis for unsupported standing in paraplegia. Poles of the combined plant-controller system are located in the left half-plane as shown in figure 5; there are more details in figure 6.

According to the separation principle, the linear quadratic regulator and the Kalman filter can both be designed and tested separately to validate the performance. Therefore, the poles of the Kalman estimator are independent of those of the controller. It can be observed from figures 3 and 6 that the designed control system shifts only the poles of the double inverted pendulum structure, while the poles pertaining to the activation dynamics and signal processing remain unchanged.

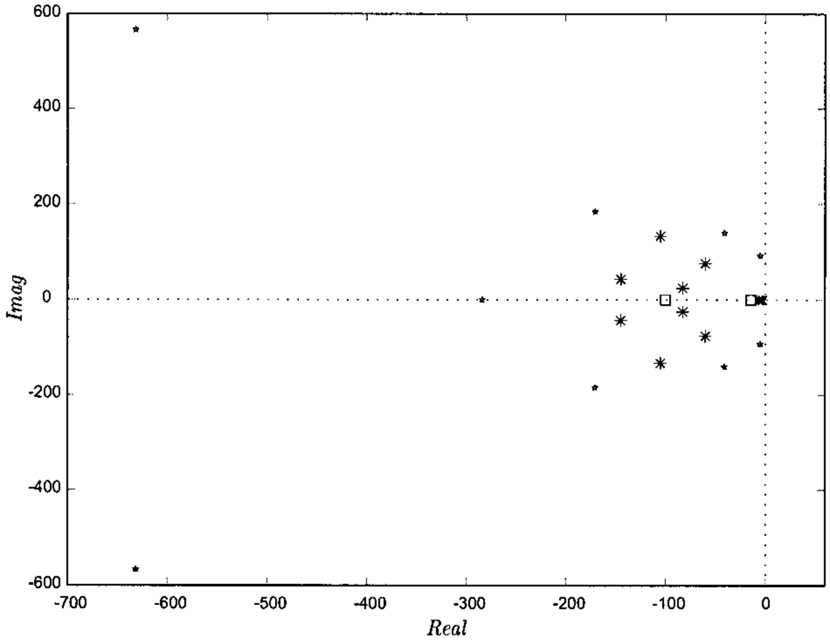

Figure 5. System and controller poles: Kalman estimator (☆), double inverted pendulum (×), muscle dynamics (□), processing delay (*).

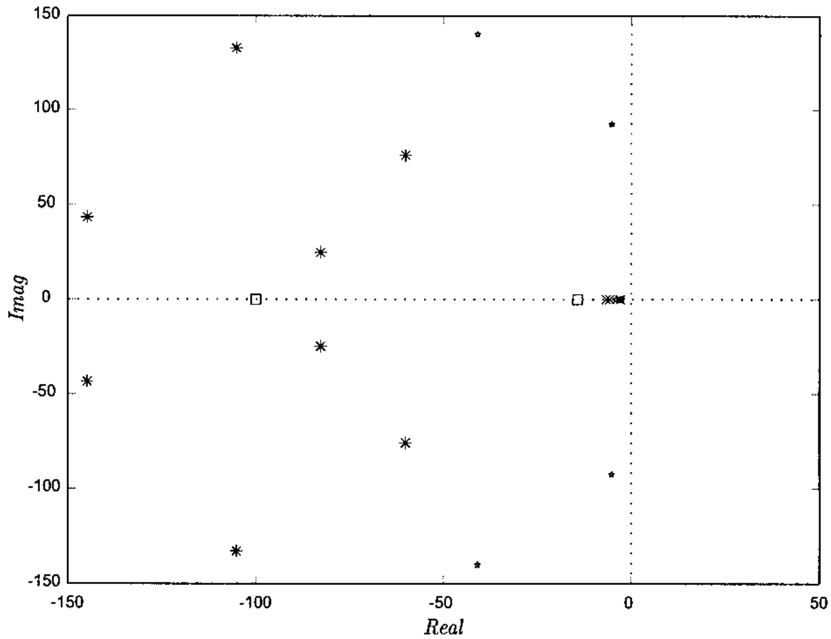

Figure 6. System and controller poles in details: Kalman estimator (☆), double inverted pendulum (×), muscle dynamics (□), processing delay (*).

The system was found to stabilize the unsupported standing of a complete spinal cord injured subject [10, 13]. However, in this study we will use the designed optimal control system to investigate the balancing strategy of an unimpaired subject. Therefore, responses of the artificial controller and the subject to identical input signals introduced through the ankle torque perturbations will be evaluated.

Experimental set-up

Figure 7 shows the experimental set-up with an intact subject placed in the mechanical rotating frame (MRF). The device constrained the body by allowing movements only in upper trunk and ankle joints, while the arms were folded on the chest.

The device consisted of a base fixation, a rotating frame and a hydraulic actuating system. The rotating part of

Figure 7. Experimental set–up.

the device consisted of a frame providing bracing to the lower body and forcing the knees and hips into an extended position with aluminium bars. The hydraulic actuator mounted in the ankle rotation axis provided the torque required to produce various disturbances. More details on the MRF are given in [5, 8]. Two force–plates (AMTI, Advanced Mechanical Technology) were mounted in the base of the device, in order to allow a measurement of reaction forces and torques separately for each foot. The movement kinematics was assessed by the optical position measuring system OPTOTRAK (Northern Digital) Two infrared markers were attached to the rotating frame, one on the bearing axis and the other on the vertical bar of the bracing system at the height of the subject's lumbosucral joint axis. Two additional markers were attached to the subject's trunk with the first located on the midline of the rib cage half way between the iliac crest and the shoulder. The second marker was then located five centimetres below the first. The joint angles were defined as in figure 2. Positive ankle angles correspond to the ankle dorsiflexion and the positive trunk angles to trunk flexion.

Results

The initial evaluation of the posture control system was performed in a simulation based study. This was followed by an investigation on neurologically intact subjects.

Simulation results

In a previous study we analysed responses of neurologically intact subjects to different perturbations in anterior and posterior directions [5]. The perturbation magnitudes were selected in a manner that elicited either the ankle or the combined ankle–hip strategies as a response to perturbation. In the present study we used the results of those measurements to validate the control system performance in a simulation based study, which allowed the comparison of the unimpaired subject balancing strategy to the designed controller operation. The nonlinear plant model was used in combination with the designed controller. The model initial standing posture was perturbed using the perturbation magnitude and duration from the experiments with the intact subject. The average subject responses based on ten measurements and a model output are presented in figure 8. Model and human angle and torque time courses generally followed the same trajectory with only small deviations.

The initial oscillations in the trunk angle and torque were the artifact of the perturbation. The model output was generally found to be smoother than the corresponding human response. The human response was slightly more oscillatory compared to the model output due to the complex CNS signals, in comparison with the simplified system model comprising only two segments. The coefficient of correlation between the model output and the subject response was 0.97 for both angles and indicated that there was a statistically significant similarity between the responses. The analysis of torque responses showed the same results, with the coefficient of correlation even higher for ankle torques (0.98) and slightly lower for trunk torques (0.93). The latter was probably the result of relatively inaccurate computation of the actual subject trunk torque, due to errors in the anthropometric and kinematic data.

Intact subject perturbed standing

In the next part of the study we compared the selected control methodology to the intact subject's perturbed stance. The person was constrained as shown in figure 7 with standing perturbed by random disturbances in the anterior/posterior direction. The subject was instructed to respond with voluntary and reflex activity based on the perception of the disturbances. The control system had no immediate information about the perturbation, and the output was generated in real time based on information about current subject posture. This output was used as a reference signal for muscle activation. However, since in this case the reference was not fed to the muscle, the signal was not influenced by the muscle

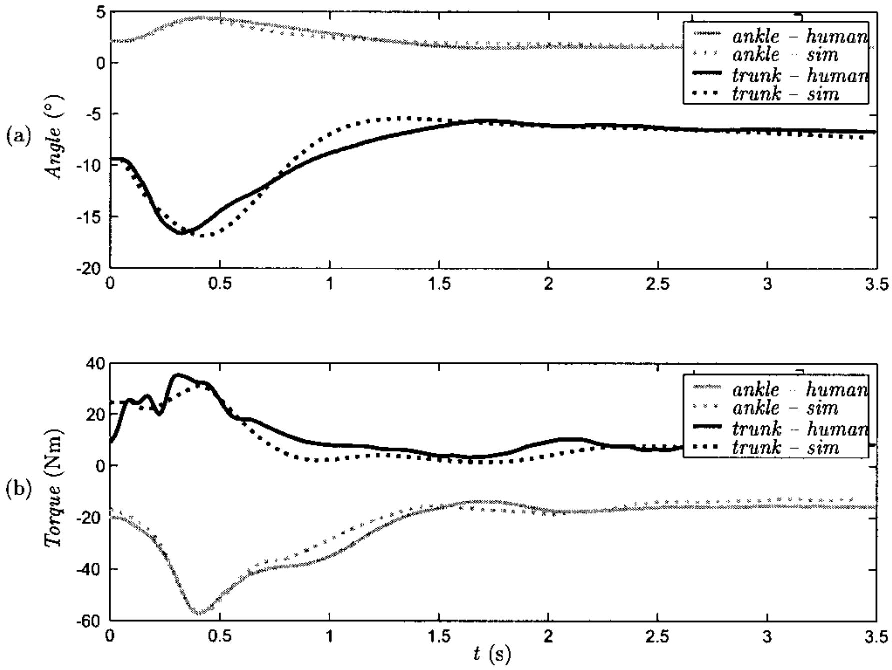

Figure 8. Simulation results and unimpaired subject responses to the same perturbation applied at time 0 s: simulated and measured ankle and trunk angles (a), simulated and measured ankle and trunk torques (b).

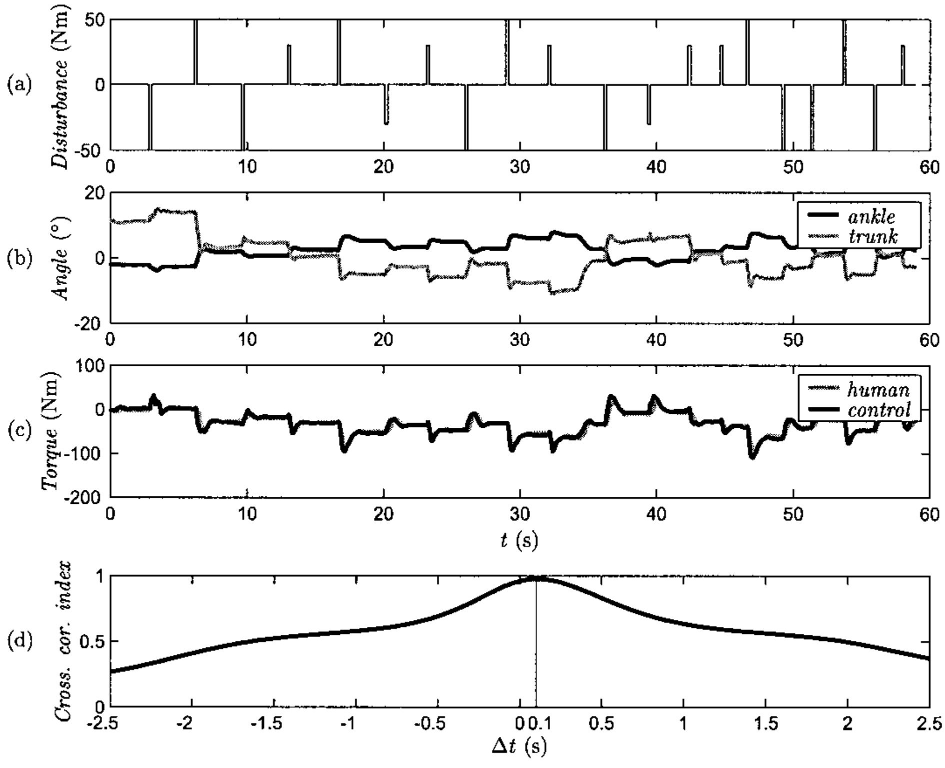

Figure 9. Comparison between human and LQG: sequence of perturbations (a), ankle and trunk angles (b), measured and predicted ankle torque (c), and cross correlation index for measured and predicted ankle torque (d).

activation dynamics. Therefore the signal was actually a prediction of the measured unimpaired muscle torque. A detailed comparison of the control system output and ankle torque generated by the subject is presented in figure 9. Plot (a) shows the distribution of disturbances over time. Two different magnitudes of disturbance were applied either in the anterior or posterior direction in a random order. Each time the disturbance was applied, the posture was perturbed and the subject responded with a voluntary and reflex

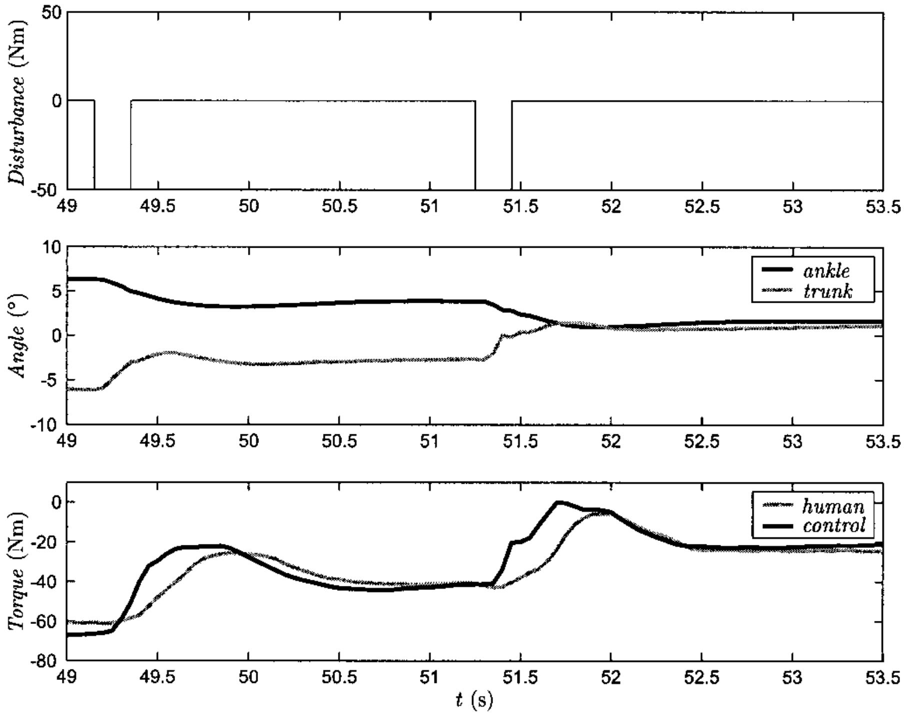

Figure 10. Comparison between human and LQG—detail: sequence of perturbations (a), ankle and trunk angles (b), and measured and predicted ankle torque (c).

activity in order to maintain equilibrium. Ankle and trunk angles are displayed in plot (b). The ankle angle mostly spanned the entire feasible range that still enabled the subject to maintain a vertical equilibrium. The unimpaired subject ankle torque and controller output are shown in plot (c).

A noticeable similarity can be observed in the torque amplitude either in response to the disturbance or in quiet standing between the disturbances. However, as already noted we expected small differences in the timing of the responses. This can be easily observed by visual inspection of the time courses (figure 10). Additional confirmation was provided by a cross-correlation analysis of the signals. The analysis indicated that the artificial control system response precedes the subject generated torque by at least 100 ms. At its maximum the correlation index amounted to 0.98, and proved the equality in the control strategies. The difference in timing was small but extremely important, since the muscle activation dynamics amounted to approximately the same value. There was a good agreement with the results of EMG-based analysis of timing of the responses to perturbations, where EMG latencies of ankle muscles were determined to be 105 ± 20 ms [5]. Therefore, the response time of the artificial control system, which includes the posture control loop and the muscle activation dynamics, would be equal to the response time of a neurologically intact subject. An enlarged detail of the time courses in figure 9, indicating the responses to two perturbations, is presented in figure 10. The similarity in the amplitudes of responses and the small difference in the timing can be verified.

Discussion

The assumption and maintenance of upright posture is such a common occurrence among human beings that it is perhaps the most universally accepted measure of normality [14]. An important reason for the renewed interest in standing is an appreciation of the central importance of the actions of the neurologically intact neuromuscular system of the upper body. The realization that the number of degrees of freedom of the paraplegic body are such that their posture may still be controllable by the intact neuromuscular system suggests that artificial controllers for standing should be designed in a way to fully exploit the subject's preserved natural resources [15]. Until now, research in this field has been limited to synthesis of an artificial control system usually without considering voluntary activity.

An important step toward understanding the control mechanism of unsupported standing in paraplegia was to consider body dynamics in a double inverted pendulum structure. Such interpretations result in an under – actuated mechanism with an active joint in the upper body and a passive one in the paralysed ankle joints. The control proposed in this paper is novel in the sense that it integrates natural and artificial control. These two concurrent controllers are acting in parallel. The physiological system which acts under voluntary control is being followed in its actions by the artificial support system. The aim of the feedback control is to recover and maintain a certain body posture, which can be specified by the user.

The selection of control method was based on analytical and empirical results, which indicated the ground reaction force position as an ideal criterion to be included in the optimal control cost function. The control criterion assures on one hand the minimization of muscle effort, with the purpose of prolonging the time of unsupported standing and, on the other hand, it also guarantees the stability of the posture by maintaining the centre of pressure position beneath the subject's feet.

The aim of this work was to compare the artificial control of unsupported standing with that of an unimpaired subject's response to perturbations in the sagittal plane. Even though we used a fairly simple model of standing for the design of the artificial control system, we observed similarity of responses that were statistically significant. Many of the differences between the experimental and simulation trajectories can be attributed to simplifications in the modelling. Errors in prediction mostly arise from lack of rigidity within human body segments and approximations in anthropometric data. Additionally, the LQG controller produces an optimal return trajectory, being a function of the states. To study the responses to perturbations it is necessary to provide reasonable estimates of the initial and final conditions of the states, which are used to compute the return trajectory. Since 14 parameters determine the system state, small deviations in the initial and final angle as well as torque values are the result of an estimation process.

Both regulator and estimator portions of the LQG controller are implemented using the steady-state feedback gain, which are appropriate in the middle portion of a long trajectory. Linear quadratic regulators are expected to use time-varying gains near the end of the trajectory, when terminal objectives, which are not part of the objective in the current model, may have precedence. The use of steady-state gains greatly simplifies computation of the controller. However accuracy could be improved by implementing terminal objective and time-varying gains.

The cost function proposed in equation 4 has been shown to produce reasonable approximation of human natural behaviour. The controller introduced here regulates the centre of pressure position within the support surface. By maintaining the centre of pressure close to the ankle joint axis, the objective of the upright stance is fulfilled with the minimal ankle muscle energy cost. Nevertheless, the control objective does not exclude use of an ankle or a hip strategy. Selection of the response strategy is dependent on the perturbation magnitude and is left to the controller. The resulting controller behaves functionally like the CNS, choosing the ankle strategy for smaller disturbances and switching to the hip strategy for larger disturbances. Nevertheless the response of the unimpaired subject to perturbation is based on some combination of feedforward and feedback control signals; the artificial controller produces the same trajectories with only the feedback component.

It is quite possible that such a cost function results in a controller approximating very well to natural behaviour. The position of centre of pressure seems a reasonable biomechanical variable on which the unimpaired subjects can rely for their control of unsupported standing. From the control point of view, the position of *CoP* is easily controlled through activation of the ankle plantarflexor and dorsiflexor muscle groups. Furthermore its position can be fairly easily measured with pressure sensors under the foot. Nevertheless one should not attach too much importance to an interpretation of cost function, because there are many other possible interpretations that could produce the same result.

More careful modelling of the cost function or the nonlinear cost function can produce predictions with greater fidelity to experimental results, but the identification of significantly complicated objective functions are ultimately limited to feasibility of the experiments and inter-subject variability.

Considering the task requirements, the control strategy for standing might change in different situations (for example when feet are not aligned), but when quiet standing is required and a rejection of disturbances is necessary, minimization of the *CoP* distance from the ankle joints axis seems a reasonable option. Not only is it important in understanding unimpaired subject standing strategy, but also, notably, the design of an artificial controller for the unsupported standing of a paraplegic patient is simplified, since the centre of pressure position becomes the variable to measure and control.

Conclusions

The paper analyses a novel control strategy for a closed-loop control for the restoration of unsupported standing in spinal cord injured subjects from a perspective of an unimpaired subject. The presented algorithm integrates the preserved upper body motor and sensor functions with the artificial control of the paralysed ankle joints and ensures stable standing of the paraplegic subject. However, the control algorithm does not only enable unsupported standing of a paralysed subject, but also gives a good insight into the intact subject control strategy for standing. The control system implements the ankle strategy when the body sway is small, and generates the combined ankle hip and hip strategy when rejection of large perturbations is required. The successful performance of the designed controller does not imply that the central nervous system functions as a linear-quadratic-Gaussian controller. The only presumption is that for the specific operating conditions for which the artificial controller was designed, the LQG control system achieves the same functionality as the CNS. The system achieves its goal by stabilizing the body against different perturbations.

Acknowledgments

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