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Modelling Seated Postural Stability for Complete Spine Cord Injury

* M. Blandeau, * T. M. Guerra, * P. Pudlo, * F. Gabrielli, ** V. Estrada-Manzo

* LAMIH UMR CNRS 8201 University of Valenciennes and Hainaut-Cambrésis
Le Mont Houy, 59300, Valenciennes, France (mathias.blandeau@univ-valenciennes.fr)
** Department of Electrical and Electronics Engineering, Sonora Institute of Technology
5 de febrero 818 Sur, CP 85000, Obregón, Mexico (victor.estrada@itson.edu.mx)

Abstract

This paper introduces a modelling approach to investigate seated postural stability for subject suffering a complete spine cord injury (SCI). A mechanical model, variation of the double inverted pendulum, is used to represent a subject in the sagittal plane moving back and forth his uppers limbs and head to maintain his stability. A non-linear descriptor system is obtained by developing the Lagrangian equations and then written as a Takagi-Sugeno model. An Unknown Input Observer (UIO) is then used to estimate the system state and the unmeasured internal stabilizing force. Simulations results attest that this technique can be used to non-invasivaly re-create unmeasured data and contribute to a better understanding of how people with complete spine cord injury maintain their stability.

Keywords

Postural Control, Sitting Stability, Spine Cord Injury, Double Inverted Pendulum, Takagi-Sugeno models, Unknown Input Observer

1. Introduction

Every year, between 250 000 and 500 000 people suffer from spine cord injury (SCI) [1]. More than 30% of them will suffer a complete injury, which is the loss of all mobility and sensitivity under the injury level, and will need a wheelchair. Injury risks linked to seated position is very high for SCI people. It has been observed that 90% of wheelchair related falling injuries occur during normal use of the wheelchair [2]–[4], this figure shows that this tool still needs some improvements. Wheelchair users are often exposed to perturbation during activities of daily living (transportation, transfers…) but depending on the level of their injury, they can lose the ability to contract their abdominal and dorsal muscles which are the main stabilizing muscles of the inherently unstable spine [5]. Instead, they can use their upper limbs and head to keep their balance (Fig. 1), a classical re-adaptation exercise and objective for SCI patients [6].

The vast majority of studies on human postural control concerns the standing position with a focus on lower limbs joints which are less impacted by the different structures of the body (muscles, bones, soft tissues, organs…) than the vertebral joints where the major activity of the sitting control occur [7].

Several teams tried to deal with this issue by using a simple mechanical model: an inverted pendulum with an active torque on the abdominal region [8]–[10]. Not only is this representation erroneous when considering SCI people above a certain injury level, it has also been shown that people moved their head when exposed to perturbation while sitting [11], [12]. A new mechanical model appears to be needed. It could be used to study the impact of various element of postural control and to analyze the impact of hypothesis, for instance, showing by simulation if a specific perturbation would lead to a fall or not. Moreover, such a model could be used to continue experimentations beyond the methodological or ethical limitations. In order to better understand the sitting control of people with SCI we need to create a mathematical model of postural sitting control considering the upper segments of the body: the head and upper limbs.

We know that biological system are intrinsically delayed, Peterka showed that a delayed closed-loop control law could generate human-like posturographic results when standing [13]. The activation delay can be defined by the sum of three sub-delays, the neural transmission time, the central nervous system processing time and the eletromechanical delay (time needed for an excited muscle to generate force) [14], [15].

We hence propose the following H2AT (head two arms and trunk) model as a variation of the double inverted pendulum in order to take into account the upper limbs and head displacement impact, this displacement being generated by a time-varying delayed force. There is no non-invasive and ecological experimental way to measure directly the value of this kind of internal force hence we have to rely on model calculation methods. This model goal is the estimation of non-measured variables of the model (e.g.: segment speed or acceleration, internal control forces and torques…) for people with SCI to reproduce their motor behavior in a simulation environment.

In biomechanics, inverse dynamic is the traditional method to calculate joint forces and torques by using the acceleration of body segments. This technique presents several limitations with among other, the risk of information loss because of the double derivation of experimental segments center of mass positions but also, because of its iterative nature, the accumulation of error at each new joint [16]. In this article we choose a control theory approach which exactly represents the dynamic of the system and estimates the internal variables. We then need to rewrite the mechanical equation into the fuzzy formalism of Takagi-Sugeno and use an Unknown Input Observer (UIO) to estimate both the state values and unmeasured variables. This method has been used once for the study of standing postural control [16], but the obtained equations led to a Bilinear Matrix Inequality (BMI) condition which are difficult to solve mathematically. The works of [17] and [18] have succeeded in resolving this problem with Linear Matrix Inequalities (LMI) for which robust solution-finding algorithms exists.

The main goal of this article is to present this new modelling method for a biomechanics problem and test by simulation its abilities in estimating unmeasured values. This might lead us to a better understanding of the contribution of upper segments in the balancing process of SCI subjects.

2. Material and Methods

2.1 The H2AT model

The H2AT model (Fig. 1) is a variation of the classical double inverted pendulum in the sagittal plane. It is composed of 2 rods articulated one to the other. The first rod representing the trunk is linked to the support (the seat) by a revolute joint standing for the lumbosacral joint (represented by angle θ). The second rod is linked to the first one by a prismatic joint and its displacement is represented by the variable x. This value represents the ability to move back and forth the upper segments center of mass (head and upper limbs combined) in the sagittal plane.

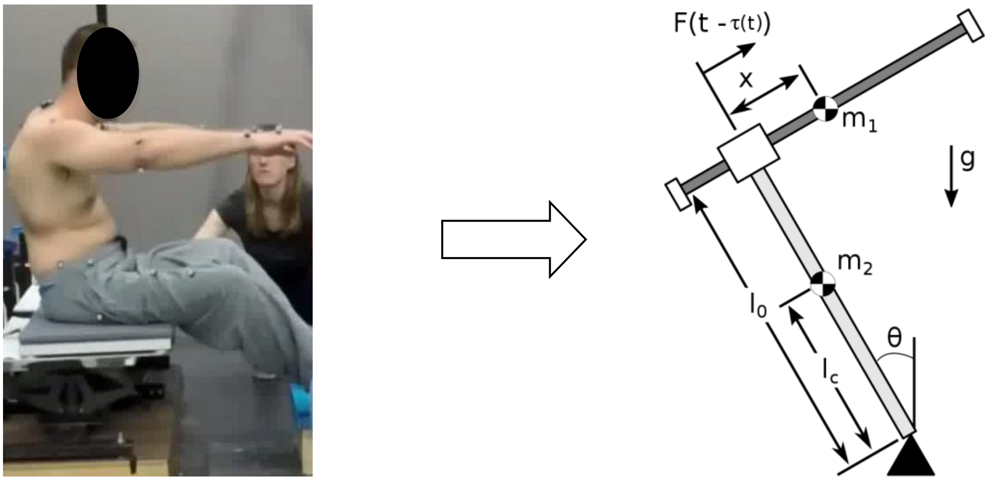

Fig. 1. A SCI subject keeping sitting balance with the upper segments and the H2AT model

In this case, the only dynamic parameters of the model are the gravity and the force $F\left(t-\tau(t)\right)$ which allow the displacement of the upper rod. Because we study subjects with a complete SCI above the abdominal level, there is no muscle torque around the revolute joint.

In order to obtain the dynamic equations of the system, we calculate the Lagrangian, $L=K-U$ with K and U, respectively the kinematic and potential energy of the system. We consider $K=K_1+K_2$ with $K_1=\frac{m_1}{2}\left(l_0^2\dot{\theta}^2+\dot{x}^2+x^2\dot{\theta}^2-2l_0\dot{x}\dot{\theta}\right)$ and $K_2=\frac{1}{2}m_1 l_C\dot{\theta}^2$ and $U=U_1+U_2$ with $U_1=m_1 g\left(l_0\cos(\theta)+x\sin\sin(\theta)\right)$ and $U_2=m_2 g l_c\cos(\theta)$. Then by calculating $\frac{d}{dt}\frac{\partial L}{\partial\dot{x}}-\frac{\partial L}{\partial x}=F\left(t-\tau(t)\right)$ and $\frac{d}{dt}\frac{\partial L}{\partial\dot{\theta}}-\frac{\partial L}{\partial\theta}=0,$ we end up with the following system:

$$\begin{cases}0=m_1\ddot{x}-m_1 l_0\ddot{\theta}-m_1 x\dot{\theta}^2+m_1 g\sin(\theta)-F\left(t-\tau(t)\right)\\ 0=-m_1 l_0\ddot{x}+J(x)\ddot{\theta}+2m_1 x\dot{x}\dot{\theta}-\left(m_1 l_0+m_2 l_c\right)g\sin(\theta)+m_1 g x\cos(\theta),\end{cases}\tag{1}$$

With $x=x(t),\ \theta=\theta(t),\ \dot{x}=\dot{x}(t),\ \dot{\theta}=\dot{\theta}(t),\ \ddot{x}=\ddot{x}(t),\ \ddot{\theta}=\ddot{\theta}(t)$ and $J(x)=m_1\left(l_0^2+x^2\right)+m_2 l_c^2$. We can rewrite (1) by considering the state vector $X=\left[x(t)\quad\dot{x}(t)\quad\theta(t)\quad\dot{\theta}(t)\right]^T$, in the following non-linear descriptor form system:

$$\begin{cases}E(X)\dot{X}=A(X)X(t)+BF\left(t-\tau(t)\right)\\ y(t)=CX,\end{cases}\tag{2}$$

where

$$A(X)=\begin{bmatrix} 0 & 1 & 0 & 0 \\ 0 & 0 & -m_1 g\dfrac{\sin(\theta)}{\theta} & m_1 x\dot{\theta} \\ 0 & 0 & 0 & 1 \\ -m_1 g\cos(\theta) & -2m_1 x\dot{\theta} & \begin{pmatrix} m_1 l_0 + \\ m_2 l_c \end{pmatrix} g\dfrac{\sin(\theta)}{\theta} & 0 \end{bmatrix}, B=\begin{bmatrix} 0 \\ 1 \\ 0 \\ 0 \end{bmatrix}, \ C=\begin{bmatrix} 1 & 0 \\ 0 & 0 \\ 0 & 1 \\ 0 & 0 \end{bmatrix}^T,$$

$$E(X)=\begin{bmatrix} 1 & 0 & 0 & 0 \\ 0 & m_1 & 0 & -m_1 l_0 \\ 0 & 0 & 1 & 0 \\ 0 & -m_1 l_0 & 0 & J(x) \end{bmatrix}.$$

This model has been created in a parametric way so that it needs very few biomechanichal values to be used. For the simulation purposed, we considered an average 80Kg man. Because the masse of upper segments does not significantly change between healthy and SCI subject [19], we can use classical regression rules to obtain the length and mass of each segments [20]. This way we obtain $m_1 = 16.1\ \mathrm{Kg}$ for the mass of upper limbs and head combined; $m_2 = 26.64\ \mathrm{Kg}$ for the trunk mass; $l_0 = 477\ \mathrm{mm}$ for the trunk length and $l_c = 276.66\ \mathrm{mm}$, for the distance between the trunk center of mass and the revolute joint. A complete flexion and extension of both upper limbs and neck results respectively in $x = 105.27\ \mathrm{mm}$ and $x = -75.18\ \mathrm{mm}$ [21].

2.2 Fuzzy Takagi-Sugeno model

One way to deal with this kind of non-linear system in (2) is to use the Takagi-Sugeno (TS) formalism [22]. This method consists in writing the non-linear system as the sum of several linear sub-systems which are constant in time and then multiply them by time-varying weighting functions called membership functions. Because the number of sub-systems is exponentially linked to the number of non-linearity, we choose to leave the descriptor matrix $E(x)$ on the left side of the equation, hence keeping this number (and with it, the complexity of the TS model) low [22]. This formalism gives us an exact representation of the system (2) with the following form:

$$\begin{cases} \sum_{k=1}^{e} v_k(z) E_k \dot{X} = \sum_{i=1}^{r} h_i(z)\big(A_i X + B_i F(t-\tau(t))\big), \\ y(t) = \sum_{i=1}^{r} h_i(z) C_i X, \end{cases} \tag{3}$$

where e and r are the numbers of sub-systems for the left and right side of the equation and $h_i(z)$ and $v_k(z)$ care the membership functions. Matrixes E_k, A_i, B_i and C_i are constant in time.

The stability analysis and the control law design of this TS model are based on Lyapunov theory and the definition of LMI conditions [23].

2.3 Unknown Input Observer (UIO)

In control theory, an UIO is a mathematical tool that provides an estimate of a system internal state, and also of unmeasured values. In our case, we measure neither the states $\dot{x}(t)$ and $\dot{\theta}(t)$ nor the internal force F but with the UIO we can estimate them. From measurements of the output, it reconstructs the state values by minimizing the error between the real output and the estimated one (Fig. 2). The works from [18] and [17] can be used to design an UIO for TS-descriptor models, by convention, estimated data are written with a « ^ ».

$$\bar{E}\dot{\hat{X}} = \bar{A}\hat{X} + L\left(y - \hat{y}\right), \qquad \hat{y} = \bar{C}\hat{X}, \tag{4}$$

2.4 Control Law with time-varying delay

As someone with complete SCI in seated position, the H2AT model is inherently unstable and hence must be stabilized. Classical inverted pendulum have already been balanced by various control law [13] but we are interested by robust time varying control law for we can see in (2) that force F is delayed by a time-varying parameter $\tau\left(t\right)$. This delay represents the sum of times for information processing, neural transport and the electromechanical delay [24]. A control law was proposed by [25] with the following equation:

$$u\left(t\right) = K\left[X + \int_{t-\tau_0}^{t} e^{A(t-s-\tau_0)} Bu\left(s\right) ds\right]. \tag{5}$$

Specific technical details about the Takagi-Sugeno formalism, the implementation of the control law, observer gain calculation and system stabilization with LMI conditions are off the topic of this paper and are described in [26], [27].

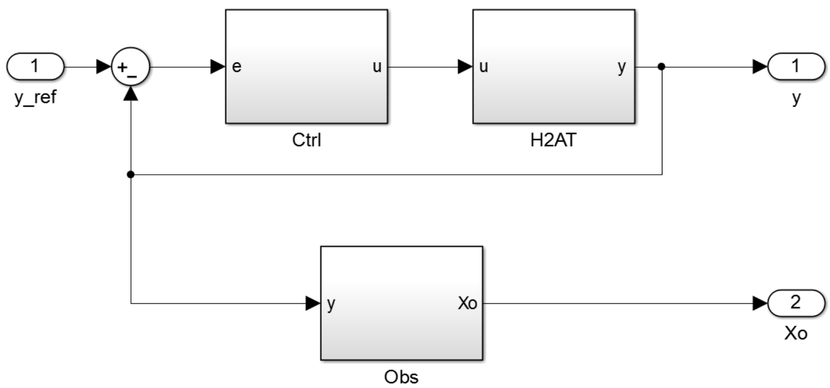

Fig. 2. Schéma d'un OIE classique

2.5 Simulation

In order to validate numerically both the behavior of the model and the TS observer, we simulated during 2 seconds the stabilization of system (2) with the $60 \pm 10\text{ms}$ delayed control law (5) through a numerical simulation software (Simulink v8.2, MathWorks). Initial conditions of the system and observer were respectively $X(0) = [0 \quad 0 \quad 0.005 \quad 0]^T$ and $\hat{X}(0) = 0_{4\times1}$. System output $y(t)$ is injected in the UIO which then estimates the state and force F. At the end of the simulation, the observer estimations are compared to the real simulated values and both the precision and convergence speed are analyzed.

3. Results

The UIO estimation of system (2) simulated with the control law (5) are presented on Fig. 3 and Fig. 4. Figure 3 shows the evolution in time of the force F (in black) and its estimate (in dashed-grey). Three points are to be highlighted; first we can see at the beginning of the simulation that F is null because of the delay; second the general behavior of the control law is decreasing which indicates stabilization over time. Finally the observer estimation is very close to the real command after 1 second of simulation.

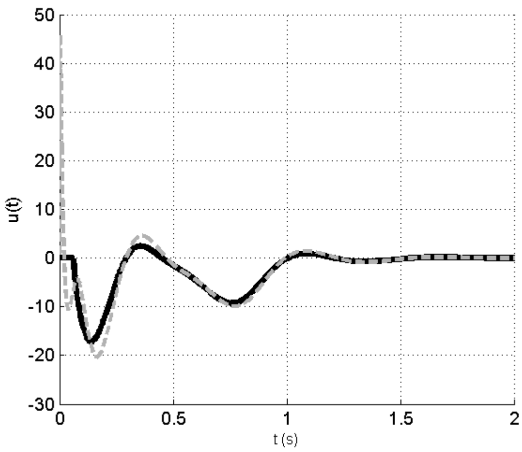

Fig. 3: Simulated control law (black) and estimated (dashed-grey)

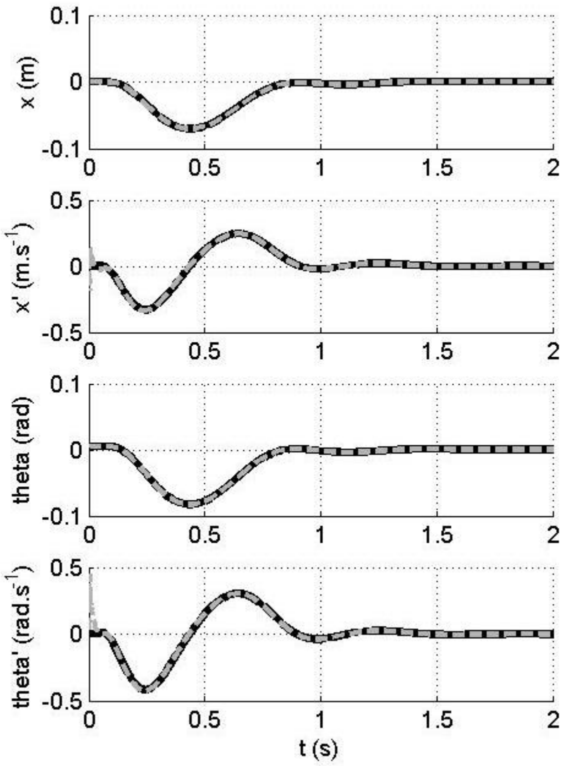

Fig. 4: Simulated state values (black) and estimated (dashed-grey)

Fig. 4 depicts the variation of the 4 components of the state in black and their estimates in dashed-grey. Let us remind that $X_1 = x$ and $X_3 = \theta$ are measured as output of the system. Not only can we see the stabilization on the system as all the states go to zero but the observer estimations on X_2 and X_4 converge very quickly even though the initial conditions are not the same.

4. Discussion

Our objective is to develop technological device to facilitate balance control in seated position for people with SCI. Before doing that, we need a better understanding of how these people manage to keep their stability without any voluntary muscle activation below their injury level. This is why we created a variation of a double inverted pendulum to model the stabilization of the trunk by the upper limbs and head displacement. A Takagi-Sugeno fuzzy unknown input observer was designed to estimate the force generated in the model at the superior joint of the model and simulated with a varying-time delayed control law.

Simulation results showed the UIO is fitted for the estimation of internal and unmeasured data from the output information of the model which can be measured with motion analysis techniques.

This is interesting in the fact that we do not have to derivate twice the experimental data to obtain the acceleration of the segment like it is done in the inverse dynamic.

They are some limitations to this technique; first of all, we define the lumbosacral joint as a perfect revolute joint, without any passive resistance to flexion or extension. Although it is known that flexion of the trunk can be followed by a resistive force created by the passives structure of the human body (soft tissues, tendons, intra-abdominal pressure…) [7], [8], this assumption is often done in biomechanics for the study of the ankle [13] or the trunk [10]. Then there is the point of the time-varying bounded delay. The value of the delay depends on numerous factors: the neural path time between the sensors of the muscles, the processing time by the central nervous system and the time to generate force after the muscle triggering [14]. It is known that a delay in a closed-loop control brings instability of a mechanical system, but previous works showed that the attentional demands of balance control (which could be assumed as the processing time) vary depending on the complexity of the task and the type of secondary task being performed [28]. By considering theses points, we could design a more coherent control law to stabilize human like non-linear models.

The H2AT represents a preliminary but essential step in order to understand the underlying parameters which define seated postural stability for people with SCI. Simulation results indicates that a mechanical model in descriptor form combined with an TS-UIO can be used to estimate internal unmeasured data. We wish to continue this work with an experimental step which would validate both the model and observer by feeding them with x and θ data measured by motion capture on real subjects with SCI. Eventually we will have the tools to identify individual stability control parameters and test them in various situation (perturbation, increased delay…) in order to better understand postural stability and be able to design better technological devices to facilitate balance control is seated position.

Conclusion

The Unknown Input Observer technique, an approach coming from the control theory community, is very seldom used in biomechanics but its advantages are to be highlighted. We can estimate internal forces in our model without having to compute the velocities and accelerations of the segments. This is, to our knowledge, the first attempt to understand how people with complete SCI maintain their sitting via an Unknown input observer in descriptor form.

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