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IFAC-PapersOnLine 49-5 (2016) 175–181

Unknown Input Observer for Understanding Sitting Control of Persons with Spine Cord Injury

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Abstract: The present paper introduces a simple model to study sitting control for persons with complete thoracic spine cord injury. The system is obtained via Lagrangian techniques; this procedure leads to a nonlinear descriptor form, which can be written as a Takagi-Sugeno model. A first attempt to estimate the sitting control in disabled people is done via an unknown input observer. The conditions are expressed as linear matrix inequalities, which can be efficiently solved. Simulation results validate the proposed methodology as the observations are coherent with and without perturbations.

Keywords: Unknown input observer, Takagi-Sugeno models, Inverted pendulum, Spine Cord Injury, Sitting movement, Descriptor Form.

1. INTRODUCTION

Sitting stability of the upper body is critical for performing activities of daily living (e.g. in transportation), this is especially true for persons with complete thoracic spinal cord injury (SCI) who lose both muscle control and sensitivity below the injury level. Because of its unstable nature (Crisco et al. 1992), the human spine must be stabilized even without external perturbation (e.g. braking in a car). These fine adjustments are usually done via activation of the trunk and intervertebral muscles (Silfies et al. 2003) but people with complete SCI cannot rely on any muscle activity below their injury level. Instead they can use their upper limbs and head in order to maintain the stability of the upper body as they are trained in rehabilitation (Janssen-Potten et al. 2001). The stability of the upper body has been described as the combination of various subsystems (e.g. redundant voluntary joint torques, passive viscoelastic behaviour of the joint… (Panjabi 1992)). Due to this complexity, linked-segment models of the human trunk are used to systemize thinking and test hypothesis one by one which is difficult to do with human observation. The obtained mathematical model is therefore needed in order to estimate the role of all the upper body segments (trunk, head and arms) in sitting stability under external perturbation.

Models used to study sitting stability are usually based on an active torque at the lumbar joint to stabilize the upper body with head, arms and trunk represented by one rigid segment, (Cholewicki et al. 1999; Tanaka and Granata 2007; Vette et al. 2010). Not only this representation is inapplicable to people with complete thoracic SCI because of their lack of trunk muscle activity but it is also both too simplistic and erroneous: it has been shown that under perturbations a seated subject's head will move (Thrasher et al. 2010) and we can predict that the arm will do the same. A model taking into account the action of the upper limbs and head is required for this specific topic.

Therefore, in this paper, the effect of the head and upper limbs displacement on the stabilization of an individual with SCI is modelled via an H2AT configuration, which can be seen as a variation of the inverted pendulum. Such displacement is generated by a force with time varying delay. This delay is inherent in biomechanical systems and represents the time to sense a perturbation and respond with muscle activation to stabilize the system (Reeves et al. 2007).

On the other hand, modeling mechanical systems often leads to nonlinear descriptor systems with the property of the inertia matrix being invertible (Lewis et al. 2003). The use of Takagi-Sugeno (TS) models has been widely employed for the analysis and design of nonlinear systems (Tanaka and Wang 2001). Effectively, via the sector nonlinearity methodology an exact TS representation can be obtained (Ohtake et al. 2001). The extension to TS descriptor models (Taniguchi et al. 1999) have been already used in biomechanics, for example, in (Guelton et al. 2008) an unknown input observer (UIO) has been developed for estimating variables, inputs. Nevertheless, the conditions are

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10.1016/j.ifacol.2016.07.109

in terms of bilinear matrix inequalities (BMI), which are difficult to solve. Linear matrix inequality (LMI) conditions have been recently derived in (Guerra et al. 2015). Those conditions can be efficiently solved via convex optimization techniques (Boyd et al. 1994). Later, results in (Guerra et al. 2015) have been extended to an UIO (Estrada-Manzo et al. 2015). This work adopts such methodologies. Its objective is to understand the way an individual with complete thoracic SCI controls his position by recovering internal variables of force generation. This task is performed via UIO and TS descriptor models.

The SCI open-loop being unstable, a first control law – that will act as an "internal" control law – being compatible with the observed human behaviour has to be derived in order to get a stable closed-loop system. The design of this control law is based on a time-varying input control law from (Yue and Han 2005) and has been adapted for the descriptor form. This design is out of the scope of this paper. To the best of our knowledge, there are no results in the literature for TS models in descriptor form with that kind of delay to study SCI stabilization.

This paper is organized as follows: Section 2 presents the modeling of H2AT via Lagrangian techniques, Section 3 explains the way the TS descriptor model is obtained as well as the main results to derive a UIO for the observation of the stabilization force, Section 4 provides the simulation results, Section 5 discusses the obtained results and future works, and Section 6 concludes the paper.

2. PROBLEM STATEMENT

The goal is to estimate the variables that make a seated person with SCI controls his position via the top of his body without invasive measurements.

2.1 Modeling

The H2AT pendulum is an extended version of the planar inverted pendulum consisting of two rods. The first one represents the trunk as a classical inverted pendulum while the second rod represents the head and arms slides at the top of the first one. The controlling force $F(t)$ will make the upper rod slide on the lower one. Figure 1 shows the H2AT system scheme.

This model is generic and just requires a minimum of biomechanical parameters. For the simulations, we consider a $80\,\text{Kg}$ male subject. As arms and trunk mass do not change between control subject and SCI subject (Jones et al. 2003), we can use regression rules to get segment mass and length (Dumas et al. 2007): $m_1 = 16.1\,\text{Kg}$ stands for the mass of the upper segment, corresponding to the head, neck, and arms; $m_2 = 26.64\,\text{Kg}$ is the mass of the trunk; $l_0 = 477\,\text{mm}$ is the length of the trunk; and $l_c = 276.66\,\text{mm}$ is the length of the centre of mass of the trunk. A full neck flexion with both arms stretched gives a value of $x = 105.27\,\text{mm}$ whereas an extension of the neck and arms gives $x = -75.18\,\text{mm}$ (Kapandji 2005). The resulting compact set is $\Omega_X = \begin{Bmatrix} -0.075 \le x \le 0.105 \\ -0.087 \le \theta \le 0.122 \\ -0.209 \le \dot{\theta} \le 0.209 \end{Bmatrix}$ in meters, radians and rad/s.

To obtain the dynamic equations of the system, we calculate its Lagrangian $L = K - U$ where K, U are the kinetic and potential energies of the system, respectively. Thus, consider $K = K_1 + K_2$ with $K_1 = \frac{m_1}{2}\left(l_0^2\dot{\theta}^2 + \dot{x}^2 + x^2\dot{\theta}^2 - 2l_0\dot{x}\dot{\theta}\right)$, $K_2 = \frac{1}{2}m_1 l_C \dot{\theta}^2$; and $U = U_1 + U_2$ with

$U_1 = m_1 g\left(l_0\cos(\theta) + x\sin\sin(\theta)\right)$, $U_2 = m_2 g l_c \cos(\theta)$.

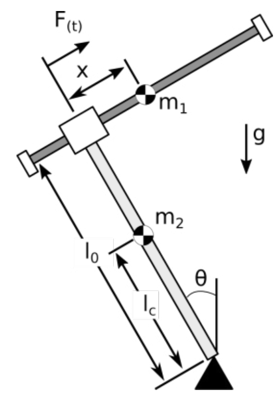

Fig. 1. H2AT Pendulum

Hence, by considering

$$\frac{d}{dt}\frac{\partial L}{\partial \dot{x}} - \frac{\partial L}{\partial x} = F(t-\tau) \quad \text{and} \quad \frac{d}{dt}\frac{\partial L}{\partial \dot{\theta}} - \frac{\partial L}{\partial \theta} = 0,$$

we obtain

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

where $J(x(t)) = m_1\left(l_0^2 + x^2(t)\right) + m_2 l_c^2$. The input includes a time-varying delay $\tau(t)$ due to neural transmission and the muscle force generation and is varying according to the individual; a classical range is for example $60\text{ms} \pm 10\text{ms}$. Of course, due to the absence of control of the trunk and intervertebral muscles, the model exhibits unstable open-loop behaviour, as shown for example, in Fig. 2, using the H2AT initial parameters at $t = 0\,\text{s}$: $\theta = -0.2\,\text{rad}$, $x = 0\,\text{mm}$, and $F = 100N$. Because of the gravity effect, the trunk should have continued in negative values but with x increasing fast, the trunk rotates in the opposite direction and ends up falling down.