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EFFECTS OF A CLOSED-LOOP PARTIAL POWER ASSISTANCE ON MANUAL WHEELCHAIR LOCOMOTION

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Abstract: In manual wheelchair locomotion, the large upper extremity loads and the repetitions of the propulsion movement increase the incidence of upper limbs injuries, pain and muscle fatigue. The main goal of this study was to investigate the influence of a closed-loop partial power assistance for manual wheelchairs through predictive simulations of a dynamic four-bar model. The applied control law applied can be seen as an impedance-like control, but it does not require force measurement. The simulation results indicate that this strategy can reduce joint torques without significantly altering the typical kinematic pattern of manual wheelchair locomotion.

Keywords: closed loop control, wheelchair locomotion, optimal control, biomechanics

1. INTRODUCTION

Data from IBGE (Brazilian Institute of Geography and Statistics) show that about 14 million Brazilians have motor impairment. The main assistive technology used for locomotion is the wheelchair, but manual propulsion is inefficient and require large energy consumption, according to van der Woude et al. (2001). Besides, manual wheelchair locomotion can cause upper extremity injuries and pain (Boninger et al., 2002; Cooper et al., 1999).

To overcome these issues, wheelchairs with partial assistance, also called *Pushrim Activated Power Wheelchair* (PAPAW), were developed. This type of wheelchair typically has an electromechanical element coupled to the rear wheel axle. The wheelchair with partial assistance still requires the user to apply force to the pushrim, but part of the propelling action is carried out by the motor. This solution differs from fully assisted electric wheelchairs, which do not allow the user to engage in physical activity and is, therefore, detrimental to his health.

The most common strategies used for partial wheelchair assistance are (i) the constant assistance that is established when the user touches the propulsion rim, and (ii) the proportional assistance that generates extra torque in the rear wheel proportional to the torque applied by the user. These two strategies were compared in Guillon et al. (2015).

Cuerva et al. (2016) compared these strategies to a mechanical impedance control-based strategy. The particularity of the strategy based on impedance control is that it allows changing the apparent mass or friction properties of the wheelchair-user system.

Cuerva (2017) proposed improvements to the assistance strategy using a modified wheelchair-user model. Instead of representing the system as a lumped mass subject to resistive and propelling forces, an approach commonly found in the literature, he proposed to represent it more realistically by means of a four-bar system, based on a model developed in Ackermann et al. (2014) to study the influence of pavement inclination and mass on system performance in steady-state locomotion.

The strategy proposed by Cuerva et al. (2016) requires the measurement of the force exerted by the user on the pushim and, therefore, specific instrumentation such as the use of load cells is needed. The present work intends to analyze the performance of a control system based only on the speed of the wheelchair, a more accessible variable. The performance of the proposed closed-loop partial assistance strategy is evaluated via predictive simulations obtained by solving an optimal control problem, considering the parameters of the controller also as free optimization variables of the problem.

Section 2 of this article contains the model of the wheelchair locomotion and the formulation of the optimal control problem. In section 3, the predictive simulation results are presented. Section 4 contains a discussion of the results and section 5 concludes the paper.

2. METHODS

2.1 Model of the wheelchair-user system

The multibody model employed in this study is shown in Fig.1. The wheelchair-user system is contained in the sagittal plane and consists of four rigid bodies: arms, forearms, rear wheel, and body structure with the wheelchair, considering

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bilateral symmetry. Therefore, the system becomes a moving four-bar system: the arm, the forearm, the wheel and the rigid body representing the wheelchair structure and the user's body. The shoulder and elbow joints and the wheel's axle were considered ideal hinge joints and there is no slip between the wheels and the floor, according to Ackermann et al. (2014).

The model is actuated by elbow and shoulder moments as well as by the wheel motor, as indicated in Fig.1.

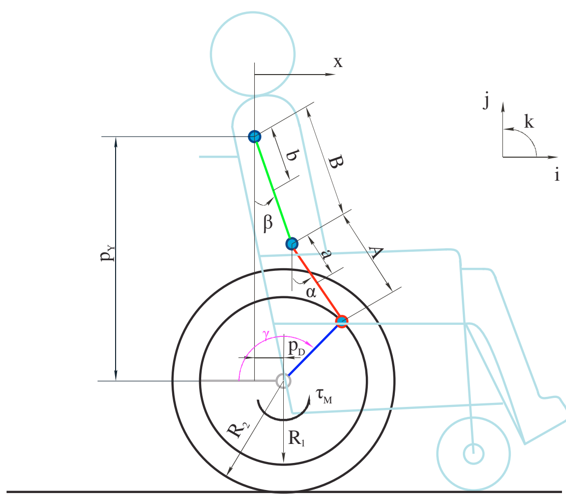

Fig. 1. Model of the wheelchair-user system in the propulsion phase: wheels (black), arm (green), forearm (red).

The wheelchair locomotion is divided into two phases, the propulsion phase and the return phase. In the propulsion phase, the hands are in contact with the propulsion rim and this interaction. In the return phase, the hands are not in contact with the propulsion rim.

In the return phase, the model has three degrees of freedom, where β is the angle between the arm and the vertical, α is the angle between the forearm and the vertical, and x is the horizontal displacement of the shoulder, linked to the wheelchair, Fig. 1. The generalized coordinates are represented by the vector $q=\left[x \quad \beta \quad \alpha\right]^T$, used to describe the multibody system configuration.

In the propulsion phase, the contact between the hand and the rim is modeled as a hinge joint and the model has a single degree of freedom, characterized by the horizontal displacement *x* of the chair.

The segment's centers of mass, moments of inertia, lengths and masses were taken from the literature (Winter, 2009; Holzbaur et al., 2005; Delp et al., 2007) for a 1.70 m, 70 kg person.

The equations of motion representing the dynamics of the multibody system were determined using the Newton-Euler formalism (Schiehlen, 1997), resulting in

$$M\left(q\right)\ddot{q}+k\left(q,\dot{q}\right)=k_e\left(q\right)+H\left(q\right)\begin{bmatrix}\tau_M\\ \tau_O\\ \tau_C\end{bmatrix}+Q\left[F_{rol}\right] \quad (1)$$

for the return phase, and

$$M\left(q\right)\ddot{q}+k\left(q,\dot{q}\right)=k_e\left(q\right)+G\left(q\right)\begin{bmatrix}F_X\\ F_Y\end{bmatrix}+H\left(q\right)\begin{bmatrix}\tau_M\\ \tau_O\\ \tau_C\end{bmatrix}+Q\left[F_{rol}\right]. \quad (2)$$

for the propulsion phase, where M is the matrix of mass, k is the vector of generalized centrifugal and Coriolis forces, k_e is the vector of applied generalized forces, including those related to gravity, G transforms the components of the contact forces $\left(F_X \text{ e } F_Y\right)$ in generalized forces, H transforms the moments at the shoulder, elbow and motor (τ_M, τ_O e τ_C, respectively) into generalized forces, and Q transforms the rolling resistance force (F_{rol}) into a generalized force.

In the contact phase, the hinge joint between the hand and the pushrim imposes two additional kinematic constraints, $c\left(x,\beta,\alpha\right)=0$, which reduced the number of degrees of freedom from 3 to 1 and is associated with the two components of the reaction forces $\left(F_X \text{ e } F_Y\right)$.

2.2 Formulation of the optimal control problem

The predictive simulation problem used to test the performance of the proposed control law represent a transient maneuver starting from rest and achieving 0.5 m/s at the end of the first propulsion phase.

An optimum control formulation was used to reproduce the implicit optimization of the movement performed by a human during the propulsion cycle with a motorized closed loop assistance based on the speed measurement of the chair. The goal of this formulation was to provide the optimal generalized coordinates profiles $\left(q,\dot{q}\right)$, and the shoulder, elbow and motor moment estimations. In this work, the total energy demanded by the motor along the investigated maneuver was limited to save energy.

The variables of the optimal control problem are organized into two groups: control variables, which are the torques variables of the multibody system (τ_M, τ_O e τ_C), and state variables, which are composed by the position vector $\left(q\right)$, generalized velocities $\left(\dot{q}\right)$, and as components of the contact force $\left(F_X \text{ e } F_Y\right)$.

The simulation refers to a complete cycle of the wheelchair locomotion (propulsion and return), whose results are obtained through the solution of the optimal control problem. To solve the optimal control problem, we used the software PROPT (http: /tomdyn.com) that transcribes the optimal control problem into a nonlinear programming problem, resulting on a standard nonlinear programming problem that can be solved, for example, by SNOPT solver.

constraints of the optimization problem are: i) the equations of motion (1) and (2); ii) the two kinematic constraints imposed by the link between the hand and the pushrim in the contact phase; iii) continuity between phases, ensuring that the final states of one phase are the same as those in the beginning of the other one; iv) it was imposed that the system is initially at rest ($v(0)=0$), v) the final speed at the end of the propulsion phase if 0.5 m/s; vi) the positions of the hands at the beginning of the contact phase on the pushrim corresponding to an angle $\gamma = 70^\circ$ (Fig.1) and at the end of the corresponding contact phase at an angle $\gamma = 120^\circ$; v) we also imposed an upper bound of 5 J to the energy consumed from batteries by the motor along this one-cycle maneuver.

We defined for this problem a cost function given by the integral of the squared moments for both phases of the propulsion cycle.

$$J = \int_0^{T_1} \left(\tau_O{}^2 + \tau_C{}^2 \right) dt + \int_{T_1}^{T_1+T_2} \left(\tau_O{}^2 + \tau_C{}^2 \right) dt \quad (3)$$

where T_1 is the duration of the propulsion phase, and T_2 is the duration of the return phase.

Aiming at real applications, the control law of the optimum control problem formulation was defined in closed loop based on measurements of the chair speed, as depicted in Fig.2.

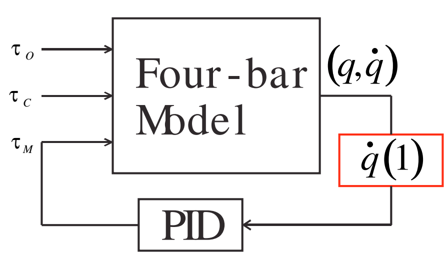

Fig. 2. Block diagram of the closed-loop control.

The block highlighted in red is a liner transformation that selects the wheelchair speed from the generalized coordinates and velocities vector.

A PD-type control law was chosen because it is quite usual, but also because it can implicitly represent an impedance control law, as pointed out by Monteiro Junior (2017). An impedance control aims to maintain the relationship between force and kinematics of motion, apparently altering the parameters of the equation of motion, such as mass and coefficient of friction. If we consider that the wheelchair can be represented approximately by a concentrated mass block, the PD feedback apparently alters these parameters of the wheelchair-user system. This control law can be represented in state space through the following ODE

$$\tau_M = k_P \cdot \dot{x} + \left(\frac{k_D}{T_D} \right) \cdot \left(\dot{x} - n \right) \quad (4)$$

where k_P is the constant related to proportional action, k_D is the constant of derivative action, T_D it is the filtering time constant of the derivative action which helps avoiding noise amplification ($T_D = 0{,}1\ s$), and n is the control state variable of derivative action.

3. RESULTS

In this session, the predictive simulation results obtained by the solution of the corresponding optimal control problem, are presented. A rolling resistance force of 15 N was included.

Predictive simulations were generated for three scenarios: for the PD action, for the open-loop assistance and for the wheelchair without assistance. It was found that the PD controller implies in a performance close to the reference performance. This reference was established by the solution of the optimal open-loop control problem, that is, without any structural constraints imposed by the control law and without any linkage to a measured variable. Therefore, this situation was considered here as a performance limit to evaluate the controller action.

Some control actions were investigated, but since the PD action was the one that produced the best performance, the following figures deal only with the PD action in comparison to the wheelchair without assistance. The simulated data of the open wheelchair assisted wheelchair represent the best expected results, but they are not practical.

Tables 1 and 2 summarize the results for the three investigated scenarios: the proportional-derivative action (PD), the open-loop assisted wheelchair and the manual wheelchair without assistance. Fig. 3, in turn, illustrates the kinematics of the movement of the wheelchair-user system obtained by solving of optimal control problem, where the element in red represents the arm and the element in black represents the forearm, and the green circles indicate points of contact at the propulsion ring.

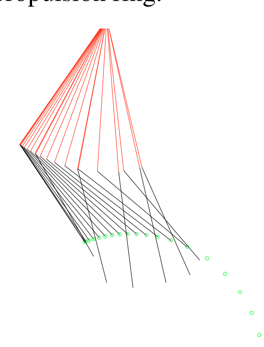

Fig. 3. Simulated kinematics of upper limb.

Fig. 4 shows the predicted joint angles and wheelchair displacement for the two practical scenarios: optimal maneuver with the PD action and optimal locomotion without assistance. Fig.5 shows the corresponding joint moments and motor torque (only for the PD action scenario).

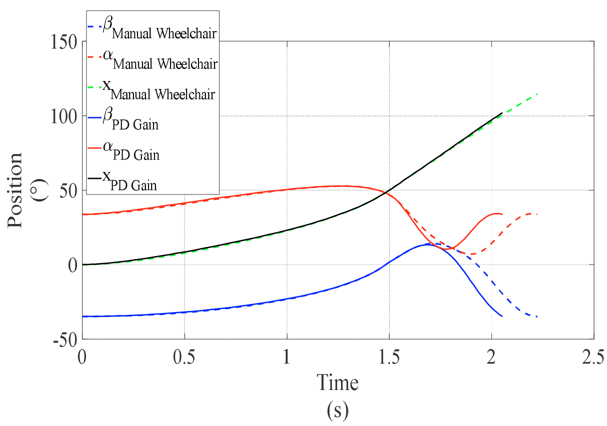

Fig. 4. Predicted kinematics.

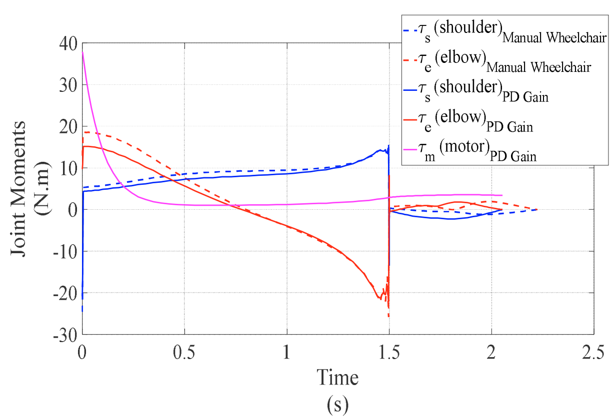

Fig. 5. Predicted joint and motor moments.

The results related to the cost function and the controller gains are presented in Tab.1 and Tab.2, respectively.

Table 1 – Cost function

Assisted wheelchair		Assisted wheelchair		Wheelchair without assistance	
PD controller		Open loop			
Cost:	258.8	Cost:	166.6	Cost:	323.2

Table 2 – Controller gains

k_P Gain:	6.99	k_D Gain:	-0.48

The controller remains on and its gains are the same for both phases. In this way, it is possible to implement a closed-loop control strategy that does not require a sensor connected to the pushrim.

4. DISCUSSION

The assistance problem was formulated as an optimal control problem for both the open loop and the closed-loop scenarios. The open loop problem was solved to establish a maximum performance limit and the closed loop problem was proposed so that the assistance did not have to be redesigned for each new maneuver. For comparison purposes, an optimal control problem was also solved for the unassisted wheelchair. The nominal design of the closed-loop control law was performed for a single maneuver, but it is assumed that the performance in other maneuvers would still be advantageous when compared to an unassisted maneuver.

The plots in Fig.4. illustrate that the wheelchair with the PD controller reproduces approximately the same cyclic movement performed by the manual wheelchair without assistance, due to the kinematic constraints that impose this movement pattern. It should be noted that the propulsion cycle is executed more quickly with the assistance.

The results in Fig. 5 show that motor assistance is used to overcome the inertia of the system at the beginning of the propulsion phase and then decaying exponentially until a certain instant, after which it remains constant for the remainder of the propulsion cycle. Furthermore, it illustrates that the human propulsion effort is lower compared to the one in a manual, unassisted wheelchair. This demonstrates the usefulness of an assisted wheelchair, also because joint loads are one of the major factors associated with increased risk of injury.

Table 1 illustrates that the PD controller reduces the cost function value in relation to the unassisted wheelchair. It is noted that the closed-loop control strategy resulted in a cost value between the open-loop performance limit and the one for the unassisted wheelchair, indicating that the strategy studied is beneficial.

Table 2 presents the optimal values for the controller gains of both proportional and derivative actions. The gains are identical in the propulsion phase and return phase. Note that by doing so it is not necessary to implement a sensor to indicate the instant of switching the control law at the end of each phase. However, keeping fixed the gains during the two phases may represent a reduction of performance when compared to an assistance that allows the switching of control law.

It is noted that the joints moments are smaller for the closed-loop solution compared to the optimal unassisted solution. The PD control law was chosen because, if the wheelchair model is approximated by a block of lumped mass, the PD control law can alter the apparent inertia and viscous friction of the model. The proposed control law differs from that proposed by Cuerva et al. (2016) to control the impedance of the wheelchair in the sense that it does not need the information of the force produced by the user.

5. CONCLUSIONS

The objective of this work was to investigate the effect of a closed-loop power assistance with a PD action requiring only velocity measurements along on a transient propulsion cycle.

This study applied the an optimum control formulation in order to predict and investigate the effects of a closed-loop PD controller on the joint moments and movement pattern of a wheelchair-user system for a transient maneuver.

It is noted that the kinematic pattern of motion is not strongly altered by the inclusion of the motor assistance inserted by the control law because the cyclic movement is given by kinematic constraints that strongly impose this pattern. However, the controller proved useful for reducing the cost function. The solution of the optimal control problem showed that the joint moments found in the closed loop scenario are close to those obtained for the open-loop optimal assistance scenario, considered here as a reference of performance.

The optimal open-loop assistance scenario does not lend itself to a real implementation of assistance, but it did highlight the maximum limit that the closed-loop strategy could achieve.

As a continuation of this research, it is proposed to investigate the design of the control law for a more complete maneuver that involves acceleration, steady-state locomotion and braking. Other future work would be to investigate the design of the control law without having to solve an optimal control problem to find its parameters. It is argued that it is possible to find a law of control that alters explicit portions of the equations of motion, manipulating them conveniently for any maneuver.

REFERENCES

Ackermann, M., Leonardi, F., Costa, H., Fleury, A. (2014). Modeling and control formulation for manual wheelchair locomotion: The influence of mass and slope on performance. *International Conference on Biomedical Robotics and Biomechatronics (2014 5th IEEE RAS & EMBS)*, p. 1079-1084, São Paulo, Brazil.

Ackermann, M., Leonardi, F., Costa, H. (2015). A modelling framework to investigate the radial component of the Pushrim force in manual wheelchair propulsion. *MATEC Web of Conferences,* v.35, 02008, Lisbon, Portugal.

Boninger, M.L., Souza, A.L, Cooper, R.A., Fitzgerald, S.G, Koontz, A.M., Fay, B.T. (May 2002). Propulsion patterns and Pushrim biomechanics in manual wheelchair propulsion. *Arch Phys Med Rehabil,* v. 83, p. 718-723.

Boninger, M.L., Cooper, R.A., Shimada, S.D., Rudy, T.H. (May 1998). Shoulder and elbow motion during two speeds of wheelchair propulsion: a description using a local coordinate system. *Spinal Cord,* v. 36(6), p. 418-426.

Cooper, R.A., Quatrano, L.A., Axelson, P.W., Harlan, W. (1999). Research on physical activity and health among people with disabilities: a consensus statement. *Jornal of Rehabilitation Research and Development,* v. 36(2), p. 142.

Cuerva, V.I., Ackermann, M., Leonardi, F. (2016). A comparison of different assistance strategies in power assisted wheelchairs using an optimal control formulation. *Proceedings of the Sixth IASTED International Conference Modeling, Simulation and Identification (MSI 2016)*, v. 840-842, Campinas, Brazil.

Cuerva, V.I. (2017). *Controle da locomoção assistida de cadeiras de rodas manuais por meio do controle de impedância: análise via controle ótimo*. p. 167, Thesis (Mechanical Engineer Master Degree) Centro Universitário FEI, São Bernado do Campo.

Cuerva V.I., Ackermann, M., Leonardi, F. (2017). The influence of speed and slope angle on wheelchair propulsion patterns: an optimal control study. *24th ABCM International Congress of Mechanical Engineering.* P. 1154, Curitiba, Brazil.

Delp, S.I., Frank, C.A., Allison, S.A., Loan, P.,Habib, A., John, C.T., Guendelman, E., Thelen, D.G. (2007) Opensim: open-source software to create and analyse dynamic simulations of movement. *IEEE Transactions on Biomedical Engineering.* v. 54(11), p. 1940-1950.

Guillon, B., Van-Hecke, G., Iddir, J., Pellegrini, N., Beghoui, N., Vaugier, I., Figére, M., Pradon, D., Lofaso, F. (2015). Evaluation of 3 pushrim-activated power-assisted wheelchairs in patients with spinal cord injury. *Archives of physical medicine and rehabilitation.* v. 96(5), p. 894-904.

Holzbaur, K.R.S., Murray, W.M., Delp, S.I. (2005). A model of upper extremity for simulating musculoskeletal surgery and analysing neuromuscular control. *Annals of Biomedical Engineering.* v. 33(6), p. 829-840.

IBGE, (2010). *Censo demográfico: Características gerais da população, religião e pessoas com deficiência*. Rio de Janeiro, Brazil.

Monteiro Junior, S., Delijaicov, S., Ackermann, M., Leonardi, F. (2017) Impedance Control For Assistance In Cargo Handling. *Proceedings of the 24th International Congress of Mechanical Engineering*, São Paulo, Brazil.

Schielen, W. (1997). Multibody system dynamics: roots and perspectives. *Multibody System Dynamics.* v. 1, p. 149-188.

Van der Woude, L.H.V., Veeger, H.E.J., Dallmeijer, A.J., Janssen, T.W.J., Rozendaal, L.A. (2001). Biomechanics and physiology in active manual wheelchair propulsion. *Medical Engineering and Pshysics*. v. 23, p. 713-733

Winter, D.A., (2009). Biomechanics and motor control of human movement. *John-Wiley & Sons Inc.* New York, 4th ed.