Journal of Mechanical Engineering and Biomechanics, Volume 3, Issue 5, Page 50-56, 2019
ISSN- 2456-219X

A model-based control strategy for assisted manual wheelchairs: a simulation study

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Special Issue: 6th Brazilian Meeting of Biomechanical Engineering (ENEBI 2018)

Research Article

Article History
Received 09 /10/2018
Revised 17/12/2018
Accepted 08/01/2019

Recommended by Editors Adriano A. G. Siqueira and Zilda de Castro Silveira

Abstract

Power-assisted wheelchairs constitute a promising alternative to traditional manual wheelchairs, which often require excessive effort from the user and may lead to upper extremity pain and injury. The assistance strategies reported in the literature include providing an assistance proportional to the applied pushrim torque and the imposition of a 1st order dynamics between wheelchair speed and applied force, both requiring the measurement of the force applied by the user on the pushrim and, therefore, associated to higher complexity and costs. These assistance strategies are suitable for some types of maneuvers, but may result in poor performance because of the unnatural interaction with the user. This paper proposes a PD-like control strategy able to change the values of apparent system parameters such as mass and friction while not requiring the measurement of pushrim forces. The effects of the resulting control strategy on locomotion performance were investigated using an optimal control approach in two operating conditions, a transient maneuver and steady-state locomotion. The results show that the proposed closed-loop strategy is able to reduce upper limb joint moments and decrease propulsion frequency, potentially mitigating upper extremity loads and the resulting risk of injury..
DOI:https://doi.org/10.24243/JMEB/3.5.213

Keywords: power-assisted wheelchairs; wheelchair locomotion; predictive simulations; optimal control

1 Introduction

In Pushrim Activated Power Wheelchairs (PAPAW) the human propulsion is assisted, rather than replaced, by a motor. These wheelchairs are promising as they provide the essential physical activity to the user while avoiding excessive loads on the upper extremity which may lead to the well-known high incidence of pain and injury in long-term manual wheelchair users [1].

The adequate amount of assistance and the appropriate interaction dynamics between the user and the wheelchair are central to the performance of the PAPAW. Some of the assistance strategies reported in the literature involve amplifying the torque applied by the user [2]. Others aim at imposing a favorable first order dynamics between the forces applied by the user on the pushrim and the resulting wheelchair speed [3], [4]. It has been reported that, although showing good performance in steady-state locomotion, PAPAWs performed poorly in transient conditions, characterized by recurrent acceleration and deceleration phases in confined environments [5]. Other reported weaknesses of the PAPAWs

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are related to control strategies that result in unnatural interaction with the user [5]-[7], and poor compensation of external disturbances such as the gravitational force component in ramps [8]-[10].

This work proposes a PD controller whose gains can be associated to changes in critical system parameters such as apparent mass and friction, without changing the original nonlinear system dynamics and, therefore, providing a natural interaction with the user. Another advantage is that the proposed approach requires measurement of wheelchair speed only. Unlike other approaches reported in the literature, no pushrim force measurements are needed. The strategy performance is assessed via predictive simulations of a transient maneuver and steady-state locomotion, obtained by solving optimal control problems.

Section 2 of this paper introduces the wheelchair-user model adopted in the predictive simulations to assess the wheelchair performance, explains the control strategy proposed, and presents the optimal control problem to be solved to generate the predictive simulations. Section 3 presents the predictive simulation results, and sections 4 and 5 contain the discussion and the conclusion, respectively.

2 Methods

2.1 Wheelchair-user Model

Wheelchair locomotion is divided into two phases, the contact or propulsion phase, in which the hands are on contact with the pushrim, and the return phase, in which the hands are not in contact with the pushrim and the upper limbs are repositioned in preparation to the next contact phase. The adopted wheelchair-user model (Fig. 1) assumes bilateral symmetry and is composed of 4 rigid bodies in planar motion (arms, forearms, wheels and body+wheelchair) connected by ideal hinge joints at shoulder, elbow and wheel axis. In the contact phase, the contact of the hand with the pushrim is represented by a fourth hinge joint. The model is actuated by moments at the shoulder τ_s and elbow τ_e, as well as by the motor torque τ_m applied on the wheels. The shoulder joint is assumed fixed to the wheelchair and no wheel slipping is allowed.

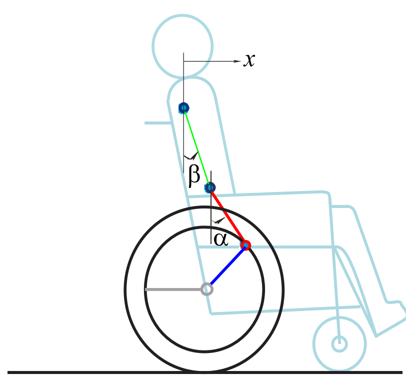


Fig.1 Model of the wheelchair-user system in the contact phase: wheels (black), arms (green), forearm (red).

In the return phase, the model has three degress of freedom (DoF) and its kinematics is characterized by three generalized coordinates in $q = [x \quad \beta \quad \alpha]^T$, where x is the horizontal displacement, β is the shoulder flexion angle, and α is the angle between the forearm and the vertical. In the contact phase, the model has a single DoF due to the additional constraints arising between the hand and the pushrim. The multibody system model parameters were adopted from [11], and were determined for a 12 kg wheelchair and using anthropometric parameters extracted from [12] for a 1.7 m, 70 kg user. The rolling resistance force was assumed as 20 N according to [13], [14].

The equations of motion for both phases were determined using the Newton-Euler Formalism [15] yielding

$$M(q)\ddot{q} + k(q,\dot{q}) = k^e(q) + H(q)\begin{bmatrix}\tau_s \\ \tau_e \\ \tau_m\end{bmatrix} + Q[F_{roll}] \quad , \quad q = \begin{bmatrix}x \\ \beta \\ \alpha\end{bmatrix} \tag{1}$$

for the return phase, and

$$M(q)\ddot{q} + k(q,\dot{q}) = k^e(q) + G(q)\begin{bmatrix} F_x \\ F_y \end{bmatrix} + H(q)\begin{bmatrix} \tau_s \\ \tau_e \\ \tau_m \end{bmatrix} + Q[F_{roll}] \quad , \quad c(x,\beta,\alpha) = 0 \qquad (2)$$

for the contact phase, where M is the mass matrix, k is the vector of generalized Coriolis and centrifugal forces, k^e is the vector of generalized forces including gravity and viscous friction forces, H transforms the joint and motor torques into generalized forces, Q transforms the rolling resistance force F_{roll} into generalized forces and G transforms the pushrim contact force components F_x and F_y into generalized forces. In the contact phase, the additional constraints $c=0$ are active.

2.2 Model-based Feedback

The proposed control strategy (Fig.2) is similar to a proportional-derivative (PD) controller, excepting the term related to a constant force in the opposite direction of the velocity $\dot{x}$, whose function is compensating opposing constant applied forces such as the rolling resistance force. Note that there is no reference signal for the feedback controller so that the parameters k_D, $k_P$and k_R are directly associated to changes in the apparent system parameters mass, viscous damping and rolling resistance force, respectively, with no change in the system original dynamics. Because of this, controller gain tuning is not required.

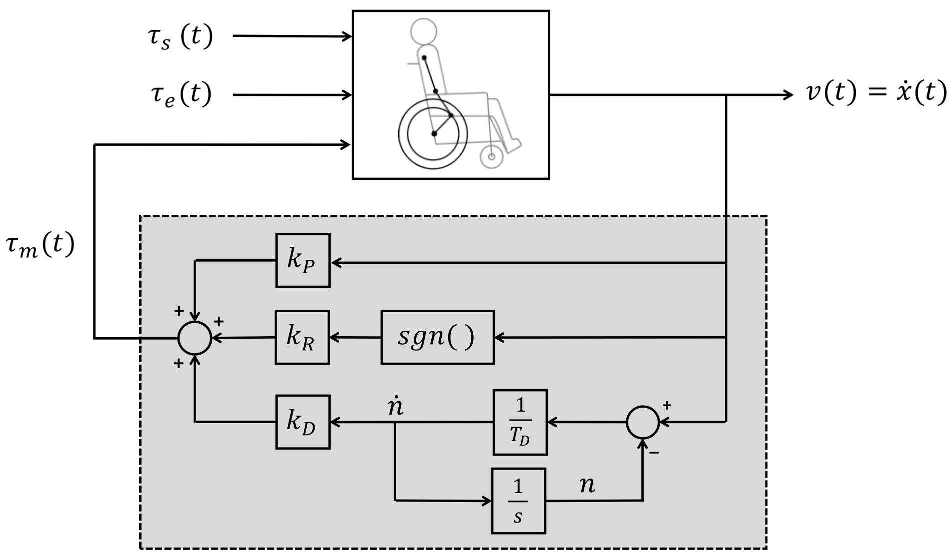


Fig.2 Block diagram of the proposed closed-loop control

The speed is the only measurement needed for the implementation. The acceleration is estimated by means of an observer through a first-order dynamics to avoid noise amplification (Fig. 2), with a time constant of T_D = 0.01 s.

2.3 Simulation Approach

Numerical simulations were performed to assess the performance of the proposed approach in two representative operation conditions: i) steady-state, periodic locomotion, with an average speed of 0.5 m/s; ii) a transient maneuver in which the user starts from rest and achieves 0.5 m/s after a complete propulsion cycle. These simulations required the solution of an optimal control problem to predict system response under the different investigated conditions. The optimal control package PROPT **[16]** which transforms the problem into a Nonlinear Programming Problem (NLP) using the direct collocation method was used.

The optimal control problem approach, similar to the one reported in **[7]**, consists in searching for the time histories of the controls $\tau_s(t)$, $\tau_e(t)$ and $\tau_m(t)$, system states $q(t)$, $\dot{q}(t)$ and $n(t)$, contact forces $F_x(t)$ and $F_y(t)$ as well as the durations of the contact and recovery phases, T_1 and T_2, that minimize a cost function representing user effort as

$$J = \int_0^{T_1} ({\tau_s}^2 + {\tau_e}^2) dt + \int_{T_1}^{T_1+T_2} ({\tau_s}^2 + {\tau_e}^2) dt \quad (3)$$

The problem is subject to a series of constraints including: *i*) the equations of motion in both phases, Eqs. (1) and (2); *ii*) the first-order dynamics imposed by the control law; *iii*) the kinematic constraints between the hands and the pushrims$c(x, \beta, \alpha) = 0$ in the contact phase; *iv*) the continuity between phases; *v*) the initial and final positions of the hand on the pushrim ($\gamma_i = 70°$ e $\gamma_f = 120°$); *vi*) maximum motor torque of 37.62 Nm. Additional constraints are added for each operation condition. For the steady-state locomotion, constraints that ensure periodicity and an average locomotion speed of 0.5 m/s are included. For the transient maneuver, initial and final velocities are set to 0 and 0.5 m/s, respectively.

In this study, we performed three simulation types for each operation condition (steady-state and transient): (i) reference simulation without assistance, for which $\tau_M = 0$, representing the pure manual locomotion; (ii) reference simulation with assistance, where an optimal motor torque profile $\tau_m(t)$ is determined in open-loop, providing the minimum achievable effort profiles; and (iii) simulation of assisted locomotion in closed-loop for assessment of the proposed controller performance, where the gains are set so as to reduce the apparent mass, viscous damping and rolling resistance force by 45%. Simulation types (i) and (ii) provide a basis for the comparison of the proposed closed-loop control law. Simulation (i) provides the upper bound on muscle effort as there is no motor assistance, while simulation type (ii) provides the lower bound on muscle effort as the motor torque is allowed to assume an optimal profile.

3 Simulation Results

Figure 3 illustrates the predicted kinematics of the arm and forearm along a complete gait cycle for the reference simulation without assistance, i.e. for simulation type (i), for the steady-state locomotion.

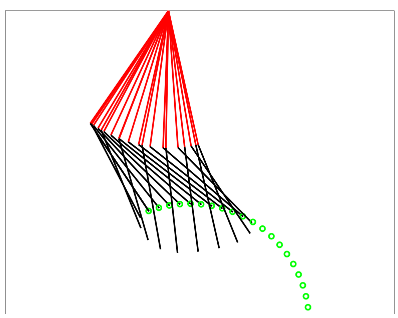


Fig.3 Predicted kinematics of upper extremity for the reference simulation without assistance (type i).

Table 1 shows the cost function values, representing user effort, for each combination of simulation type (no assistance, assistance in open-loop and proposed assistance in closed-loop) and operation condition (steady-state locomotion and transient maneuver).

Table 1 Cost Function (*J*) value (N^2m^2s)

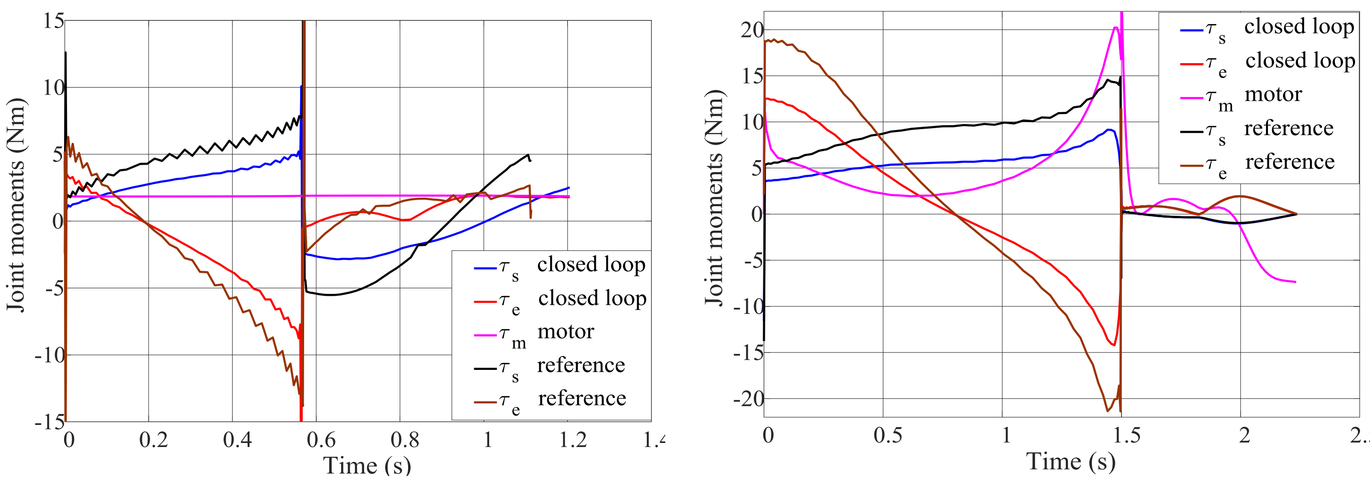
Simulation type	steady-state locomotion	transient maneuver
Type i (no assistance)	67.2	335.5
Type ii (open-loop assistance)	8.1	117.8
Type iii (closed-loop assistance)	18.6	128.7

Table 2 reports the recovery phase (T_1), the contact phase (T_2) and the complete cycle (T_C) durations for the steady-state locomotion.

Table 2 Phase durations for predicted steady-state locomotion

	No assistance Type i	Open-loop assistance Type ii	Closed-loop assistance Type iii
Contact phase (T_1)	0.572 s	0.529 s	0.566 s
Recovery phase (T_2)	0.480 s	0.626 s	0.639 s
Complete cycle ($T_C = T_1 + T_2$)	1.052 s	1.155 s	1.205 s

Figure 4 shows the predicted joint moments and motor torque profiles for the steady-state locomotion (left) and for the transient maneuver (right). Type ii simulation results (for open-loop assistance) were omitted for clarity.



a) Steady-state locomotion ***b) Transient maneuver***

Fig.4 Predicted joint moments and motor torque profiles for closed-loop assistance

4 Discussion

The stick-figure in Fig. 3 shows that the simulation strategy was successful in predicting realistic locomotion patterns and is a useful tool to investigate the effect of closed-loop control strategies for power-assisted wheelchairs. According to the results in Table 1, the proposed closed-loop strategy showed the expected behavior in terms of required user effort (cost function value) as it remained between the upper and lower bounds represented by the cost function values corresponding to the pure manual propulsion and the optimal open-loop assistance, respectively, for both, the steady-state locomotion and for the transient maneuver. This means that the user effort for the proposed control law was reduced compared to the pure manual propulsion in both operation conditions, but was greater than the one resulting from a hypothetical optimal motor torque profile, as expected.

The reduction in user effort is also evidenced by the joint moment profiles in Fig. 4. In both operation conditions, a substantial reduction in the elbow and shoulder moments is observed along the whole propulsion cycle for the proposed assistance strategy when compared to the pure manual propulsion. Table 2 reveals another potential benefit of the proposed control strategy. An increase in total cycle duration due to a longer recovery phase in the steady-state locomotion is observed, indicating that the proposed strategy could lead to a reduced motion frequency, mitigating the risk of upper extremity injuries [1].

Note that the proposed control law alters the system apparent parameters mass, viscous damping and rolling resistance force without changing the overall nonlinear system dynamics, an approach that should lead to a more natural interaction between the user and the assisted wheelchair and a more straightforward interpretation and adjusting of the degree or intensity of assistance. Moreover, the proposed control law does not require the measurement of the forces applied by the user on the pushrim, which constitutes a major advantage for practical implementation and affordability. Therefore, this approach differs from others reported in the literature, because it does not impose a fixed dynamic or require the measurement of the torque applied by the user. Thus, potentially providing a more natural assistance to the user, and favoring the practical implementation of the approach.

5 Conclusion

A PD-like control strategy for power-assisted wheelchairs was proposed and investigated by means of predictive simulations of steady-state locomotion and a transient maneuver, obtained by solving an optimal control problem. The resulting performance was compared to reference simulations of the pure manual propulsion and of the propulsion with an optimal, open-loop motor assistance. The results show a successful reduction in joint moment profiles and user effort in both operation conditions, as well as a decreased propulsion frequency in the steady-state locomotion. These positive outcomes are added to two desirable features of the proposed control law, the natural interaction with the user resulting from the maintenance of the original system dynamics, and the unneeded measurement of pushrim forces, making this a promising control strategy for power-assisted wheelchairs.

Future studies will investigation the controller performance under a larger set of different conditions such as inclines, curves and locomotion in different velocities. An experimental verification of these results is also planned.

Acknowledgements

The authors acknowledge the Brazilian National Council for Scientific Development (CNPq) for its support through the grant 458717/2013-4.

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