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Modeling and Optimal Control Formulation for Manual Wheelchair Locomotion: the Influence of Mass and Slope on Performance*

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Abstract — **A framework to generate predictive simulations is proposed to investigate the influence of system's mass on manual wheelchair locomotion. The approach is based on a model of wheelchair propulsion dynamics and an optimal control formulation. In this study, predictive simulations of steady-state wheelchair locomotion are generated for different combinations of model mass and uphill slope inclination angle. The results show that the influence of system's mass is negligible in level surfaces in steady-state, a finding which agrees with experimental observations in the literature. On the other hand, the results show that the influence of mass on slopes is critical, with large increases in propulsion effort with system's mass, even for slight inclination angles. This shows the importance of reducing wheelchair mass for improving locomotion performance, particularly in overcoming obstacles and ramps. Decreasing the wheelchair's mass may not be sufficient. Therefore, and on the light of these findings, we propose the reduction of system's apparent mass through the implementation of an impedance control scheme in power-assisted wheelchairs.**

I. INTRODUCTION

In Brazil, the number of estimated long-term wheelchair users exceeds 5 million. In spite of the importance of this assistive device in the lives of millions of people around the world, wheelchair locomotion with conventional manual wheelchairs requires large energy consumption and is one of the least efficient means of locomotion [1]. Furthermore, the prevalence of upper-extremity injury and shoulder pain is large among manual wheelchair users [2, 3].

In order to improve wheelchair locomotion, many studies have investigated the influence of different adjustments on quality of locomotion and proposed modifications in conventional wheelchair design as well as alternative propulsion concepts [4]. Reducing weight through new materials and structural improvements has been the focus of wheelchair manufacturers. However, experimental studies on the effect of mass on locomotion performance on level surfaces [5, 6] have been inconclusive. Recent computational studies using wheelchair locomotion models [7-9], on the other hand, have not addressed the influence of system's mass on performance.

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More recently, "power-assisted wheelchairs" have been proposed [10], whose aim is to assist users of manually propelled wheelchairs by providing actuation to the wheels. This concept provides additional moment on the wheels to reduce user's effort, but does not replace user's actuation as in motorized wheelchairs. This approach can potentially allow patients to remain active without exceeding their physical capacity, with potential cardio-vascular benefits and improved mobility. However, the adequate amount of assistance and the best actuation strategies will depend on user's fitness, disability, and terrain conditions. One possible actuation strategy, which is currently under investigation by this group, is to reduce the system's apparent mass or inertia through the implementation of an impedance control scheme [11]. Here again the question of the adequate apparent inertia for different conditions arises. As mentioned previously, the benefits of mass reduction on even surfaces are still unclear [5, 6]. On the other hand, although the benefit of mass reduction for overcoming a hill is more intuitive, the extent to which the mass reduction affects performance has not been well characterized in the literature.

In order to shed some light on the influence of system's mass on locomotion performance under different slope conditions, we propose in this manuscript a model of wheelchair propulsion dynamics and an optimal control formulation to generate predictive simulations to estimate how the person propels the wheelchair under these different conditions. The resulting framework is used to objectively investigate the influence of model's mass under different slope inclination angles.

In Section II of this manuscript, the model of wheelchair locomotion is presented (Section II.A) and the formulated optimal control strategy introduced (Section II.B). In section III.A the approach is illustrated by the documentation of the predicted patterns for locomotion on even surface at an average speed of 0.5 m/s. In sections III, the results on the influence of mass (section III.B) and slope inclination angle (section III.C) are presented. A discussion of the results is performed in section IV and some concluding remarks presented in section V.

II. METHODS

A. Model of Wheelchair Locomotion

The manual wheelchair locomotion cycle is characterized by two phases, the contact phase and the recovery phase [4]. The contact phase corresponds to the period of the cycle in which hands are in contact with the pushrim propelling the wheelchair. The recovery phase corresponds to the period of the cycle in which hands and arms are repositioned and there is no contact between hands and rim.

The wheelchair/user model (Fig. 1) is planar and, assuming there is bilateral symmetry, it is composed of four rigid bodies: both wheels, both forearms, both upper arms, and the remaining segments along with the chair. The shoulder and elbow articulations are modeled as ideal hinge joints and it is assumed there is no slip between the wheels and the floor. The shoulder joint is assumed fixed to the wheelchair.

The model for the recovery phase has three degrees of freedom and the corresponding generalized coordinates are depicted in Fig. 1, the angle between the upper arm and the vertical β, the angle between the forearm and the vertical α, and the horizontal displacement of the shoulder x. The mass, moment of inertial and center of mass location of the segments are estimated using anthropometric data in [12] for a person of 70 kg of mass and 1.7 m height. The wheelchair dimensions and mass properties are estimated from data on conventional manual wheelchairs available in the market. All adopted values and properties are documented in Tab. I of the Appendix. The rolling resistance is modeled as a constant horizontal force of 15 N applied to the wheelchair, which is an average value for data reported for the following floor surfaces as in Fig. 7 of [1]: "tiles (cafe)", "tarpaulin (fitness)" and "low pile carpet". Moments τ_s and τ_e are applied at the shoulder and elbow joint, respectively.

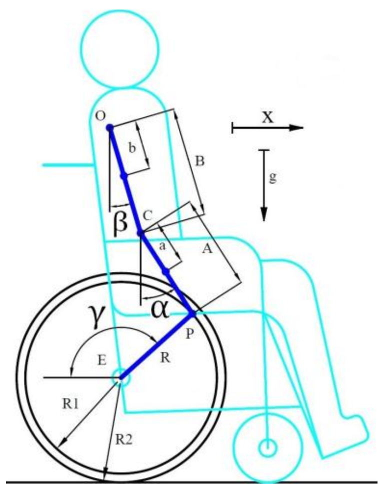

Figure 1 – Schematic representation of wheelchair/user model.

The resulting equations of motion in minimal form for the recovery phase read as

$$M(q)\ddot{q} + k(\dot{q}, q) = k_e(q, \tau_s, \tau_e), \tag{1}$$

where q is the vector of generalized coordinates as

$$q = [x \; \beta \; \alpha]^T, \tag{2}$$

M is the 3x3 mass matrix, k is the vector of generalized Coriolis and centrifugal forces, and k_e is the vector of generalized applied forces. The equations of motion in (1) were derived symbolically using Matlab's Symbolic Toolbox from the Newton-Euler equations using the d'Alembert's principle [13].

In the contact phase, hands are in contact with the pushrim as in Fig. 1, forming a hinge joint with hands and wrist lumped into a single contact point with the rim. A closed kinematic loop arises with the model turning into a four-bar mechanism composed of forearm, upper arm, wheel and wheelchair. This imposes two kinematic constraints on x, β and α as

$$c(x, \beta, \alpha) = 0, \tag{3}$$

where c is a two-dimension vector. In order to represent the dynamics in the contact phase, the reaction forces arising between the hands and the rims are added to (1) as

$$M(q)\ddot{q} + k(\dot{q}, q) = k_e(q, \tau_s, \tau_e, F_x, F_y), \tag{4}$$

where F_x represents the horizontal component and F_y the vertical component of the contact force at the interface between hands and rims. This leads to a set of differential algebraic equations (DAE's) composed of (3) and (4) which represents the dynamics of the model in the contact phase.

B. Optimal Control Formulation

The models for both phases introduced in the previous section are used to generate predictive simulations of the whole cycle of wheelchair propulsion. The resulting optimal control problem was solved by transforming it into a nonlinear programming problem using direct transcription [14] and the optimal control solver PROPT (tomdyn.com).

The optimal control problem consists of searching for the time series of the generalized coordinates q(t), the generalized velocities, the joint moments τ_s and τ_e, the contact forces in the contact phase F_x(t) and F_y(t), and the durations of the contact and recovery phases, T_c and T_r, respectively, that satisfy the constraints and minimize a cost function representing a performance criterion.

The set of constraints is composed of: *i*) the equations of motion, (1) in the recovery phase and (4) in the contact phase; *ii*) the additional contact constraints in the contact phase, (3); *iii*) the continuity constraints between phases which guarantee there is no discontinuity in position or velocity in phase transitions, considering there is no collision in the contact event of the hand with the rim, i.e. assuming relative velocity of hand and rim just before contact is zero; and *iv*) a constraint that ensures prescribed average speed. The prescribed speed selected in this study was 0.5 m/s, as a relatively slow speed which allows uphill locomotion in the investigated conditions. Also, it was imposed that initial hand contact occurs at $\gamma = 70^o$ in Fig. 1 and that the hands are released from rims at $\gamma = 120^o$ to avoid extreme arm positions.

The cost function adopted in this study as a performance criterion representing effort is the time integral of joint moments squared as

$$J = \int_0^{T_c} \left(\tau_s^2 + \tau_e^2\right) dt \; + \int_{T_c}^{T_c+T_r} \left(\tau_s^2 + \tau_e^2\right) dt \,, \tag{5}$$

based on cost functions frequently used to estimate muscle forces, refer e.g. to [15].

III. Results

A. Horizontal wheelchair locomotion

The predicted patterns for locomotion on an even surface at an average speed of 0.5 m/s are documented in this section. Fig. 2 shows stick figures illustrating the predicted patterns. Figs. 3 and 4, in turn, show the predicted time series of the three generalized coordinates for both phases, with contact phase duration of 0.575s and whole cycle duration of 1.165s.

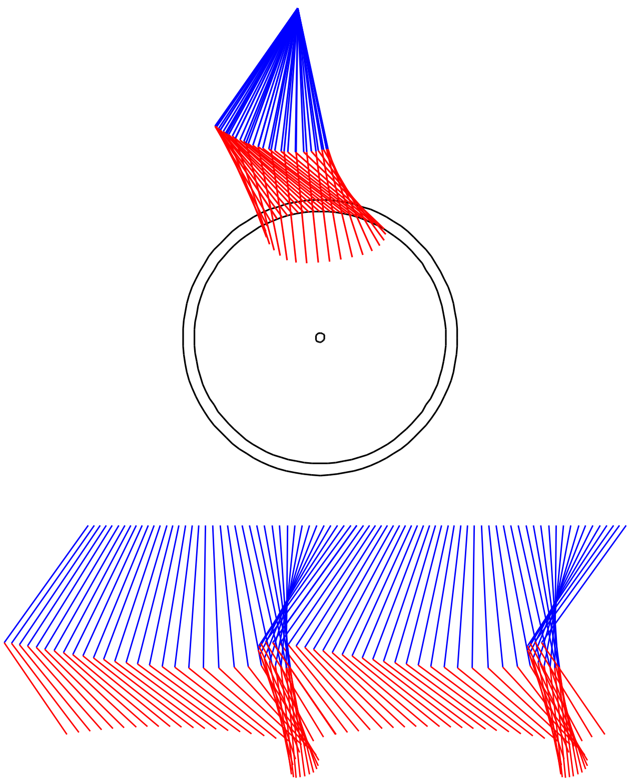

Figure 2 – Stick-figures showing predicted patterns for locomotion on an even surface at 0.5 m/s: depicted with static shoulder (top) and with moving shoulder (bottom).

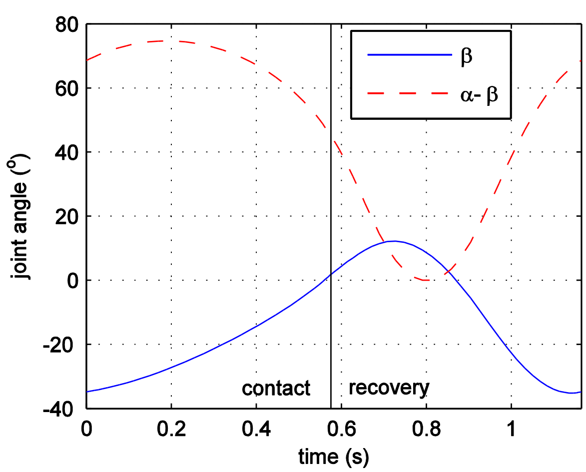

Figure 3 – Predicted joint angles for locomotion on even surface at 0.5 m/s.

Fig. 5 shows the predicted shoulder and elbow joint moments. Note that the shoulder moment is positive (anteflexion) providing propulsion with flexor muscles contracting concentrically. In the recovery phase, shoulder moment becomes negative in the first half to accelerate arm's backwards and positive in the second half to accelerate arms forwards so that contact between hands and rims occurs at zero relative velocity. The elbow moment is negative (extension) in most of the contact phase, which also means elbow extension muscles are working concentrically as elbow is extended in this phase (see Fig. 3).

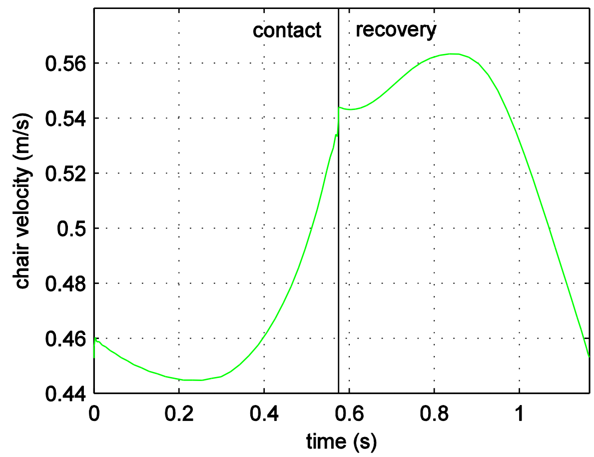

Figure 4 – Predicted chair and body speed for locomotion at average speed of 0.5 m/s on an even surface.

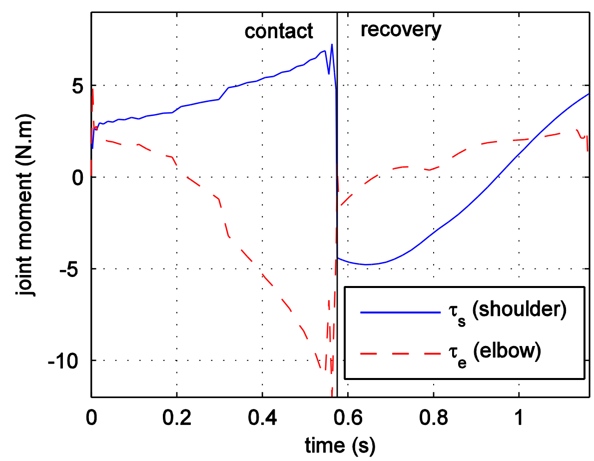

Figure 5 – Predicted joint moments for locomotion at average speed of 0.5 m/s on an even surface.

B. The influence of slope

The first investigation comprised varying slope inclination angle from 0 to 6 degrees in order to quantify the isolated effect of slope angle on propulsion effort, (5), at an average locomotion speed of 0.5 m/s. A strong dependency on slope angle can be observed with steep increases in the performance criterion values (effort) as slope angle becomes greater, Fig. 6.

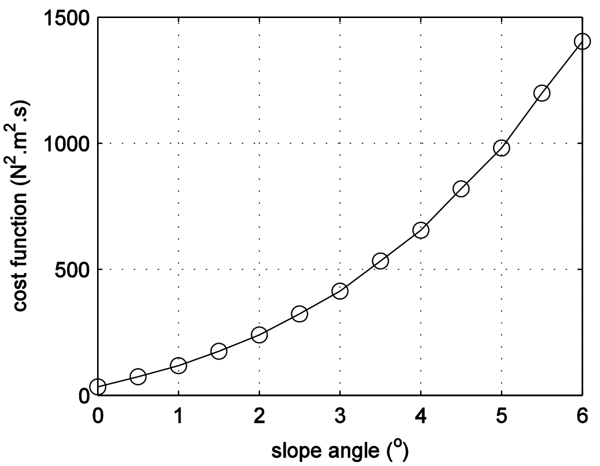

Figure 6 – Optimal cost function value for different slope inclination angles.

C. *The influence of mass*

The next investigation comprises the study of the combined effects of varying mass and slope on locomotion performance. Fig. 7 shows optimal values for the cost function, (5), for 0-, 1- and 2-degree inclination angles and whole system's mass varying from 25 % to 175 % of the original system mass reported in Table I. Note that the mass variation were undertaken on the mass of the rigid body comprising the chair (without wheels) and the body without arms. As this rigid body contains most of the system's mass, from now on the mass variation will be referred to as system's mass variation. It can be observed that the mass influence on slopes is critical, with effort increasing substantially with mass increases, but the influence of mass increases on even surfaces is negligible.

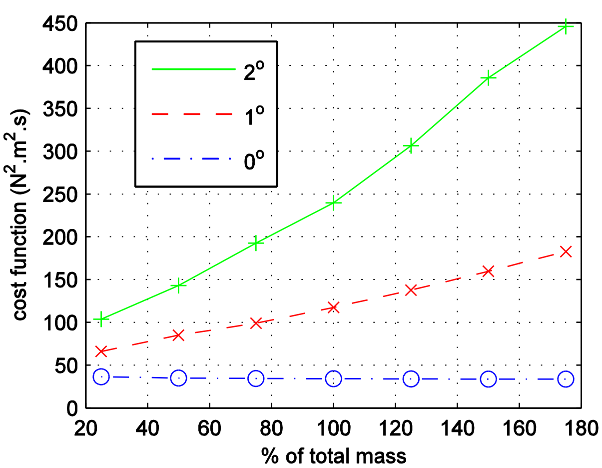

Figure 7 – Optimal cost function value for different system's mass values at three slope inclination angles.

IV. Discussion

Kinematic patterns, Figs. 2-4, are consistent with data in the literature, although the low simulation speed 0.5 m/s adopted in this study to allow for uphill locomotion at severe inclination angles, hinders direct comparison to normative data in the literature, usually reported for locomotion at larger average speeds. Fig. 2 reveals a semi-circular stroke pattern [2], which is observed for all simulations performed at even surfaces. It is interesting to note that the optimal pattern changes to a "single looping over propulsion" (SLOP) pattern for all the simulations performed at slopes.

Fig. 3 shows shoulder anteflexion and predominant elbow extension (α-β) during the contact phase, while Fig. 5 shows joint moments in the same directions in this phase, i.e. anteflexion for shoulder and extension for elbow. This means contraction is predominantly concentric in the contact phase and mechanical power is positive with increase in system's mechanical energy during this phase as expected. Fig. 4 shows the resulting chair (shoulder) speed, averaging 0.5 m/s over the whole locomotion cycle. It can be observed that the chair is accelerated in the second half of the contact phase with large applied moments and positive mechanical power by the elbow and shoulder joints. On the other hand, shoulder is decelerated in the recovery phase where power input ceases and energy is dissipated by the rolling resistance force.

The results on effort, (5), as a function of slope inclination angle, Fig. 6, reveal a strong dependency. The maximal inclination angle, six degrees, corresponds to approximately the maximum inclination recommended in standard regulations, often allowed over short distances only (refer e.g. to the *Americans with Disabilities Act Accessibility Guidelines* - ADAAG).

The results observed in Fig. 7 for locomotion on even surfaces suggest that there is almost no influence of system's mass on the cost function, i.e., the propulsion effort adopted in this study as the locomotion performance criterion. This is in conformity with the experimental results observed in other studies [5, 6] and can be explained by the fact that in locomotion on even surfaces the only net energy that needs to be supplied over a complete locomotion cycle to sustain steady-state is the energy dissipated by friction, in this case the rolling resistance. Although there is a dependency between weight on the wheels and rolling resistance [1], this dependency was neglected in this study and average values in [1] were adopted. Future studies should include a model of the effect of weight on rolling resistance to confirm the small influence of system's mass on steady-state locomotion performance on even surfaces.

The scenario is quite different when inclined surfaces are considered, even at relatively small angles as shown in Fig. 7. The influence of system's mass on propulsion effort on ramps is clearly critical. In these conditions, optimal cost function value or effort increases roughly linearly with system's mass, within the mass variation limits analyzed, and becomes more important as slope angle increases. The simulations also reveal an increase in the frequency of stroke with increasing mass for all conditions (plots not included), which might be an important factor for wheelchair users and will be the focus of further investigation. These results show the negative effects of overweight on performance on overcoming ramps and obstacles. More importantly, the results evidence the benefits of decreasing wheelchair mass, either through wheelchair design optimization or through the apparent reduction of system's mass through the implementation of an impedance control scheme in a power-assisted manual wheelchair. The benefits of mass reduction would probably hold true in transients, for instance, when subjects accelerate from a resting position, during maneuvers or in overcoming obstacles.

The purpose of the impedance control differs from the one of standard control systems in the sense that its main objective is to make the system mimic a reference model rather than ensuring an input reference tracking. The main objective in this strategy is to control the relationship of the force and the velocity. In some cases this relationship is adopted in the form of a linear system as a mass-spring-damper system [16]. One way to implement the impedance control is by controlling the velocity and using the desired impedance to generate the velocity reference by measuring forces. This strategy requires that both the kinematic variables as well as the forces involved are available. This may be achieved, for example, by using a force and torque sensor mounted on the axis as in [10]. It is expected that the reduction of the apparent mass achieved by the implementation of an impedance control strategy is more effective and natural for a wheelchair user compared with simpler strategies such as that of amplifying the external force applied by the user.

Although some of the predicted patterns are generally consistent with data in the literature, the difference in speed locomotion in experimental data reported in the literature compromises the comparison. For this reason, a validation of the model including the choice of cost function is still required to reduce the speculative nature of the conclusions drawn in this study. Model improvements are proposed as future work, some of which are currently under investigation: incorporating muscle models and contraction dynamics, allowing collision of the hands with the rims, allowing for optimization of propulsion contact and release angles of the hands on the rims, allowing for shoulder joint displacement with respect to the chair and investigating the influence of different cost functions and rolling resistance models on predicted patterns.

V. Conclusion

Wheelchair locomotion with conventional manual wheelchairs requires large energy consumption and may cause shoulder pain or even injuries. Reducing weight is one option for improvement regarding quality of locomotion. To better understand the influence of system's mass on locomotion performance under different slope conditions, we proposed an optimal control formulation to predict manual wheelchair locomotion patterns.

The results indicate system's mass has negligible influence on locomotion performance on even surfaces in steady-state conditions, in line with experimental results reported in the literature. On the other hand, the study shows a strong dependency on system's mass on slopes, even at relatively modest inclinations. This indicates the importance of mass reduction to improve locomotion quality. This can be achieved through structural improvements to the wheelchairs, with limited effectiveness as the mass of modern wheelchairs is already small compared to subject's mass. As an alternative, we propose the implementation of an impedance control scheme to reduce system's apparent mass or inertial through proper actuation in a power-assisted manual wheelchair.

Appendix

TABLE I. ANTHROPOMETRIC AND WHEELCHAIR DATA.

Person Height	1.70 m
Upper arm length (B in Fig. 1)	0.3196 m
Upper arm CM location (b in Fig. 1)	0.1393 m
Forearm length (A in Fig. 1)	0.2465 m
Forearm CM location (a in Fig. 1)	0.1060 m
Handrim radius (R1 in Fig. 1)	0.2794 m
Rear wheel radius (R2 in Fig. 1)	0.3048 m
Moment of inertia of rear wheel	0.1395 kg.m²
Upper arm moment of inertia	0.021 kg.m²
Forearm moment of inertia	0.020 kg.m²
Person mass	70.0 kg
Forearm mass (with hand)	1.54 kg
Upper arm mass	1.96 kg
Rear wheel mass	3.00 kg
Complete wheelchair mass	12.0 kg
Gravity acceleration	9.81 m/s²
Shoulder to axle distance - horizontal	0.05 m
Shoulder to axle distance - vertical	0.73 m
Total rear wheels rolling resistance	15 N

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