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Torque Sensor Free Power Assisted Wheelchair

Daniel Petersson, Jonas Johansson, Ulf Holmberg and Björn Åstrand

Abstract—A power assisted wheelchair combines human power, which is delivered by the arms through the pushrims, with electrical motors, which are powered by a battery. Today's electric power assisted wheelchairs use force sensors to measure the torque exerted on the pushrims by the user. This leads to rather expensive and clumsy constructions. A new design, which only relies on velocity feedback, thus avoiding the use of expensive force sensors in the pushrims, is proposed in this paper. The control design is based on a simple PD-structure with only two design parameters easily tuned to fit a certain user; one parameter is used to adjust the amplification of the user's force and the other one is used to change the lasting time of the propulsion influence. Since the new assisting control system only relies on the velocity, the torque sensor free power assisted wheelchair will besides giving the user assisting power also give an assistant, which pushes the wheelchair, additional power. This is a big advantage compared to the pushrim activated one, where this benefit for the assistant is not possible.

I. INTRODUCTION

For many years, there have only been three viable options for functionally impaired people, who are unable to propel an ordinary wheelchair. The three options are; using an electric powered wheelchair, driving a scooter, or being pushed by an assistant. But recently, a new type of wheelchair has been developed that combines human power and electrical power. This new type is called power assisted wheelchair.

A power assisted wheelchair [1] combines human power, which is delivered by the arms through the pushrims, with electrical motors, which are powered by a battery. The power assisted wheelchair is aimed at customers, who have used a regular wheelchair for a long time, but who have become weaker or just need additional power when driving uphill. This kind of wheelchair will provide additional power for users, which will spare their wrists, elbows and shoulders.

Today's electric power assisted wheelchairs use force sensors to measure the torque exerted on the pushrims by the user [1-5]. This torque is then increased with the help of electrical motors. The wheelchair's speed is also measured, with rotational encoders, to calculate how much power and for how long the power should be applied. The force sensors in the pushrims are rather expensive and this approach also makes the wheels a little bit clumsy.

The objective with this paper is to present a new, better and cheaper solution that does not use expensive torque sensors in the pushrims. The idea is inspired by [6] where a torque sensor free solution is suggested. In this paper, it is demonstrated that the approach in [6] might not work in practice due to gearbox play, model sensitivity and rather

difficult design parameters. A solution to overcome the problem is first suggested. Then, a PD-controller parameterization is proposed with only two easily tuned design parameters. It is also illustrated how to rebuild a joystick controlled wheelchair into a torque sensor free prototype.

II. DESIGN APPROACHES

Yoichi Hori and Schoon Oh [6] have presented the so-called "sensor free" solution, which is based on velocity control. The authors have designed a controller that is sensitive to disturbances, which can be used in power assisted systems.

Usually a power assisted system uses sensors, in most cases torque sensors, to measure the external forces caused by for example human influence. These systems that use force sensors are expensive and therefore a cheaper solution is desirable. In the paper, Hori and Oh have presented a power assisting controller that does not use these expensive force sensors. Their controller only relies on the external changes in the feedback signal. In other words, the controller interprets and amplifies the external disturbances, compared to an ordinary control system that suppresses disturbances.

In the case with the pushrim activated power assisted wheelchair, the expensive torque sensors in the pushrims are used as a feedforward signal to the control system. With a velocity disturbance observed controller, these torque sensors can theoretically be removed. The wheelchair's new control system will only rely on the wheel's velocity values, when controlling the electrical motors.

A. Hori & Oh Design

Hori and Oh propose a continuous-time controller design of the velocity disturbance controlled system in [6]. The proposed system is discretized to make it implementable. The discrete-time form of the control design is shown in Fig. 1. The design consists of five blocks. The first block *B1* is the motorized wheelchair process that is to be controlled. The wheelchair model (process) presented is

$$P(q^{-1}) = \frac{B(q^{-1})}{A(q^{-1})} = \frac{b_1 q^{-1}}{1 + a_1 q^{-1}}$$

where q^{-1} is the backward-shift operator.

The input signal to the process is the control signal together with the human's force (external disturbance) and the output signal from the process is the wheelchair's real velocity. The *B2* block removes the wheelchair's dynamics and outputs the process' input signal, which theoretically is

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Fig. 1. Discrete-time version of Hori & Oh design

the control signal and the external disturbance, delayed one sample.

$$\frac{A}{b_1} \cdot v_r = \frac{A}{b_1} \cdot \frac{b_1 q^{-1}}{A} (u + d) = q^{-1}(u + d)$$

After subtracting the $B2$'s output signal with the control signal, which is delayed one sample by the block $B3$, only the disturbance signal remains.

$$q^{-1}(u + d) - q^{-1}u = q^{-1}d$$

The obtained disturbance signal, then passes block $B4$ that contains the dynamics of a desired model of the wheelchair, which outputs a desired velocity.

$$v_m = \frac{b_m}{A_m(q^{-1})} q^{-1}d = \frac{b_m q^{-1}}{1 + a_m q^{-1}} d$$

The difference between the desired velocity and the real velocity will then become the new control signal and its effect can be altered by changing the constant in block $B5$.

The main idea of the design by Hori and Oh, is that it will send a positive feedback signal to the real wheelchair, which regulates the speed so the desired velocity is equal to the real velocity. In other words, the dynamics of the power assisted wheelchair can be modified, by just changing the parameters in the desired wheelchair model block. However, the controller will only work correctly if the model of the wheelchair is accurate and remains the same during the drive.

The design by Hori and Oh can be reduced to one controller block, shown in Fig. 2. This structure will be shown below to be similar as a proportional derivative (PD) controller with positive feedback. Although the structure is the same as for a PD-controller, the parameterization is more complex.

B. PD-Control Design

Consider a continuous-time PD-controller

$$u = K_p y + K_d \dot{y}_d$$

where $y_d = \frac{d}{dt}y$. But neither is it possible to implement a pure differentiation, nor is it desirable due to sensitivity to

Fig. 2. Power assist with Hori & Oh controller

Fig. 3. Power assist with PD-controller

high frequency noise. Instead, the derivative action is filtered as

$$\tau \frac{d}{dt} y_d(t) + y_d(t) = \frac{d}{dt} y(t)$$

where $\tau = 0$ would correspond to a pure differentiation ($y_d = \frac{d}{dt}y$) and a larger value on the time-constant τ limits the high-frequency gain. In discrete-time, using backward difference approximation of the derivative

$$\frac{d}{dt} \rightarrow \frac{1 - q^{-1}}{h}$$

where h is the sampling interval, this results in

$$\begin{aligned} [\tau \frac{1-q^{-1}}{h} + 1] y_d &= \frac{1-q^{-1}}{h} y \\ \Rightarrow y_d &= \frac{c_1(1-q^{-1})}{1-c_2 q^{-1}} y \end{aligned}$$

where $c_1 = \frac{1}{\tau+h}$, $c_2 = \frac{\tau}{\tau+h}$.

The PD-controller with the high frequency gain limitation filter will then be

$$\begin{aligned} u(k) &= K_p y(k) + K_d y_d(k) \\ &= K_p y(k) + K_d \frac{c_1(1-q^{-1})}{1-c_2 q^{-1}} y(k) \\ &= \frac{K_p(1-c_2 q^{-1}) + K_d c_1(1-q^{-1})}{1-c_2 q^{-1}} y(k) \\ &= \frac{(K_p + K_d c_1) - (K_d c_1 + K_p c_2)q^{-1}}{1-c_2 q^{-1}} y(k) \end{aligned}$$

The control system with the PD-controller including the limitation filter is shown in Fig. 3. In both Hori & Oh and PD designs the feedback controller is of the structure

$$u(k) = \frac{S(q^{-1})}{R(q^{-1})} y(k) = \frac{s_0 + s_1 q^{-1}}{1 + r_1 q^{-1}} y(k)$$

but with different design parameters.

Although the structure is the same between Hori and Oh design and PD design there is a significant difference

Fig. 4. Block scheme of the original joystick controlled wheelchair's control hardware

in parameterization. In Hori and Oh design the controller parameters are

$$\begin{aligned} s_0 &= K(b_m/b_1 - 1) \\ s_1 &= K(a_1 b_m/b_1 - a_m) \\ r_1 &= a_m + K b_m \end{aligned}$$

while for PD design it is

$$\begin{aligned} s_0 &= K_p + K_d c_1 \\ s_1 &= -(K_d c_1 + K_p c_2) \\ r_1 &= -c_2 \end{aligned}$$

Notice that the controller pole $-r_1$ for PD design is fixed at c_2 , only depending on the noise reduction time constant τ , while the controller pole for Hori and Oh design depends on all design parameters a_m , b_m and K . Therefore, the PD parameterization is likely to have an advantage for tuning of sensitivity to noise and high frequency unmodeled dynamics.

The Hori and Oh design parameters are nonlinear functions of the controller parameters s_0 , s_1 and r_1 . Solving for a_m results in the equation

$$a_m^2 + (s_0 b_1 - r_1 - a_1) a_m + (a_1 r_1 - b_1 s_1) = 0$$

which has two solutions. The parameters K and b_m can then be solved from

$$\begin{aligned} K &= (r_1/b_1 - s_0) - a_m/b_1 \\ b_m &= (r_1 - a_m)/K \end{aligned}$$

III. WHEELCHAIR RECONSTRUCTION

An ordinary joystick controlled motorized wheelchair was used to build the new power assisted wheelchair, which only will rely on velocity feedback.

A. Joystick Controlled Wheelchair

A block scheme of the original joystick controlled wheelchair's control hardware is shown in Fig. 4. The user, driving the motorized wheelchair, controls the wheelchair with a joystick module. The joystick module sends the user's commands to a power module via a serial communications protocol. The power module consists of two cards; a main power card that holds all power electronic devices and an

Fig. 5. Block scheme showing modifications of the original design.

external processor card, which handles the communication between the joystick module and the main power card. The power module, which is supplied by a 24 volts battery, translates the received commands from the joystick module into control signals (power signals) that drive the two electrical motors on the wheelchair. Each motor is connected to a gearbox, with a resulting gear ratio of 81:1. Each gearbox is in turn connected to a wheel with a pushrim, which also can be used to propel the wheelchair. Each wheel is exposed to different loads, e.g. weight, friction or up- and down-hill driving, that affects the motor's torque. On each motor there is also a solenoid brake mounted, which is used to brake the motor axis when the power module gets a braking command from the joystick module.

B. Feedback Controlled Wheelchair

A block scheme of the new power assisted wheelchair's control hardware is illustrated in Fig. 5. All significant modifications of the control hardware are highlighted in grey. The joystick module has now been removed and replaced by a programmable controller card that will be used to develop the new control system. The controller card is "directly connected" to the power module's hardware instead of using the serial communication. The external processor card, which was placed inside the power module and handled the serial communication protocol translation, has been replaced with an external middle card that translates control signals from the controller card before the signals enter the main power card. The external middle card also works as a safety card that protects the power module from being destroyed by, for example, careless coding of the control system. The solenoid brakes, placed on each electrical motor, were removed to be able to mount rotary encoders instead. The speed values from the rotary encoders will be used as feedback signals to the control system.

IV. EXPERIMENTAL RESULTS

A. Modeling

1) *Step response*: The wheelchair model was obtained by investigating the wheelchair's velocity response when

Fig. 6. The wheelchairs velocity step response. The large spikes are due to gearbox play. (1 m/s = 5370 pulses/s)

applying a step signal to the motors. The wheels velocity, which is measured with the rotational encoders, is recorded in MATLAB by using the control card's wireless serial communication. The motors' speed is altered by changing the duty cycle of the square wave signals, from the control card, that are fed to the pulse width modulated motor controller. The duty cycle on the control card has a 10-bit resolution (0-1023), where the duty cycle 511 results in a non rotating motor and the values 1023 and 0 results in a maximum rotation in forward and backward direction respectively. The duty cycle will now be defined as a function of the control u .

$$duty\_cycle = 511 + u, 0 < duty\_cycle < 1023$$

The step responses of the power assisted wheelchair when applying two step changes, in the control signal u , from 0 to 250 and back to 0, is shown in Fig. 6. The large spikes are caused by play in the wheelchair's gearboxes. The play causes the motor axis, on which the rotational encoder is mounted, to rotate freely, a couple of revolutions before the wheel starts to rotate. Since the torque sensor free control systems only relies on the velocity values from the encoders these spikes in the velocity have to be reduced otherwise the control system will make the wheelchair feeling very jerky. This problem has to be solved in software, since the play in the gearbox cannot be reduced. Therefore a play passing method that does not measure the velocity when the play problem occurs was developed.

2) *Play Passing Method*: The main idea of the method is that before the control system starts to measure the velocity it will apply a small force that is enough to rotate the motor axis past the play but without start rotating the wheels. In other words, the motor axis together with the rotational encoder will not rotate freely on the start, which will reduce the large spikes in the beginning of the velocity response.

For the play passing method to work in both forward and backward direction it has a direction sensing function. This function tells the motors to rotate past the play in either forward or backward direction. Therefore, when the wheelchair user wants to change direction, this play reduction method

Fig. 7. The state graph of the power assisted system with the play passing method

Fig. 8. The filter that slows down the step signal, which is used to pass the gearbox play, and the jump back threshold

will first close the power assisting control system, and then change direction of the play passing force and finally start the power assisting control system again.

As a result of this, the control system for the power assisted wheelchair will have two modes; the controller mode and the play passing mode, which the system jumps between. The state graph with the two modes is shown in Fig. 7. When the system is in the *controller mode*, the regular power assisting controller runs, which gives the user additional power. If the user changes direction, which results in a changed sign of the wheelchair's velocity $y(k)$, the system jumps to the *play passing mode*. In the *play passing mode* a small motor force is applied, which will rotate the motor axis together with the rotational encoder past the play, by changing the control signal $u(k)$ to $d_0 \cdot \text{sign}(y(k))$. To prevent this direction change from feeling jerky, the step signal $d_0 \cdot \text{sign}(y(k))$ is run through a filter, which is shown in Fig. 8. The response of the filter is altered by changing a_f , which should be chosen $-1 < a_f < 0$. This filter slows down, or smoothes, the direction change. When the control signal $u(k)$ reaches a certain threshold value $c_{th} \cdot d_0$, the system will jump back to the controller mode. The threshold value can be adjusted with the threshold constant c_{th} , which should be chosen $0 < c_{th} < 1$. When the system jumps back to the controller mode, the wheelchair's power assisting controller is enabled again, and the user can drive in the other direction without the play in the gearbox disturbing the control system.

The play reduction method works individually on each motor, which makes it possible for the user of the power assisted wheelchair to rotate the wheels in different direction, thus making a tight turn with the wheelchair. The only disadvantage with this play passing method is that it will cause a small delay, depending on a_f in the filter, before giving the assisting power when changing direction.

The step response with play passing method is shown in

Fig. 9. The wheelchairs velocity step response with the play passing method. The spikes are reduced (compare Fig. 6).

Fig. 10. The wheelchair's velocity step response with different user weights

Fig. 9. This play reduction method works well. The method reduces the spikes in the velocity response, especially in the increasing phase. However, the method will give a larger gain compared to the original step response with the same control signal change of 250. This larger gain is due to the extra play passing force with $d_0 = 45$, which actually gives a control signal change from 45 to 295 and back to 45.

This play passing method, with the parameter settings $d_0 = 45$, $a_f = -0.22$ and $c_{th} = 0.98$, will be used in all experiments with step responses and control systems.

3) *User Weight Influence:* The step response for different user weights is shown in Fig. 10. This experiment shows that the final velocity is almost the same for all different user weights. But, the time response will become slightly slower for increasing weight.

Thus, the wheelchair model will not be significantly affected by small changes in the user's weight. This is important since the control system should not require retuning when the user changes weight.

The model with the user weight 82 kg will be used when simulating and implementing the control systems. This response gives the wheelchair model parameter values $a_1 = -0.949$ and $b_1 = 1.29$ when using the sampling period $h = 0.05$ s.

Fig. 11. The real and the simulated disturbance response with the controller by Hori and Oh ($a_m = -0.99$, $b_m = 1.8$, $K = 1$) compared to the response without any controller

B. Hori & Oh

The main idea of the Hori and Oh controller is that it obtains the external disturbance caused by the user, by using a process eliminating function and then feeds the disturbance to a desired process model of the wheelchair. This control design, shown in Figs. 1-2, contains three control parameters; a_m and b_m , which are used in the desired wheelchair model, and the constant K , which is used to regulate the power assisting controller's influence. The three control parameters a_m , b_m and K should be chosen within some limitations to give the desired effect. The desired gain factor b_m should be chosen greater than the real wheelchair's gain factor b_1 to give a higher velocity on each push. The desired a_m should be closer to -1 than a_1 to achieve a longer lasting response of each push. The constant K should be greater than zero.

This control design works fine when simulated. The response to a simplified user propulsion disturbance signal, which is a square pulse with a top value of 200 and a length of one second, is shown in Fig. 11. With the chosen parameters $a_m = -0.99$, $b_m = 1.8$ and $K = 1$, the simulated controller gives a larger maximum velocity and a longer lasting response compared to the same wheelchair without the assisting controller.

The control design by Hori and Oh is implemented in the controller card on the power assisted wheelchair and the same disturbance pulse signal that was used in the simulation is fed to the wheelchair's control system. The wheelchair's disturbance response is measured and the measured data is compared to the simulated data in Fig. 11. But, the power assisted wheelchair's velocity does not match the simulated response at all. The first order wheelchair model that they have proposed does not take play problem and other real life factors into account and therefore differs too much from the real wheelchair process, which results in the controller not function properly on the wheelchair, despite the play passing method.

Fig. 12. The real and the simulated disturbance response, with PD-controller ($K_p = 0.030$, $K_d = 0.010$, $\tau = 0.5$) compared to the response without controller

C. PD-Controller

The PD-control design has a simpler parameterization with two gain parameters; the proportional gain K_p , which is used to change the lasting time of the propulsion influence and the derivative gain K_d that can be used to alter the amplification of the user's force. The derivative part in the PD-controller is implemented as a high pass filter with a limited noise feedback, which reduces high frequency noise in velocity.

The simulated and the real response for the gain constants $K_p = 0.030$ and $K_d = 0.010$ and with the frequency limiter parameter $\tau = 0.5$ when the control system is fed with the disturbance pulse signal is shown in Fig. 12. The measured velocity data matches the simulated disturbance response well. The difference after five seconds has to do with a slightly uneven floor, which makes the wheelchair turn a little and vary in velocity. The two spikes, one in the beginning and one on the top, are caused by the play problem.

The PD-controller, implemented on the power assisted wheelchair, works well. The controller amplifies the user's propulsion and generates a longer lasting propulsion influence. If K_p is increased, the amplification of the initial force is increased, but that will also make the influence of the disturbance to last longer. However, K_d in the PD-controller can be used instead of K_p to increase the initial force without changing the lasting time so much. The influence of the proportional controller with the gain value $K_p = 0.030$ and the two different K_d values 0.05 and 0.015 is shown in Fig. 13. As shown in the figure, when increasing K_d , the amplification of the initial force is increased.

D. Design Comparison

Since the controller structure for both designs are the same, it is interesting to compare what parameter choice for Hori and Oh design would correspond to PD design. In the PD design example above, $K_p = 0.03$, $K_d = 0.01$ and $\tau = 0.5$, which gives $s_0 = 0.048$, $s_1 = -0.055$ and $r_1 = -0.91$. This corresponds to two different parameter

Fig. 13. The real and the simulated disturbance response, with the PD-controller values $K_p = 0.030$, $K_d = 0.015$, $\tau = 0.5$ (the upper curves) compared to the response with the values $K_p = 0.030$, $K_d = 0.005$, $\tau = 0.5$

choices for Hori and Oh design:

$$\begin{array}{ll} a_m = -0.98 & a_m = -0.94 \\ b_m = 8.85 & b_m = -1.15 \\ K = 0.0082 & K = -0.026 \end{array}$$

Negative values for b_m and K are excluded by the design guidelines by Hori and Oh. The case with $a_m = -0.98$ is a reasonable choice considering that $a_1 = -0.949$. However, following the guidelines, the gain b_m should be chosen slightly larger than $b_1 = 1.29$. But the extremely large $b_m = 8.85$ and small $K = 0.0082$ would hardly be natural choices.

V. CONCLUSION

This paper presents a new kind of power assisting wheelchair, shown in Fig. 14, that does not use torque sensors in the pushrims. This torque sensor free power assisted wheelchair only relies on one feedback signal. This feedback signal is the wheelchair's velocity, which is measured with rotational encoders. The proposed design is a PD-controller with positive feedback. The simple parameterization makes the controller easy to tune. The assisting power of the wheelchair can be adjusted in both how much the user's force should be amplified and how long time the influence of the user's propulsion should last. These adjustable parameters make it possible to configure the power assistance to suit all users, from those who just need a little additional assistance to those who need much assisting power.

The torque sensor free power assisted wheelchair will behave as an ordinary wheelchair, but the user propelling the torque sensor free wheelchair will feel like his strength is increased sharply. As a result of this, it will be easier to propel the wheelchair in uphill direction, since the control system amplifies the user's propelling movements. It will also become easier to brake the wheelchair in downhill direction since the control system amplifies the braking force applied by the user.

This torque sensor free power assisted wheelchair has a big advantages over the pushrim activated one. For example,

Fig. 14. The new power assisted wheelchair

if an assistant wants to push a functionally impaired user sitting in a power assisted wheelchair with torque sensors in the pushrims, the assistant will not benefit from the assisting control system. But, in the case with the disturbance observed controller system, which relies on the wheelchairs velocity, the wheelchair will also give additional power for the assistant.

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