Simulation Model of a Lever-Propelled Wheelchair

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Abstract— **Wheelchair efficiency depends significantly on the individual adjustment of the wheelchair propulsion interface. Wheelchair prescription involves reconfiguring the wheelchair to optimize it for specific user characteristics. Wheelchair tuning procedure is a complicated task that is performed usually by experienced rehabilitation engineers. In this study, we report initial results from the development of a musculoskeletal model of the wheelchair lever propulsion. Such a model could be used for the development of new advanced wheelchair approaches that allow wheelchair designers and practitioners to explore virtually, on a computer, the effects of the intended settings of the lever-propulsion interface. To investigate the lever-propulsion process, we carried out wheelchair lever propulsion experiments where joint angle, lever angle and three-directional forces and moments applied to the lever were recorded during the execution of defined propulsion motions. Kinematic and dynamic features of lever propulsion motions were extracted from the recorded data to be used for the model development. Five healthy male adults took part in these initial experiments. The analysis of the collected kinematic and dynamic motion parameters showed that lever propulsion is realized by a cyclical three-dimensional motion of upper extremities and that joint torque for propulsion is maintained within a certain range. The synthesized propulsion model was verified by computer simulation where the measured lever-angles were compared with the angles generated by the developed model simulation. Joint torque amplitudes were used to impose the torque limitation to the model joints. The results evidenced that the developed model can simulate successfully basic lever propulsion tasks such as pushing and pulling the lever.**

I. INTRODUCTION

Wheelchairs are commonly used mobility devices. However, more than 50% of all manual wheelchair users have experienced upper limb pain or injury. Results of recent studies show that wheelchair users who push on the handrim an average of 2000–3000 times a day often develop injuries in their shoulders, wrists, and hands, including carpal tunnel syndrome [1-3]. As a response to the problem, numerous theoretical and experimental researches for development of technical solutions minimizing the effects the wheelchair propulsion were initiated.

Lever-propelled wheelchairs, where user's force is transmitted to the driving wheel via a lever and one-way clutch mechanism, seem a promising alternative to the standard hand-rim wheelchairs. Such wheelchairs require less driving efforts because the user applies force to the end of a lever which length is usually greater than the handrim radius. It was demonstrated with many studies that lever propulsion wheelchair lower the risk of secondary disorders [4, 5]. Lever-propulsion mechanisms can be customized easily. The lever length, grip orientation and handle's shape, as well as the gear ratio can be selected individually to match the upper limb abilities of the particular patient. However, the adaptation of lever-propelled wheelchairs comprises individual selections of many parameters and wheelchair adjustment is usually done by experienced rehabilitation engineers. As an alternative to the standard adjustment practice of lever-propelled wheelchair, in this paper we focus on a wheelchair tuning approach where the individual optimization parameters are defined via computer simulation of the interaction between the wheelchair mechanism and the individual. It is expected that such approach will increase the tuning accuracy, will improve patients' experience, and will reduce the adjustment time. The development of precise models of the human motions during wheelchair propulsion could enable accurate wheelchair adaptation by computer simulation and virtual "modification" of the wheelchair on the computer. As a first step toward the development of the idea, in this paper we present some initial results from the design of a model of human- lever propulsion.

II. ANALYSIS OF WHEELCHAIR LEVER PROPULSION

A. Measurement and analysis method

For the design of the model of human- lever propulsion we need to have enough clarity on the kinematic and dynamic features of lever propulsion. In order to explore relations and conditions that are needed for the model development, we conducted propulsion experiments with a group of subjects who were asked to perform wheelchair lever propulsion motions. The outputs for selected kinematic parameters were recorded during the exercises. The data were used for calculations of the kinematic and dynamic motion parameters that were used for the designed model of wheelchair lever propulsion.

In the conducted propulsion experiments we used a special wheelchair simulator (Figure 1). It allows adjustment of the wheelbase length, seat height, and simulates different driving resistance of the wheels. Five male adults without disabilities (age 23.0 ± 1.4 years old, height 171.0 ± 6.7 cm, body weight 57.6 ± 4.3 kg, mean ± SD) participated in the experiments. For the experiments, the seat position of the wheelchair simulator was adjusted individually. For that purpose, the subject was asked to keep his arm straight downwards and the seat was moved until his middle finger touches the wheel axle [6].

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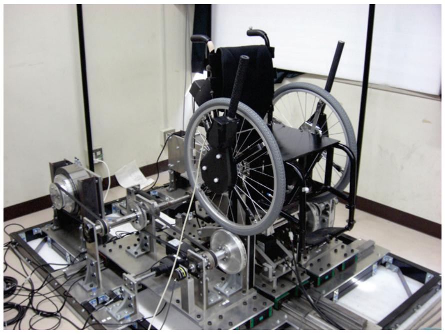

Figure 1. Wheelchair simulator used for lever propulsion experiments.

Figure 2. Musculoskeletal model for wheelchair lever propulsion.

Next, the subject was asked to keep his upper arm vertical while bending his elbow at 90^{o} angle. The wheelchair lever was set in vertical direction and the lever-propulsion module of the simulator was moved until the subject was able to grasp the lever. For the test group, the average distance from the center of the wheel axis position to the lever handle was 37.5 ± 2.3 cm. Participants were asked to propel the "wheelchair" by pushing and pulling the lever forward and backward sequentially. The measured wheelchair "speed" was shown to the subject on a special display placed in front of the wheelchair simulator. Participants were asked to keep target speed 2 rad/s. For the experiments, the driving wheel resistance of the simulator was set to correspond to movement on a 6° slope. During the lever propulsion tests, the body motions were measured by using a motion capture system OptiTrack (Natural Point Inc.) with 12 infrared cameras. Infrared reflective markers were attached into 31 feature points on the body. For the tests we used e-arm B02 lever system (Tohotechnos Co. Ltd.). A six-axis force sensor (IFS-67M25T50-140-ANA; Nitta Corp.) was incorporated with the lever to measure the three-directional forces and moments applied to the lever. The signals from the force sensor were recorded. The sampling frequency of the force measurement was set to 100 Hz.

Next, the joint angles, joint torque, and muscular tension, were calculated from the measured human motion parameters and driving force via inverse kinematic and dynamics analysis. The musculoskeletal model of the wheelchair lever propulsion was developed by using 3D CAD software (SolidWorks Corp.) and musculoskeletal modeling software (SIMM, Nac Image Technology Inc.) (Figure 2). In the model, each arm was presented as a mechanism with 7 degrees of freedom.

B. *Analysis of the Results*

Results showed that during the propulsion exercise the joint angles of both arms and the lever angles changed on a cyclic way (Figure 3). The average driving cycle for all five subjects was about 0.9 s. The driving phase occupied 47.4% from the driving cycle while the recovery phase was 52.5%. During the execution of the movement task, the lever movement remained mainly in the 2D plane when the arm performed 3D motion by involving all 7 degrees of freedom. These results confirmed that the simulation model should be constructed as a 3D model.

The ranges of joint torque during the lever propulsion exercise are presented in Table 1. The experiments for propulsion on a 6^{o} slope showed that the subjects applied much physical efforts to maintain the same speed. For the simulation of the lever propulsion the range of joint torque was constrained. Thus, we reduced the number of the search parameters for motion generation and decreased the calculation time. In addition to that, the imposed torque constrain suppresses the generation of eventual unnatural and unfeasible driving motion signals of the model.

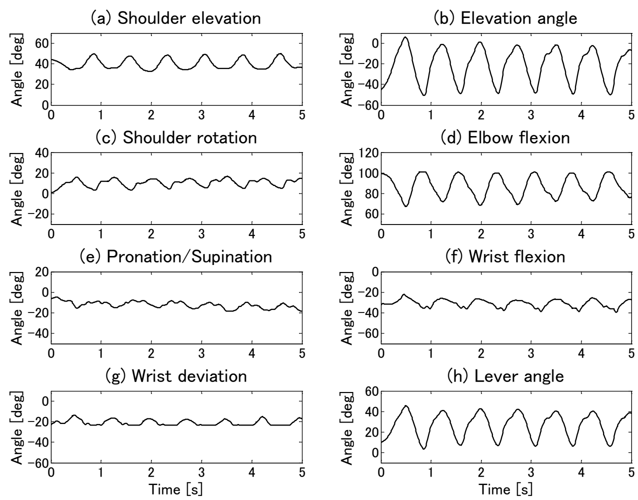

Figure 3. Kinematic features of the human motions during wheelchair lever propulsion.

Table 1 Range of joint torque.

Joint	Joint torque [Nm]
Shoulder elevation	-14.3 ± 2.7 ~ 8.3 ± 1.7
Elevation angle	-6.9 ± 0.6 ~ 12.1 ± 3.0
Shoulder rotation	-6.6 ± 2.4 ~ 7.8 ± 2.1
Elbow flexion	-12.6 ± 1.6 ~ 11.0 ± 1.8
Pronation/Supination	-3.6 ± 1.3 ~ 5.1 ± 2.0
Wrist flexion	-2.7 ± 1.2 ~ 3.9 ± 0.8
Wrist deviation	-2.5 ± 0.6 ~ 3.5 ± 0.9

III. WHEELCHAIR DRIVING SIMULATION

For the design of the wheelchair driving model we followed the approach proposed by Hase et al. for synthesis of walking simulators [7, 8]. The parameters of the simulated human motions for propulsion of the lever-propelled wheelchair were defined by using a genetic algorithm (GA).

A. Lever-propelled wheelchair model

For the analytical model shown in Figure 2, a spring floor for generation of reaction force was imposed to all areas where the musculoskeletal model and the wheelchair are in contact (grip, backrest, seat, and footrest). The spring points in the contact areas of the wheels with the road surface were set to 72 and the floor reaction forces were calculated. That allowed the calculation of the reaction force conveyed from the lever handle to the hand during the lever motion (driving force). In order to reflect in the model that the lever-propelled wheelchair mechanism transmits the force from the lever to the wheel only in one direction due to the one-way clutch, we imposed the following two conditions to the lever part of the developed model:

$$\dot{\theta}_L \geq 0 \quad \text{and} \quad \dot{\theta}_W \geq 0 \tag{1}$$

Therein, $\dot{\theta}_L$ represents the angular velocity of the lever and $\dot{\theta}_W$ denotes the angular velocity of wheel. Since it was not expected from the user to move the lever on an angular velocity greater than the wheel's angular velocity, in the model we set the following constraint condition:

$$\dot{\theta}_L \leq \dot{\theta}_W \tag{2}$$

B. Neural model

Usually, for forward dynamics simulation, we use a human-body model where the joint torque or the muscular tension is used as an input of the same model and the human motions caused by these inputs are calculated. However, this general approach may lead to generation of unnatural movements if the inputs are not selected correctly. The driving motion of the lever-propelled wheelchair might be regarded as a cyclic rhythmic motion as confirmed in the graphs of Figure 3. In view of that, for the design of human-like motion we used the approach of the central pattern generator that is applicable to the modelling of cyclic rhythmic motions.

Figure 4 shows an outline of the neural model used in this study. The neural model consists of a rhythm generation system and a sensory feedback system. The rhythm generation module consists of neural oscillators which number corresponds to the degrees of freedom of each joint. Each neural oscillator creates a basic rhythm pattern for the driven joint. In Figure 4, mark 'o' denotes one neuron and mark '=' represents inhibitory binding of two neurons. The state of each neuron of the neural oscillator is expressed with the following equations:

$$n_i = y_{i1} - y_{i2} \tag{3}$$

$$y_{i1} = \max(0, u_{i1}) \tag{4}$$

$$y_{i2} = \max(0, u_{i2}) \tag{5}$$

$$\tau_{i1}\dot{u}_{i1} = -u_{i1} - \delta y_{i2} + feed_i \tag{6}$$

$$\tau_{i2}\dot{u}_{i2} = -u_{i2} - \delta y_{i1} - feed_i \tag{7}$$

Therein, i denotes the number assigned to the neural oscillator, n_i signifies the output of the neural oscillator, u_{i1} and u_{i2} denote the internal state of the neuron, τ_{i1} and τ_{i2} represent time constants, δ stands for the weight coefficient of mutual inhibition, and $feed_i$ is the feedback signal to the ith neuron pair controlling the ith joint.

The neural model also contains a sensory feedback system that creates feedback signals corresponding to the kinetic state. The feedback signals are produced by the rhythmic pattern of the neural oscillator. The sensory feedback signals $feed_i$ were defined by referring to the approach proposed by Hase et al. [7, 8]. They are based on proportional and derivative (PD) control and consider gravity compensation as shown in (8):

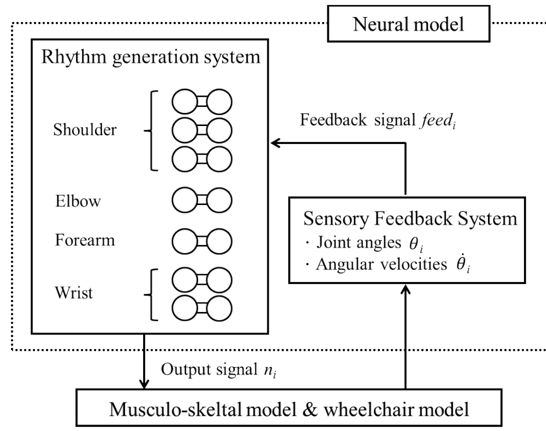

Figure 4. Outline of neural model.

$$feed_i = k_i\theta_i + c_i\dot{\theta}_i + p_i + INV_i(0,0,\theta) \tag{8}$$

In the above equation, k_i represents the proportional coefficient for the ith joint, c_i denotes the derivative coefficient, p_i denotes the stationary signal, and θ_i and $\dot{\theta}_i$ represent the joint angle and joint angular velocity respectively. INV_i (0, 0, θ) is the joint torque needed for maintaining the current posture.

The output signal of the rhythm generation module defines the driving force of the human module of the musculoskeletal model. The joint angles and the angular velocities of each joint are calculated by forward dynamics simulation. Rhythm pattern of the neural oscillator in the subsequent time point are calculated by referencing to the motion parameters. The neural system and the musculoskeletal system of the model work in cooperation.

C. Creation of driving motion

For calculation of the wheelchair lever propulsion motions we need to define the parameters τ_{i1}, τ_{i2}, δ, k_i c_i, and p_i of the nerve system module of the model. In order to do that, we assume the energy consumption per unit movement distance as a motion criterion and search the values of the parameters τ_{i1}, τ_{i2}, δ, k_i c_i, and p_i that minimize this criterion. This approach can be expressed with the following objective function:

$$I = W_J/D \;\rightarrow\; \min \tag{9}$$

In the above equation, D is the wheelchair movement distance and W_j stands for the energy consumption of the human.

The wheelchair driving motion consists of two phases: a driving phase when the lever is pushed forward and a recovery phase when the lever is moved back to its initial position. Therefore, the feedback parameters k_i, c_i, and p_i have different values for the driving phase and for the recovery phase. For each phase we need to define the values of the parameters τ_{i1}, τ_{i2}, δ, k_i, c_i, and p_i. That was done by GA procedure. The GA procedure included a number of iterations. Parameters with higher fitness were searched for each stage of the evolution process.

D. Simulation conditions

In order to verify the feasibility of the driving motion of the model calculated by the proposed method we performed a forward dynamics simulation. For simplification of the calculations, only 7 degrees of freedom for each upper extremity were considered, while the remaining joint movements were constrained. The joint torque on the output of the neural oscillator was used as an input of the human part of the model. Since the mechanical work of the joints determines the energy consumption, it was calculated with Equation (9). For the calculations we applied the following torque limitation to the model joints (see Table 1):

- Shoulder joint: ± 25 Nm,
- Elbow joint: ± 20 Nm
- Wrist joint: ± 10 Nm.

In this simulation, the number of individuals of the GA was set to 30, the time of the simulation was 1.0 s and the step width was set to 0.05 s. Matlab (R2009b; The Mathworks, Japan) was used for the calculation of the neural model, and for the parameter searching procedure.

IV. Results of simulation

The results from the motion generation simulation are shown in Figure 5. The beginning of the GA procedure, the generated driving motion get distorted when the lever is tilted backward at more than 40°. However, as search of a parameter progresses, the shape of the generated motion become much natural and close to the human-like motions (pushing and pulling the lever) (red line). A direct comparison between the synthesized motion and the measured motion cannot be made because the simulation experiment did not consider the driving resistance and the speed limitation, while the same parameters were set in advance in the physical test (section II.A). However, it can be observed that the motions generated by the model and the motion presented in Figure 3(h) have a similar shape. The lever propulsion motions change from -5° to 55° and the driving cycle continues about 0.6 s. Results show that the lever movement generated by the model is much coarse than the actual human motion during the physical experiment. This initial simulation model includes only basic functions but it demonstrates that the proposed simulation approach can be used for modelling of the natural movement propulsion patterns. However, the simulation model needs to be improved further to allow its application into clinical practice. The shape of the generated motion can be improved by reducing the step width, increasing the simulation time, and changing the definition of the objective function.

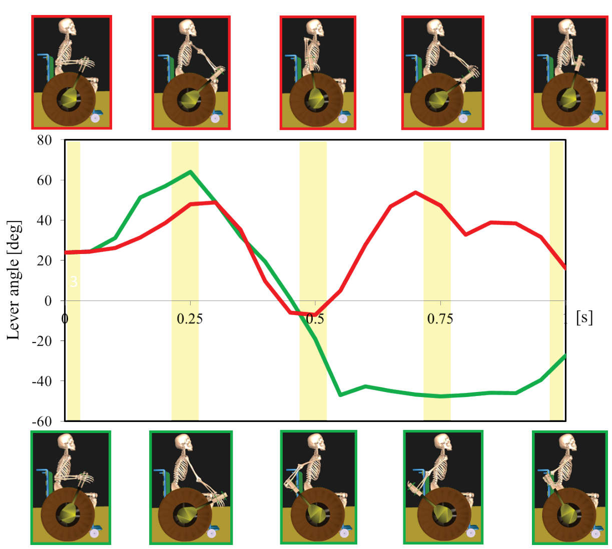

Figure 5. Results of creation of driving motions.

V. Conclusion

In this study, a model for simulation of human motions during wheelchair lever propulsion was proposed. It contains a neural part and a musculoskeletal part. At the beginning of this study we performed a wheelchair propulsion experiment that allowed us to analyze the kinematic features of lever propulsion. The results from the physical experiment were used for the development of the model. Although this initial model cannot produce very smooth motions due to design simplifications, it proves the proposed approach. It is expected that the proposed technique could be applied also to the modelling of handrim-propelled wheelchairs to be used for adaptation of the wheelchair characteristics to the individual user. Future studies will be conducted for improvement of the simulation model.

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