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Predictive Forward Dynamic Simulation of Manual Wheelchair Propulsion on a Rolling Dynamometer

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ABSTRACT

Research studies to understand the biomechanics of manual wheelchair propulsion often incorporate experimental data and mathematical models. This project aimed to advance this field of study by developing a two-dimensional model to generate first of its kind forward dynamic fully predictive computer simulations of a wheelchair basketball athlete on a stationary ergometer. Subject-specific parameters and torque generator functions were implemented in the model from dual x-ray absorptiometry and human dynamometer measurements. A direct collocation optimization method was used in a wheelchair propulsion model for the first time to replicate the human muscle recruitment strategy. Simulations were generated for varying time constraints and seat positions. Similar magnitudes of kinematic and kinetic data were observed between simulation and experimental data of a first push. Furthermore, seat heights inferior to the neutral position were found to produce similar joint torques to those reported in previous studies. An anterior seat placement produced the quickest push time with the least amount of shoulder torque required. The work completed in this project demonstrates that fully predictive simulations of wheelchair propulsion have the potential of varying simulation parameters to draw meaningful conclusions.

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INTRODUCTION

It is common for people with a spinal cord or lower limb injury to require a manually propelled wheelchair to complete activities of daily living. In Canada, a review of the Canadian Survey on Disability (2012) determined that approximately 200,000 Canadians are manual wheelchair users [1]. Improving biomechanical models for manual wheelchair propulsion is important to help paraplegic individuals complete activities of daily living with less risk of injury. Development of this research area is also sought of by coaches and athletes to improve performance and mitigate injury risk in Paralympic sport. The standard method of making a wheelchair setup change is currently a lengthy and expensive process, and is carried out on a trial-and-error basis with an athlete going through several wheelchairs over their career [2]. Therefore, the goal of this project was to develop a validated, fully predictive computer simulation of wheelchair propulsion, and explore the feasibility of using it to provide individualized recommendations of optimal wheelchair settings. The major benefit of using a computer simulation would be the drastic reduction in the cost and effort required to fit an athlete to a new wheelchair.

Since the early 1980s, mathematical models combined with experimental data have proven to be useful tools to study wheelchair propulsion [3]. 3D motion capture systems are typically used to determine body segment kinematics, while electromechanical strain-gauge wheel hubs are used to measure wheel-hand reaction forces, such as those developed by [4] and [5]. 2D models have been utilized by researchers for the ability to perform a realistic quasi-static or inverse dynamic analysis.

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Richter [6], and later Leary et al. [7], demonstrated this by using a four bar mechanism to model the wheelchair-user system. These studies found that the torque required by the shoulder exceeds the torque generated at the elbow.

One of the fundamental wheelchair parameters identified by the Canadian Sports Institute Ontario (CSIO) and previous literature [6–9] for both performance and injury prevention was the position of the hips relative to the wheel axle. In [6] and [7], the fixed shoulder position was adjusted vertically, and it was determined that a lower seat position minimized the required concentric shoulder torque but increased the required elbow torque. Munaretto et al. [9] conducted an inverse dynamic analysis of a four bar mechanism with varying shoulder positions, and found similar results to [7]. The studies of Munaretto et al. used contour maps to show the large solution space that varying parameters provide to a simulation. Ideally, an optimization method would strive to reach the best solution through the selection of optimal parameters.

Optimization methods have been used in a variety of human biomechanics inverse and forward dynamic studies such as walking [10], sit to stand [11], and various sports applications [12–16]. There has recently been an increase in the number of forward dynamic modeling studies to track wheelchair user kinematics and kinetics, mostly from Neptune et al. [17–21] using Visual3D and SIMM, however none of these studies have measured subject-specific torque capability to be included as bounds. These studies utilized a complex single-sided upper extremity model with 26 Hill-type musculotendon actuators [22], which included the trunk, upper arm, forearm, and hand. This model was verified by the close comparison of inverse dynamic results to a similar model in a different software program, the Anybody Modeling System, by [23]. Trunk motion was prescribed based on experimental data, and scapular and clavicular motions were prescribed as functions of shoulder elevation. Optimal muscle activation inputs were calculated to track experimental kinematic and kinetic data collected from paraplegic subjects performing steady-state sub-maximal propulsion on a wheelchair treadmill. The forward dynamic simulations tracked experimental data well with average errors of 1.12° and 2.36 N between joint kinematics and handrim forces, respectively. Results from these studies consistently found that the shoulder flexors are the most active during the push phase. It is of interest to observe the results of a forward dynamic model that does not track time series data, prescribe trunk motion or track experimental data in the objective function, allowing for the model to be more predictive in the effect of varied parameters.

Slowik et al. [19] investigated how seat position influenced musculoskeletal demand in wheelchair propulsion. In the computer model, the shoulder position was fixed and the push angle range was defined as a function of seat height in the same way as [6,7]. The cycle time was fixed to 1 s at a constant push frequency to simulate a steady state push and the hand position throughout the simulation was prescribed. Furthermore, an average power output at the hand was fixed at 10 W. The results of the study found that muscle stress, co-contraction and metabolic cost were all minimized at anterior horizontal positions of the shoulder between -14 cm and -3 cm from the center wheel axle, and a superior offset between -1 cm and 3 cm. Past a -10° hub-shoulder angle (posterior seat positions), it was found that upper extremity demand increased

[19]. This forward dynamic simulation method can be improved by including predictive trunk motion and allowing the average hand force to vary between different parameter value trials.

This study takes direct aim to advance the application of forward dynamic modeling to wheelchair propulsion research. The first fully predictive forward dynamic model of wheelchair propulsion is presented here, meaning the optimized muscle input activations are predicted without the requirement of prescribed hand motion, joint motion, cycle time, or hand force. Furthermore, the first push was simulated, which has received limited attention in literature, and can be considered as one the of the most important pushes in many wheelchair sports [24,25]. Direct collocation is used to determine optimal activation profiles that don't constrain the solution to the shape of pre-defined functions, such as the parameterized control inputs used in previous forward dynamic wheelchair propulsion studies [18–21,26,27]. Direct collocation has been previously used in a predictive wheelchair curling study and was found to provide realistic activation torques for a closed-chain dynamic model [28]. In addition, [29–31] suggest that higher mobility wheelchair users utilize the torso significantly in wheelchair propulsion. To accommodate this insight, an active torso joint and body segment was included in seat variation simulations to provide additional insight and accuracy to a forward dynamic model. Finally, human dynamometer testing with wheelchair Paralympic athletes was utilized for the first time to obtain personalized torque functions in a forward dynamic simulation.

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METHODS

CSIO Experimental Data

One female subject (WCB003) was used for the testing protocol of this study with a severe, non-spinal lower body knee injury (28yrs, 170cm, 67kg, Class 4-4.5), and although not a full-time wheelchair user, has played as an active member of the Canadian Women's Wheelchair Basketball Team for over a year. Ethics approval for the testing protocol was obtained from the University of Waterloo Office of Research Ethics. Data collected by the CSIO in which the subject pushed an adjustable wheelchair set to her usual configuration on top of a motor-controlled dynamometer were shared to aid in the research of this project. The ergometer provided a track simulated and inertia adjusted wheel resistance (Keku Inc.) determined from the equations developed by [32]. Athletes were instructed to push for 30 seconds at 80% effort. 3D motion capture technology was used to collect kinematic data of the upper body segments, which included data for the upper arms, forearms, hand, and wheel. 3D hand reaction forces were measured using SMARTWheel Technology (Out-Front).

System Model and Body Segment Inertial Parameters

A five segment, joint torque driven 2D planar model was developed to represent the system of wheelchair and user with MapleSim (Maplesoft, Canada), a multibody dynamics simulation software package. Five revolute joints were included to represent flexion and extension of the torso, shoulder, elbow, and wrist, as well as rotation of the wheel. The rigid bodies of the model represented the torso, upper arm, forearm, hand, and wheel, to which the hand segment was rigidly attached. The model schematic can be found in Fig. 1. From experimental data, an average contact angle for the first push between the hand and wheel segment of 53.7° counter-clockwise of the radial wheel segment component about the *z* axis was provided to the model.

The resulting model contained two degrees of freedom and was represented by four generalized coordinates coupled by two algebraic constraints. The seat position was located -11.2 cm posterior and +18 cm superior to the wheel axis, which was obtained from the experimental marker data. Since the subject's hip joint was hidden from view by the motion capture system, the location of the hip was estimated from markers placed on the outside of the wheelchair. Body segment mass was measured by dual x-ray absorptiometry (iDXA) measurements of the subject, and center of mass and inertia terms were obtained by combining measurements with anthropometric charts obtained using a gamma-ray scanning technique of college-aged Caucasian males and females [33]. 3D motion capture technology was used to measure segment lengths. Due to the 3D movement of the arm that occurs during the push phase in wheelchair propulsion, the representative body segment inertial parameters (BSIPs) for the 2D projected model are variable when viewed in a single plane. The BSIPs for the 2D model were obtained using a projection method previously employed in a study by [34]. The resulting 2D BSIPs are found in Tab. 1, where *L* is length, *CoM* is center of mass relative to the proximal joint, *M* is mass, and *I* is inertia about the center of mass. The resulting torques of an inverse dynamic analysis between the 2D projected model and CSIO show good agreement in Fig. 2. The elbow torque had the largest discrepancy, likely because the elbow and wrist coordinate systems used by Visual3D were not documented and had to be estimated.

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Joint Torque Functions

The net torque produced by each joint is described in Eqn. 1,

$$\tau = \tau_{act}(t)\tau_V(\omega)\tau_A(\theta) + \tau_P(\theta) \quad (1)$$

where $\tau_{act}(t)$ is the activation torque, $\tau_V(\omega)$ is the torque-velocity scaling, $\tau_A(\theta)$ is the torque-angle scaling, and $\tau_P(\theta)$ is the passive torque. $\tau_{act}(t)$ could take any value between the upper and lower isometric torque at the optimal angle. $\tau_V(\omega)$ and $\tau_A(\theta)$ were normalized by the maximum isometric torque of each joint. Joint torque testing was completed using the Biodex System 4 Pro ™ (Biodex Medical Systems, Inc, Shirley, NY). The testing protocol was broken into isometric and isokinetic testing of the shoulder and elbow of the dominant side of the body. Although the subject was not representative of many athletes in the sport with spinal or other upper body injuries, the time-consuming and physically demanding nature of this protocol provided initial datasets with more accuracy and repeatability as a baseline for future studies. Due to these challenges, the testing focused on the major muscle groups of the arm only; therefore, the wrist was not included in this study. The subject completed isometric testing by exerting a five second maximal flexor activation and extensor activation. The subject exerted a single 5 second maximal flexor activation against the stationary crank, which was immediately followed by a 5 second maximal extensor activation. A minimum of 15 seconds of rest was given between joint angles, with longer breaks taken by the subject when needed to ensure maximal exertion in the following rep. Three minutes of rest was provided between isometric and isokinetic testing. This was completed for 8 different angles of the shoulder (-40°, -20°, 0°, 20°, 40°, 60°, 80°, and 100°) with an

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elbow angle of 0°, and 9 for the elbow (0°, 15°, 30°, 45°, 60°, 75°, 90°, 105°, 120°) with a shoulder angle of 45°. Following this, the subject exerted a maximal effort for two repetitions of a full concentric/eccentric cycle for each joint and velocity tested (30 deg/s, 75 deg/s, 120 deg/s, 180 deg/s, 240 deg/s, 300 deg/s, and 360 deg/s) for both the flexor and extensor muscle groups. The maximal torque isometric and isokinetic data sets were obtained using the regression method proposed by [35].

The maximal data set was fit to a seven-parameter function, $\tau_V(\omega)$, which was used to express maximum voluntary torque as a function of angular velocity. This function was developed in a previous study by [35]. The parameters of this function were identified with a weighted nonlinear least-squares solver in MATLAB called *lsqnonlin*. The majority of experimental data points were forced beneath the fitted function to account for the difficulty in obtaining consistent maximum voluntary contractions during testing. For the fitting process, the coefficient of determination (R-squared) between the experimental data and fitted function was found to be 0.906, 0.855, 0.964, and 0.947, for the shoulder flexors, elbow flexors, shoulder extensors, and elbow extensors, respectively.

For the torso and wrist joints in the model, a generic concentric torque-velocity function was utilized that has been used extensively in golf swing predictive simulation research [13–15,36], and the eccentric torque-velocity function was obtained by [15]. This function was chosen due to its ease of implementation in a forward dynamic model, use in various application that include jumping [37], as well as the previous success obtained in modeling an upper body joint dependent sport movement.

A second-order polynomial was used to express the torque-angle relationship $\tau_A(\theta)$, which was similar to the functions used by [38,39]. For the unique profile produced by the flexor muscles of the shoulder in this study, a fourth-order polynomial was used. Data for the wrist joint was obtained from human dynamometer data collected by [40]. A torque-angle function for the torso was not included in this study. The corresponding R-squared value were 0.965, 0.729, 0.946, and 0.718 for the shoulder flexors, elbow flexors, shoulder extensors, and elbow extensors.

Passive Torque

A passive torque function of angle and angular velocity developed by [41] were included in the model for each joint except the torso to represent the restoring forces that are produced by muscle tissue, tendons, and ligaments that surround the joint. Function parameters were fit by data obtained from [15] for the shoulder, [14] for the wrist, and [42] for the elbow.

Ergometer Resistive Torque

The experimental data collection by the CSIO was performed on top of a wheelchair ergometer (Keku Inc.), which consisted of a high inertia roller directly coupled with a servo motor. The controller in the ergometer provided an equivalent rolling resistance, aerodynamic drag, as well as the inertial resistance of the athlete and wheelchair, which was based on the work by Fuss [32]. To prevent an initial backwards rotation of the wheel in the simulation from the constant rolling resistance term, a continuous method developed by [43] was applied to eliminate the significance of this term for wheel velocities close to zero.

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The wheel inertia for the 2D model was provided a value that produced an equivalent inertial force at the wheel, which is further described in Eqn. 2. The result of I_{m_z} is shown in Tab. 1.

$$m\dot{\omega}_{wc} r_{\omega}^{2} = I_{m_z}\dot{\omega}_{wc} \rightarrow I_{m_z} = m r_{\omega}^{2} \quad (2)$$

Although the combined athlete and chair weight was 80.2 kg, only half the value of mass and inertia was used as only the right side of the athlete was modeled.

Forward Dynamic Simulation

Optimal Control Method

MapleSim was used to obtain the differential-algebraic dynamic equations of the 2D model, which were converted to ordinary differential equations. This was done by using Baumgarte's constraint stabilization method [44]. This equation was then numerically integrated simultaneously with the ODEs from the dynamic equations. A direct collocation method (GPOPS-II) was used to solve the optimization problem. In this method, the state and control are solved by approximation over a subinterval by a nth-degree polynomial. This solution is obtained iteratively through mesh refinement methods until the objective function is minimized with satisfied constraints [45].

Initial Conditions and Bounds

The initial torso and wheel angle were selected to match the experimental data of WCB003 at the start of the first push. The initial joint angles for the remaining upper body degrees of freedom were selected to match as close as possible to the experiment. The final wheel angle was fixed to match the final experimental wheel angle of the push phase. Range of motion limits were set to approximate the range of motion of each joint. Joint angular velocity limits were similar to the values used in previous predictive biomechanical simulations [14,15,36]. Activation torque limits for the shoulder and elbow were equal to the maximum isometric torque measured of the subject for the flexor and extensor activations. The isometric torque estimate for the torso was determined from a flexor and extensor test conducted by Williams [46]. Values for the maximum wrist torque have varied by large margins in literature. In previous forward dynamic golf swing simulations, the wrist torque for a single arm swing was bounded by 90 Nm [15]. In [40], isometric wrist torque was measured at varying anatomical positions and found to peak at approximately 23 Nm. For this study, various wrist torque limits were simulated and presented in the following section. Rate of torque development (RTD) bounds were selected based on similarity of activation shape to literature [13–15,47,48] and ease of optimization (i.e., helped the optimizer reach an optimal solution with respect to mesh and solver tolerances).

Objective Function

The objective function used by [19] minimized the sum of change in joint moments and change in handrim forces in a high-fidelity muscle model. In this study, difficulties were encountered optimizing an objective function with a rate of change of hand force penalty term and the resulting hand forces were much larger than experimental values. Therefore, a modified version of this objective function was used, in which the sum of the change in activation torques, $\dot{U}$, and the hand forces, F, was minimized:

$$J = \int_{t_0}^{t_f} (\dot{U}^T \dot{U} + wF^T F)\, dt, \quad (3)$$

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where a weighting of 10, which was determined to best match experimental hand forces, was used for w. In addition, minimizing F was found to produce smoother kinematic and kinetic results.

Simulation Trials

Simulations were generated for a single push and were compared to experimental kinematic and kinetic ergometer first push data of WCB003. It was found that the initial wheel speed was not equal to zero in the experimental first push; therefore, a non-zero initial wheel speed simulation for the first push was generated for validation purposes. The non-zero initial wheel speed required non-zero initial joint angular velocities to satisfy a continuous solution. An initial angular velocity of zero at the torso was selected. Using experimental initial joint velocities as guesses, the remaining initial joint angular velocities were selected that satisfied the kinematic constraints of the model. Furthermore, the wrist torque was bounded to a range ±20 Nm.

Next, simulations were generated from rest that varied seat position in a maximum effort push by adding a penalty term to *J* of Eqn. 3 that minimized the final time of the push. The weighting of this term was increased over many simulations of the same seat position until the difference in final push time was negligible. Due to the variation in initial conditions of the model between seat positions, it was difficult to compare directly to the experimental condition. The same model for WCB003 described above was used, however the initial angles were altered to match those of a different subject in the CSIO study to provide contact and release positions before and after top

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dead center (TDC), respectively. This type of push angle range is found much more consistently in literature, and allowed for a better comparison of seat position dynamics to previous studies. In addition, the initial torso angle was selected to be upright at 5°. Further constraints were added to the optimization to help achieve a realistic push, which included the final torso velocity constrained to a range of ±60 deg/s, the final shoulder angle constrained to achieve a minimum displacement of 25°, and the wrist torque bound to a range of ±40 Nm. Simulations were generated for vertical seat positions of -5 cm and -10 cm relative to the neutral position and with respect to the inertial frame. Due to issues with the optimizer, which were likely related to the kinematic constraints, seat positions greater than the neutral position could not be tested. Simulations were also generated for horizontal seat positions of -10 cm and +10 cm from the neutral position.

RESULTS

Predictive Simulation Validation

Figure 3 represents the net torque produced by each biomechanical joint in the non-zero initial wheel velocity simulation. The root mean square deviation (RMSD) was 13.6 Nm, 4.99 Nm, and 12.9 Nm for the shoulder, elbow, and wrist, respectively. Further analysis of the simulated elbow torque shows that the joint provided a flexor activation followed by an extensor activation in the second half of the push. The use of torque-velocity-angle scaling allowed for a closer analysis of torque generation for each joint, which is found in Fig. 4 for the elbow joint.

Tangential (x) and normal (y) hand forces produced in the simulation are similar in overall magnitude to those measured experimentally, as shown in Fig. 5. The tangential RMSD was 50.4 N, and the normal RMSD was 18.2 N. The reason for the larger tangential RMSD is due to the difference in profile compared to that measured experimentally. The optimizer selected a profile that ramped up quickly to a lower maximum tangential force, whereas the experimental profile ramped up more slowly to a higher maximum force.

The kinematic results are found in Fig. 6. The RMSD was 7.60°, 3.07°, 5.64°, 13.5°, and 0.458° for the torso, shoulder, elbow, wrist, and wheel, respectively.

Varying Seat Position and Maximal Performance

Maximal effort simulations starting from rest resulted in a much larger joint torque requirement than when the initial wheel velocity is non-zero, which can be seen when comparing the difference in simulation results for a neutral seat position simulation between Fig. 3 and Fig. 7. One of the largest difference between the two simulations is found in the large torso torque generated. The net shoulder joint torque requirement was lower as the seat position decreased vertically. As discussed previously, the shoulder joint is consistently the most active in the upper extremity during a wheelchair push, and as a result is typically injury-prone and the largest contributor to performance. Therefore, increased analysis was focused on this joint. The largest peak shoulder torque was in the neutral position at 88.4 Nm, which was followed by 77.6Nm and 77.5 Nm for the -5 cm and -10 cm position, respectively. For varying horizontal seat positions, the least amount of shoulder torque required to complete the push was

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found to be in the +10 cm seat position with a peak torque of 74.0 Nm. This was followed by peak torques of 88.4 Nm and 97.0 Nm in the neutral and -10 cm condition, respectively. This can be observed in Fig. 8.

In terms of push performance, the model had a greater acceleration as chair height decreased. It was found that the neutral optimal performance time was 0.346 s, which compares well to the push time of 0.342 s measured by the athlete. Although the athlete pushed through a different wheel angle range, had an initial non-zero wheel velocity and was instructed to push at only 80% of maximal effort, this comparison provides a qualitative check of the realistic optimal push time determined in these simulations. The push time for vertical seat change of -5 cm was 0.338 s, and for -10 cm was 0.331 s. For horizontal seat positions, the push time showed a decrease as seat position moved anteriorly. The anterior position had a push time of 0.334 s, whereas the posterior position had a push time of 0.356 s. The results from these simulations are summarized in Tab. 2.

DISCUSSION

The validation simulation hand forces produced larger tangential forces than normal forces, which was expected as the propulsion force is purely tangential. Furthermore, maximum hand forces were observed much earlier in the push phase than were measured experimentally. The tangential force increased more quickly than the experimental condition, but plateaued to a lower maximum value.

Other than the torso, the kinematic RMSD was low in all joints between the validation simulation and experimental data, as shown in Fig. 6. However, to satisfy kinematic constraints, deviations in initial and final wrist and elbow angle can be observed between simulation and experiment. Although these deviations are present, the elbow reaches a maximum angle within close proximity to the experiment. The wrist angle was further compromised by the variable experimental hand angle throughout the push. However, the wrist angle reached an inflection point at a similar time between simulation and experiment.

Given the full functionality in the torso of WCB003, the large torque required falls well within the maximum torso torque generation limit, and has found to be large for able-bodied subjects (ie. [29–31]). Furthermore, previous studies have found that the wrist is the lowest torque producing joint of the upper extremity during wheelchair propulsion [3,49–51]. When left unconstrained, it was found that the optimizer used in this study favored an unreasonably larger wrist torque than all other joints. Therefore, different bounds of maximum wrist torque were used depending on the effort required by the joints in each simulation.

The change in shoulder torque pattern of this study for a vertical seat change has been observed previously in [7,9]. Furthermore, Slowik et al. concluded that an optimal vertical seat position that minimized musculoskeletal demand for daily wheelchair users in a steady state push should result in the elbow achieving a TDC angle of 110° to 120°. The TDC values in this study were considerably lower than this, which were 47.3° (neutral), 68.8° (-5 cm) and 85.4° (-10 cm). Shoulder torque decreased as TDC elbow angle approached the optimal range. Slowik et al. discussed how previous studies concluded that a posterior seat position was optimal to reduce joint loading. However,

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Slowik et al. concluded that a shoulder-wheel-hub angle range of -10° to -2.5° was optimal, and that angles lower than -10° increased musculoskeletal demand. The initial shoulder-wheel-hub angles tested in this study corresponded to -13.6° (-10 cm), -4.92° (neutral), and +3.96° (+10 cm).

It is difficult to compare directly to Slowik et al. due to the large difference in models and simulation methods, with Slowik et al. investigating daily wheelchair users and not athletes, and including the shoulder joint being held fixed during propulsion simulations in [19]. However, in a qualitative comparison, it appeared that a shoulder-wheel-hub angle lower than -10° in this study also increased demand in the shoulder. Furthermore, the anterior position simulated in this study was also outside the optimal range suggested by Slowik et al., yet produced lower peak torques in the shoulder and elbow than in the neutral simulation.

From these simulations, it seemed that an anterior or inferior seat position could be beneficial to alleviate the required torque production in the shoulder joint as well as to decrease push time. Overall, the fastest push time was produced by the -10 cm inferior offset from the default seat position, and the +10 cm anterior offset from the default seat position required the least amount of shoulder torque.

The main goal of this work was to develop a validated, fully predictive computer simulation of wheelchair propulsion. In conclusion, forward dynamic simulations were generated to provide similar profiles and magnitudes of kinematic and kinetic data between fixed final time simulations and experimental data of a sub-maximal first push. Wrist torque results remain to be further validated, as the optimizer tended to

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maximize wrist torque when it is known that the wrist plays the smallest role in energy production during a wheelchair push. Additional simulations were generated that varied the seat position and used an additional objective function term that minimized push time to simulate a maximal effort from rest. These simulations resulted in push times that compared well to experimental data for the first push. Seat heights inferior to the neutral experimental position were found to produce similar joint torque effects to those reported in previous modeling studies. An anterior seat placement to the neutral experimental position produced the quickest push time with the least amount of shoulder torque required. Variations in this model compared to those in literature, as well as the model parameter identification of only one subject, provided limited validation of these seat adjustment findings. The work completed in this project demonstrates that fully predictive simulations of wheelchair propulsion can produce realistic results, particularly for first push kinematics and hand force magnitudes. With further refinement of joint torque models, particularly for the wrist, the methods of this work demonstrate the potential of varying simulation parameters to make meaningful conclusions.

Future work should improve upon this study by continuing the validation of this method through testing more subjects and increasing the complexity of the model. Although iDXA measurements were used to obtain subject-specific masses, future studies should advance this approach by approximating center of mass and moment of inertia from iDXA measurements. One of the limitations of the 2D model is that the camber angle of the wheel could not be taken into consideration. Another improvement to future studies would be the measurement of experimental data over-ground rather than on an ergometer, as variations in measurement have been reported in literature [52,53]. In addition, a hand-grip model could be implemented to limit the tangential hand force and capture the variable hand-wheel angle observed experimentally. Furthermore, an inclusion of wrist torque to the objective function might counter the optimizers preference for maximizing wrist torque. It is common knowledge that other factors are involved in wheelchair sports that influence top performance, including agility, ball handling, and stability. The effect on these attributes of performance should be explored further with changes in seat position and push angle range. Finally, a higher-fidelity muscle model should be explored at each joint that aims to provide smoother transitions between eccentric and concentric torque generation (see Fig. 4), such as in the study by Slowik et al. [19], and combined with a fully predictive propulsion simulation.

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REFERENCES

[1] Smith, E. M., Giesbrecht, E. M., Mortenson, W. B., and Miller, W. C., 2016, "Prevalence of Wheelchair and Scooter Use among Community-Swelling Canadians," Phys. Ther., **96**(8), pp. 1135–1142.

[2] Mason, B. S., Van Der Woude, L. H. V., and Goosey-Tolfrey, V. L., 2013, "The Ergonomics of Wheelchair Configuration for Optimal Performance in the Wheelchair Court Sports," Sport. Med., **43**(1), pp. 23–38.

[3] Cerquiglini, S., Figura, F., Marchetti, M., and Ricci, B., 1981, "Biomechanics of Wheelchair Propulsion," Proc. Seventh Int. Congr. Biomech., **3A**, pp. 410–419.

[4] Asato, K. T., Cooper, R. A., Robertson, R. N., and Ster, J. ., 1993, "SMART Wheels: Development and Testing of a System for Measuring Manual Wheelchair Propulsion Dynamics," IEEE Trans. Biomed. Eng., **40**(12), pp. 1320–1324.

[5] Wu, H. W., Berglund, L. J., Su, F. C., Yu, B., Westreich, A., Kim, K. J., and An, K. N., 1998, "An Instrumented Wheel for Kinetic Analysis of Wheelchair Propulsion," J. Biomech. Eng., **120**(1), pp. 533–535.

[6] Richter, W. M., 2001, "The Effect of Seat Position on Manual Wheelchair Propulsion Biomechanics: A Quasi-Static Model-Based Approach," Med. Eng. Phys., **23**(10), pp. 707–712.

[7] Leary, M., Gruijters, J., Mazur, M., Subic, A., Burton, M., and Fuss, F. K., 2012, "A Fundamental Model of Quasi-Static Wheelchair Biomechanics," Med. Eng. Phys., **34**(9), pp. 1278–1286.

[8] Desroches, G., Aissaoui, R., and Bourbonnais, D., 2006, "Effect of System Tilt and Seat-to-Backrest Angles on Load Sustained by Shoulder during Wheelchair Propulsion," J. Rehabil. Res. Dev., **43**(7), pp. 871–882.

[9] Munaretto, J. M., McNitt-Gray, J. L., Flashner, H., and Requejo, P. S., 2013, "Reconfiguration of the Upper Extremity Relative to the Pushrim Affects Load Distribution during Wheelchair Propulsion," Med. Eng. Phys., **35**(8), pp. 1141–1149.

[10] Anderson, F. C., and Pandy, M. G., 2001, "Dynamic Optimization of Human Walking," J. Biomech. Eng., **123**(5), p. 381.

[11] Norman-gerum, V., and Mcphee, J., 2018, "Constrained Dynamic Optimization of Sit-to-Stand Motion Driven by Bezier Curves," J. Biomech. Eng., **140**(12).

[12] Neptune, R. R., and Hull, M. L., 1998, "Evaluation of Performance Criteria for

Simulation of Submaximal Steady-State Cycling Using a Forward Dynamic Model," J. Biomech. Eng., **120**(3), p. 334.

[13] MacKenzie, S. J., and Sprigings, E. J., 2009, "A Three-Dimensional Forward Dynamics Model of the Golf Swing," Sport. Eng., **11**(4), pp. 165–175.

[14] Balzerson, D., Banerjee, J., and McPhee, J., 2016, "A Three-Dimensional Forward Dynamic Model of the Golf Swing Optimized for Ball Carry Distance," Sport. Eng., **19**(4), pp. 237–250.

[15] McNally, W., and McPhee, J., 2018, "Dynamic Optimization of the Golf Swing Using a Six Degree-of-Freedom Biomechanical Model," MDPI Proc. 12th Conf. Int. Sport. Eng. Assoc., **2**(6).

[16] Jansen, C., and McPhee, J., 2018, "Predictive Dynamic Simulation of Seated Start-Up Cycling Using Olympic Cyclist and Bicycle Models," Proceedings, **2**(6), p. 220.

[17] Rankin, J. W., Kwarciak, A. M., Mark Richter, W., and Neptune, R. R., 2010, "The Influence of Altering Push Force Effectiveness on Upper Extremity Demand during Wheelchair Propulsion," J. Biomech., **43**(14), pp. 2771–2779.

[18] Rankin, J. W., Richter, W. M., and Neptune, R. R., 2011, "Individual Muscle Contributions to Push and Recovery Subtasks during Wheelchair Propulsion," J. Biomech., **44**(7), pp. 1246–1252.

[19] Slowik, J. S., and Neptune, R. R., 2013, "A Theoretical Analysis of the Influence of Wheelchair Seat Position on Upper Extremity Demand," Clin. Biomech., **28**(4), pp. 378–385.

[20] Morrow, M. M., Rankin, J. W., Neptune, R. R., and Kaufman, K. R., 2014, "A Comparison of Static and Dynamic Optimization Muscle Force Predictions during Wheelchair Propulsion," J. Biomech., **47**(14), pp. 3459–3465.

[21] Slowik, J. S., Requejo, P. S., Mulroy, S. J., and Neptune, R. R., 2016, "The Influence of Wheelchair Propulsion Hand Pattern on Upper Extremity Muscle Power and Stress," J. Biomech., **49**(9), pp. 1554–1561.

[22] Morrow, M. M. B., Kaufman, K. R., and An, K. N., 2010, "Shoulder Model Validation and Joint Contact Forces during Wheelchair Activities," J. Biomech., **43**(13), pp. 2487–2492.

[23] Dubowsky, S. R., Sisto, S. A., and Langrana, N. A., 2009, "Comparison of Kinematics, Kinetics, and EMG throughout Wheelchair Propulsion in Able-Bodied and Persons with Paraplegia: An Integrative Approach," J. Biomech. Eng., **131**(2), p. 021015.

[24] Van der Slikke, R. M. A., de Witte, A., Berger, M. A. M., Bregman, D. J. J., and Veeger, D., 2018, "Wheelchair Mobility Performance Enhancement by Changing Wheelchair Properties: What Is the Effect of Grip, Seat Height, and Mass?," Int. J. Sports Physiol. Perform., **13**(8), pp. 1050–1058.

[25] Vanlandewijck, Y., Theisen, D., and Daly, D., 2001, "Wheelchair Propulsion Biomechanics: Implications for Wheelchair Sports," Sport. Med., **31**(5), pp. 339–67.

[26] Neptune, R. R., 1999, "Optimization Algorithm Performance in Determining Optimal Controls in Human Movement Analyses," J. Biomech. Eng., **121**(2l), pp. 249–252.

[27] Pandy, M. G., Anderson, F. C., and Hull, D. G., 1992, "A Parameter Optimization Approach for the Optimal Control of Large-Scale Musculoskeletal Systems," J. Biomech. Eng., **114**(4), pp. 450–460.

[28] Laschowski, B., Mehrabi, N., and McPhee, J., 2018, "Optimization-Based Motor Control of a Paralympic Wheelchair Athlete," Sport. Eng., **2**, pp. 1–9.

[29] Guo, L. Y., Su, F. C., Wu, H. W., and An, K. N., 2003, "Mechanical Energy and Power Flow of the Upper Extremity in Manual Wheelchair Propulsion," Clin. Biomech., **18**(2), pp. 106–114.

[30] Guo, B., Duan, S., and Ghalambor, A., 2006, "Effect of Handrim Diameter on Manual Wheelchair Propulsion: Mechanical Energy and Power Flow Analysis," SPE Prod. Oper., **21**(1), pp. 107–113.

[31] Rodgers, M. M., McQuade, K. J., Rasch, E. K., Keyser, R. E., and Finley, M. A., 2003, "Upper-Limb Fatigue-Related Joint Power Shifts in Experienced Wheelchair Users and Nonwheelchair Users," J. Rehabil. Res. Dev., **40**(1), p. 27.

[32] Fuss, F. K., 2009, "Influence of Mass on the Speed of Wheelchair Racing," Sport. Eng., **12**(1), pp. 41–53.

[33] De Leva, P., 1996, "Adjustments to Zatsiorsky-Seluyanov's Segment Inertia Parameters," J. Biomech., **29**(9), pp. 1223–1230.

[34] Ghannadi, B., 2017, "Model-Based Control of Upper Extremity Human-Robot Rehabilitation Systems," University of Waterloo.

[35] Yeadon, M. R., King, M. A., and Wilson, C., 2006, "Modelling the Maximum Voluntary Joint Torque-Angular Velocity Relationship in Human Movement," J. Biomech., **39**(3), pp. 476–482.

[36] Brown, C., and McPhee, J., 2018, "Dynamics of the Golf Club-Grip Interaction,"

MDPI Proc. 12th Conf. Int. Sport. Eng. Assoc., **2**(6).

[37] Alexander, R. M., 1990, “Optimum Take-Off Techniques for High and Long Jumps,” Philos. Trans. R. Soc. B Biol. Sci., **329**(1252), pp. 3–10.

[38] King, M. A., Wilson, C., and Yeadon, M. R., 2006, “Evaluation of a Torque Driven-Model of Jumping for Height,” J. Appl. Biomech., **22**, pp. 264–274.

[39] Forrester, S. E., Yeadon, M. R., King, M. A., and Pain, M. T. G., 2011, “Comparing Different Approaches for Determining Joint Torque Parameters from Isovelocity Dynamometer Measurements,” J. Biomech., **44**(5), pp. 955–961.

[40] Garner, B. A., and Pandy, M. G., 2001, “Musculoskeletal Model of the Upper Limb Based on the Visible Human Male Dataset,” Comput. Methods Biomech. Biomed. Engin., **4**, pp. 93–126.

[41] Yamaguchi, G. T., 2001, *Dynamic Modeling of Musculoskeletal Motion: A Vectorized Approach for Biomechanical Anlaysis in Three Dimensions*, Kluwer Academic Publishers, Norwell, Massachusetts.

[42] Lemay, M. A., and Crago, P. E., 1996, “A Dynamic Model for Simulating Movements of the Elbow, Forearm, and Wrist,” J. Biomech., **29**(10), pp. 1319–1330.

[43] Brown, P., and McPhee, J., 2016, “A Continuous Velocity-Based Friction Model for Dynamics and Control with Physically Meaningful Parameters,” J. Comput. Nonlinear Dyn., **11**(5), p. 054502.

[44] Baumgarte, J., 1972, “Stabilization of Constraints and Integrals of Motion in Dynamical Systems,” Comput. Methods Appl. Mech. Eng., **1**(1), pp. 1–16.

[45] Patterson, M., and Rao, A., 2012, “GPOPS II: A MATLAB Software for Solving Multiple-Phase Optimal Control Problems Using Hp-Adaptive Gaussian Quadrature Collocation Methods and Sparse Nonlinear Programming,” ACM Trans. Math. Softw., **41**, pp. 1–37.

[46] Williams, C., 1992, “Maximum Concentric, Eccentric, and Isometric Strength of Trunk Flexor and Extensor Muscles in Athletes,” University of Alberta.

[47] Bastian, A. J., Zackowski, K. M., and Thach, W. T., 2000, “Cerebellar Ataxia: Torque Deficiency or Torque Mismatch between Joints?,” J. Neurophysiol., **83**, pp. 3019–3030.

[48] Kentel, B. B., King, M. A., and Mitchell, S. R., 2011, “Evaluation of a Subject-Specific, Torque-Driven Computer Simulation Model of One-Handed Tennis Backhand Ground Strokes,” J. Appl. Biomech., **27**(4), pp. 345–354.

[49] Kulig, K., Rao, S. S., Mulroy, S. J., Newsam, C. J., Gronley, J. K., Bontrager, E. L., and Perry, J., 1998, "Shoulder Joint Kinetics during the Push Phase of Wheelchair Propulsion," Clin. Orthop. Relat. Res., (234), pp. 132–143.

[50] Veeger, H. E. J., Van Der Woude, L. H. V, and Rozendal, R. H., 1991, "Load on the Upper Extremity in Manual Wheelchair Propulsion," J. Electromyography Kinesiol., **l**(4), pp. 270–280.

[51] Rodgers, M. M., and Tummarakota, S., 1998, "Three-Dimensional Dynamic Analysis of Wheelchair Propulsion," J. Appl. Biomech., **14**(1), pp. 80–92.

[52] Stephens, C. L., and Engsberg, J. R., 2010, "Comparison of Overground and Treadmill Propulsion Patterns of Manual Wheelchair Users with Tetraplegia," Disabil. Rehabil. Assist. Technol., **5**, pp. 420–427.

[53] Mason, B., Lenton, J., Leicht, C., and Goosey-Tolfrey, V., 2014, "A Physiological and Biomechanical Comparison of Over-Ground, Treadmill and Ergometer Wheelchair Propulsion," J. Sports Sci., **32**, pp. 78–91.