Modelling the stability and maneuverability of a manual wheelchair with adjustable seating

by

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B.Eng. (Sports and Mechanical), The University of Adelaide, 2014

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Abstract

Manual wheelchairs are generally designed with a fixed frame, which is not optimal for every situation. Spontaneous changes in seating configuration can ease transfers, increase participation in social activities, and extend reaching capabilities. These changes also shift the centre of gravity of the system, altering wheelchair dynamics. In this study, rigid body models of an adjustable manual wheelchair and test dummy were created to characterise changes to wheelchair stability and maneuverability for variations in backrest angle, seat angle, rear wheel position, user position, and user mass. Static stability was evaluated by the tip angle of the wheelchair on an adjustable slope, with maneuverability indicated by the ratio of weight on the rear wheels. Dynamic stability was assessed for the wheelchair rolling down an incline with a small bump. Both static and dynamic simulations were validated experimentally using motion capture of real wheelchair tips and falls. Overall, rear wheel position was the most influential wheelchair configuration parameter. Adjustments to the seat and backrest also had a significant impact on both static and dynamic stability. For wheelchairs with a more maneuverable (or 'tippy') initial configuration, dynamic seating changes could be used to increase stability as required.

Keywords:  Wheelchair stability; mobility devices; rigid body dynamics; simulation; motion capture; dynamic wheeled mobility
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Chapter 1. Background and theory

Wheelchairs are widely used as assistive devices; there are an estimated 1.6 million to 2.2 million users in the United States [1], and worldwide it is estimated that 65 million people require the use of a wheelchair [2]. The majority of these individuals (83%) use a manual wheelchair (as opposed to powered) [3].

For wheelchair users, the functionality and safety of their chair can drastically affect their quality of life [4]. Every year 3.3% of these users are involved in serious wheelchair-related accidents [5], some resulting in traumatic brain injury, bone fractures, or concussions [6]. Active manual wheelchair users are especially susceptible to tips and falls, with 61% of users having had at least one incident over a three year period, compared to 28% of power wheelchair users [7].

However, stability is only one aspect of wheelchair design [8]. Chronic overuse injuries (e.g. rotator cuff injuries, carpal tunnel syndrome, and median nerve damage) are also a widespread problem for manual wheelchair users [9]–[11]. Since manual wheelchair users are dependent on their upper limbs for most activities of daily life [12], propulsion efficiency and functionality are often of greater importance to the user than stability [13].

Maneuverability (e.g. propulsion efficiency) is increased by reducing resistive energy losses, which include rolling resistance, bearing resistance, tire scrub, and frame flexion [14]. For straight trajectories, rolling resistance is reduced by shifting the centre of mass towards the rear wheels [13], [15], but rear stability is increased by shifting the centre of mass forwards. Trade-offs are usually made between stability and maneuverability, with the 'optimal' wheelchair configuration varying depending on the activity being undertaken. To get the greatest benefit out of a wheelchair, the configuration therefore should change to suit the situation.
1.1. The dichotomy of wheelchair stability and maneuverability

Static stability is primarily a function of mass distribution, and is measured by the maximum incline that a wheelchair can be at rest without tipping in uphill, downhill, and lateral directions [16], [17]. If the horizontal position of the centre of mass of the user-wheelchair system remains between the axes of rotation of the wheelchair, the system will be stable at rest [18], [19] (Figure 1-1). For a wheelchair with unlocked wheels, these rotational axes correspond to the wheel axles. When brakes are applied, the axis of rotation becomes the point of contact between the wheels and ground.

![Diagram of wheelchair stability. The wheelchair system is stable when the CoG remains between the points of rotation of the wheelchair (a), and unstable when the CoG is shifted beyond one of these points (b). In this scenario the wheels are unlocked, so the axes of rotation correspond to the axles.](image)

The stability is increased when the centre of mass is shifted away from the point of rotation. This is directly related to the load distribution between the front and rear wheels. In general, manual wheelchairs are configured with most of the weight on the rear wheels [13]; backward tipping is therefore much more likely to occur in static situations than forward tipping. Variables that affect the tipping point include the position of the rear axle [19], [20], the addition of weights such as bags [19], the size of the wheels and casters.
[13], [20], the configuration of the seat and backrest [20], [21], and the mass and positioning of the user [20], [22]. Slopes and the condition of the ground, including any obstacles, also affect the wheelchair’s ability to stay upright [13], [22]. Similarly, dynamic stability refers to the ability of a wheelchair to stay upright while wheeling up or down slopes, which is also reliant on mass distribution [20]. However, dynamic stability also depends on inertia and contact characteristics.

In addition to being stable, manual wheelchairs must also be easy to push and maneuver, which reduces the risk of upper limb overuse and strain injuries [23]–[26]. Contrary to rear stability, straight motion maneuverability is improved by shifting the centre of mass towards the rear wheels to reduce rolling resistance [13], [14], [26]. Having more weight on the rear wheels also increases the ease of performing a wheelie; a necessary maneuver to wheel over everyday obstacles such as curbs [27].

The optimal configuration of a wheelchair therefore changes depending on the situation. For wheeling on slopes, the wheelchair needs to be stable enough so that it does not fall over. However, on level ground, the maneuverability of the wheelchair becomes more important. Once the minimum stability criteria is satisfied, a wheelchair should be optimized to improve maneuverability [13].

### 1.2. Manual wheelchair dynamic seating

The basic design of manual wheelchairs has remained substantially the same for the past century; the first model with metal tubing for the frame and a sling seat was invented in 1933 [28]. Since then, incremental improvements have been made in weight reduction and customization options, but few major changes have been made to the functionality (Figure 1-2). Ultralight wheelchairs are the most commonly prescribed and used manual wheelchair for active users [29], [30]. Most of these wheelchairs are manufactured with fixed frames, though some more recent models (PDG Mobility “Elevation” (Figure 1-3), ProActiv “Lift”) allow for “on-the-fly” dynamic seating, or the ability to change the seat and backrest configuration while in use [31].
Dynamic seating is mostly used for power and tilt wheelchairs. For power chairs, it has been shown that facilitating spontaneous changes to seating configuration improves independence by increasing the reach of the user, assists during transfers, and improves social interactions by elevating the wheelchair user closer to eye-level [21], [33]. These benefits would likely also extend to manual wheelchair users. Additionally, enabling the user to change their position throughout the day has physiological benefits by reducing neck strain in social situations [21], improving comfort, relieving pressure points, and altering baroreflex function [34].

Figure 1-2  Vintage wheelchair (left image, circa 1930’s [32]) compared to modern lightweight wheelchair (right image, 2005). Few changes have been made to manual wheelchair functionality.

Figure 1-3  Elevation model wheelchair by PDG Mobility. Gas springs under the seat allow for dynamic changes to the angle of the seat and backrest.
However, seat and backrest changes also shift the CoG of the system, affecting the stability and maneuverability of the wheelchair. Anecdotally, this allows users to optimize chair performance to specific use cases. For example, when travelling uphill, a wheelchair user could shift their backrest (and consequently CoG) forward, increasing stability [35]. On level ground, the user could then recline their backrest to increase maneuverability. However, there are currently no studies to quantify the sensitivity and extent of these effects.

1.3. Rigid body dynamics

Rigid body dynamics (RBD) is a method of studying the motion of interconnected bodies including the application of external forces. A key simplifying assumption, as suggested by the name, is the absence of deformation. This reduces the degrees of freedom, enabling problems to be solved efficiently without calculating the localized stresses and strains in each body. Segments are defined by a point mass, centre of mass location, and moments and products of inertia. Any external forces (e.g. gravity) are applied at the centre of mass. Interactions between segments are constrained by joints. Body surface geometry is only relevant for determining and calculating contact.

This type of analysis is commonly used when studying human motion; RBD methods have been used for biomechanical analyses as early as 1906 [36]. Since bodies are assumed to be rigid, the kinematics are defined by the positioning and orientation of a local coordinate system on each body with respect to a reference origin. For any point P on body i, the position of P in the reference space ($X_i$) is defined by

$$X_i = r_i + x_i$$  \hspace{1cm} (1-1)$$

where $r_i$ is the vector distance from reference origin to the local origin, and $x_i$ is the vector distance from the local origin to P (Figure 1-4). The vector $x_i$ is initially specified in a local coordinate system ($X_i, Y_i, Z_i$), and then redefined in the reference coordinate system using a rotational transformation matrix.
Taking the derivatives of position gives the following equations:

\[
\dot{\mathbf{X}}_i = \dot{\mathbf{r}}_i + \omega_i \times \mathbf{x}_i \quad (1-2)
\]

\[
\ddot{\mathbf{X}}_i = \ddot{\mathbf{r}}_i + \dot{\omega}_i \times \mathbf{x}_i + \mathbf{X}_i \times (\omega_i \times \mathbf{x}_i) \quad (1-3)
\]

where \(\dot{\mathbf{X}}_i\) is the velocity of \(P\) relative to the reference origin, \(\ddot{\mathbf{X}}_i\) is the linear acceleration, \(\omega_i\) is the angular velocity of \(i\) relative to the origin, and \(\dot{\omega}_i\) is the angular acceleration of \(i\). Using variations of these equations (including the kinematics of parent segments), the position, velocity, and acceleration of any segment in a linked system can be represented as a function of the preceding segments.

Deriving from classical mechanics, rigid body motion can be related to the applied forces and torques through the Newton-Euler equations of motion [37]

\[
m_i \ddot{\mathbf{r}}_i = \mathbf{F}_i \quad (1-4)
\]

\[
\mathbf{I}_i \cdot \ddot{\omega}_i + (\omega_i \times \mathbf{I}_i) \cdot \omega_i = \mathbf{T}_i \quad (1-5)
\]

where \(m_i\) is the mass of body \(i\), \(\ddot{\mathbf{r}}_i\) is the linear acceleration at the CoG, \(\mathbf{F}_i\) is the force vector, including any constraints, \(\mathbf{I}_i\) is the inertia tensor with respect to the CoG, and \(\mathbf{T}_i\) is the resultant torque. The equations of motion form a system of coupled non-linear second order differential equations [37], which can be represented by
\[ \ddot{q} = h(q, \dot{q}, t) \]  

where \( \ddot{q} \) gives the acceleration elements, \( h \) represents the equations of motion, \( q \) gives the generalized coordinates and joint degrees of freedom, and \( \dot{q} \) the velocities. The number of elements in column matrix \( q \) corresponds to the model degrees of freedom, which are determined by the joint type. These equations of motion can then be solved for position and velocity using numerical integration methods. Essentially, the processes for solving rigid body problems is an iterative method of applying forces, calculating the corresponding accelerations, integrating to give the velocity and position data, and then repeating (Figure 1-5).

**Figure 1-5**  
Simplified procedure for solving rigid body dynamic problems.

Due to the number of iterations required for an accurate model and the complexity of mathematically defining contact surfaces and properties, it is more efficient to use purpose-built software packages for solving multibody problems.
1.3.1. Madymo software

Madymo (TASS International, Livonia, MI) is a commonly used multi-body solver, often used for crash testing and accident reconstruction [38], [39]. It supports both rigid body dynamics and finite element methods [40], and has been used previously for studying wheelchair dynamics [41], [42]. In the software, segments are defined by a point mass, inertia about the CoG, surface geometry, and contact properties; constrained by joints to other segments. Given a set of initial conditions and any forces acting on the bodies (e.g. gravity), Madymo can then output all kinematics and contact forces within a system.

The kinematics and dynamics of the system are calculated in terms of the position and velocity at the preceding time point [37]. The default solver for Madymo uses an explicit Euler method. This expresses the velocity at \( t_{n+1} \) as a function of acceleration and the time step, \( t_s \). A similar equation is also used to work out the position at \( t_{n+1} \)

\[
\dot{q}_{n+1} = \dot{q}_n + t_s \ddot{q}_n \\
q_{n+1} = q_n + t_s \dot{q}_{n+1}
\]  

(1-7)  

(1-8)

For rigid body simulations, contact interaction is defined in Madymo by an elastic master-slave model [37]. In the elastic model, the slave surface (usually an ellipsoid) can penetrate the master surface (either a plane, cylinder, or ellipsoid), and a user defined force is applied as a function of the penetration. An iterative process is used to calculate the minimum penetration distance, and therefore contact force. This force-penetration characteristic can be separately defined for each surface or for the contact generally, and can be linear or nonlinear and include phenomena such as hysteresis, damping, and friction forces. Reducing the time step will improve the accuracy of calculations (particularly during impacts), but increase computational time. These iterative contact calculations and explicit kinematic equations allow the dynamics of the wheelchair, including any impacts, to be successfully modeled.
1.4. Objectives

The overall goal of this research was to assess the effect of dynamic seating changes on the stability and maneuverability of a manual wheelchair, and to determine which configuration parameters were most significant. The specific objectives included:

1. Determine the effects of seat angle, backrest angle, user position, user mass, and rear axle position on the static stability and maneuverability of an ultralight manual wheelchair.

2. Analyze the effect of these configuration changes on the dynamic stability and tip probability of a manual wheelchair rolling down a variable-angle slope and over small obstacles.
Chapter 2. Model specifications

A rigid body model of a manual wheelchair (including test dummy) was developed using Madymo to test the effects of on-the-fly wheelchair configuration adjustments (seat angle and backrest angle), fixed wheelchair configuration changes (rear wheel axle position), and user variables (user mass, user positioning) on the stability and maneuverability of an ultralight wheelchair. The simulation enabled more variable combinations to be systematically tested than experimentally achievable using a real wheelchair.

2.1.1. Geometric properties

The geometry of the wheelchair was taken from physical measurements of a manual wheelchair (first generation Elevation model, PDG Mobility, Vancouver, BC). These were double-checked using both a tape measure and 3D motion capture. Key measurements are illustrated in Figure 2-1, with blue spots indicating pivot points. An ISO wheelchair test dummy was also included in the model to simulate a user (Figure 2-2). This was comprised of three main parts; the torso, thigh, and legs. Revolute joints connected each of the components, allowing one degree of freedom.

The dummy/user was assumed to be fixed to the wheelchair seat, eliminating relative motion. The dummy was positioned with varied offset between the backrest and the base of the thigh segment for each simulation. The torso was fixed to the top of the backrest with a coupled universal and translational joint. This ensured that the thigh segment would move in synchrony with the wheelchair seat, and the torso segment would stay in contact with the backrest for all configuration changes. Fixing the distance to the top of the backrest mimicked the user changing their seated position while still using the backrest for support. Consequently, changes in user offset also changed the torso orientation.
Figure 2-1  Geometry of Elevation manual wheelchair.

Figure 2-2  Geometry of wheelchair test dummy used for experiments.
2.1.2. Mass and inertial properties

The wheelchair was considered as seven rigid bodies: the backrest, seat (including the gas springs used to change the seat and backrest angles), frame, two rear wheels, and two front wheels (Figure 2-3). Each physical component was weighed using digital scales, and the mass compared to a CAD model provided by PDG Mobility (Table 2-1). The CAD model was then adjusted to account for the discrepancies in Table 2-1, with mass added to each component as required. The adjusted CAD model was used to calculate the CoG (Figure 2-3) and inertia (Table 2-2) of each component.

![Figure 2-3 Centre of gravity for wheelchair frame and wheels.](image)

<table>
<thead>
<tr>
<th>Component</th>
<th>Measured mass (kg)</th>
<th>Mass from CAD model (kg)</th>
<th>Reason for discrepancy</th>
</tr>
</thead>
<tbody>
<tr>
<td>Front wheels (x2)</td>
<td>0.38</td>
<td>0.23</td>
<td>Different casters</td>
</tr>
<tr>
<td>Rear wheels (x2)</td>
<td>1.80</td>
<td>2.03</td>
<td>Different wheels</td>
</tr>
<tr>
<td>Seat (inc. gas springs)</td>
<td>3.21</td>
<td>2.78</td>
<td>CAD upholstery lighter</td>
</tr>
<tr>
<td>Backrest</td>
<td>1.24</td>
<td>0.64</td>
<td>CAD upholstery lighter</td>
</tr>
<tr>
<td>Wheelchair frame</td>
<td>3.19</td>
<td>3.00</td>
<td>Lighter brakes and rear axle used in CAD</td>
</tr>
</tbody>
</table>
Table 2-2  Mass and inertia of wheelchair frame and wheels.

<table>
<thead>
<tr>
<th>Component</th>
<th>Mass (kg)</th>
<th>Inertia: $I_{xx}$, $I_{yy}$, $I_{zz}$, $I_{xy}$, $I_{yz}$, $I_{xz}$ (kg.m$^2$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Front wheels (x2)</td>
<td>0.38</td>
<td>0.0005, 0.0009, 0.0005, 0, 0, 0</td>
</tr>
<tr>
<td>Rear wheels (x2)</td>
<td>1.80</td>
<td>0.0670, 0.1323, 0.0670, 0.0023, 0, 0</td>
</tr>
<tr>
<td>Seat (inc. gas springs)</td>
<td>3.21</td>
<td>0.0645, 0.0529, 0.0892, 0, 0, -0.0044</td>
</tr>
<tr>
<td>Backrest</td>
<td>1.24</td>
<td>0.0435, 0.0253, 0.0242, 0, 0, -0.0016</td>
</tr>
<tr>
<td>Wheelchair frame</td>
<td>3.19</td>
<td>0.1328, 0.1187, 0.2024, 0, 0, -0.0117</td>
</tr>
</tbody>
</table>

The torso (62.80 kg), thigh (42.16 kg), and legs (4.16 kg each) were weighed by deconstructing the test dummy, with differences of -0.07 kg, -0.53 kg, and 0.05 kg respectively compared to a CAD model of the same dummy. The CoG of each component (Figure 2-4) was calculated using a datum and scale. The differences in CoG location were [0.046 m, -0.013 m] for the torso and [0 m, 0.006 m] for the thigh compared to the CAD model. Inertia properties (Table 2-3) were taken about the COG for each component from the CAD model, and adjusted using the parallel axis theorem to account for the differences in COG positions in the actual dummy.

![Figure 2-4](image)

**Figure 2-4**  Dummy centres of gravity, as found experimentally. The torso CoG shown was (0.046 m, -0.013 m) different to the CAD model, and the thigh CoG was (0 m, 0.006 m) different.
Table 2-3  Mass and inertia of test dummy components.

<table>
<thead>
<tr>
<th>Component</th>
<th>Mass (kg)</th>
<th>Inertia: $I_{xx}$, $I_{yy}$, $I_{zz}$, $I_{xy}$, $I_{yz}$, $I_{xz}$ (kg.m²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Torso</td>
<td>62.80</td>
<td>0.9439, 0.6674, 1.3138, 0, 0, 0.0730</td>
</tr>
<tr>
<td>Thigh</td>
<td>42.16</td>
<td>0.9682, 0.5219, 1.2659, 0, 0, 0.0702</td>
</tr>
<tr>
<td>Legs (x2)</td>
<td>4.16</td>
<td>0.0182, 0.1022, 0.0871, 0, 0, 0.0163</td>
</tr>
</tbody>
</table>

2.1.3.  Wheel axial friction

Friction loads were found experimentally by spinning each of the wheels, using motion capture to record the rotation, and calculating the deceleration (and subsequently torque) due to friction. Each wheel was tested three times. The deceleration of the front wheels was approximately 50 times greater than the rear wheels (Table 2-4). Frictional loads, rather than coefficients, were used in the model as calculating the friction coefficients required accurate measurements of the pitch diameter and equivalent loads on the bearing. Attempts at calculating the coefficients using estimated bearing measurements resulted in a reduction in simulation accuracy.

The frictional torque was calculated from the mean front and rear wheel decelerations using the equation $\tau = I\alpha$. The torque on the rear wheels was greater since its inertia was much larger than that of the front wheels (Table 2-5). The values 0.000918 N/m and 0.00263 N/m were respectively used in the model as the front and rear axle frictional loads.

Table 2-4  Wheel decelerations due to friction.

<table>
<thead>
<tr>
<th></th>
<th>Mean acceleration (rad/s²)</th>
<th>Standard deviation (rad/s²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Left front wheel</td>
<td>-0.861</td>
<td>0.427</td>
</tr>
<tr>
<td>Right front wheel</td>
<td>-1.178</td>
<td>0.452</td>
</tr>
<tr>
<td>Left rear wheel</td>
<td>-0.0103</td>
<td>0.00421</td>
</tr>
<tr>
<td>Right rear wheel</td>
<td>-0.0295</td>
<td>0.00380</td>
</tr>
</tbody>
</table>

Table 2-5  Calculated average friction torque on front and rear axles

<table>
<thead>
<tr>
<th></th>
<th>Mean acceleration (rad/s²)</th>
<th>$I_{yy}$ (kg.m²)</th>
<th>Friction torque (N/m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Front wheels</td>
<td>-1.020</td>
<td>0.0009</td>
<td>0.000918</td>
</tr>
<tr>
<td>Rear wheels</td>
<td>-0.0199</td>
<td>0.1323</td>
<td>0.00263</td>
</tr>
</tbody>
</table>
2.1.4. Wheel contact properties

The contact properties of the wheels determined the behaviour of the wheelchair during, and immediately after, any impacts. For the rigid body model, the contact characteristics were defined by force-deflection loading and unloading curves. The loading values were found experimentally by measuring the deflection of the wheels as the load on the wheelchair was increased. Loads were measured with a scale under each wheel, and the displacement of the axles recorded using motion capture. Displacement was averaged between the left and right wheels, to give the values in Table 2-6.

Table 2-6 Average deflection of front and rear wheels, as measured experimentally using motion capture and scales.

<table>
<thead>
<tr>
<th>Weight on wheel (N)</th>
<th>Deflection of front wheel (m)</th>
<th>Deflection of rear wheel (m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>49</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>98</td>
<td>0.00086</td>
<td>0.00113</td>
</tr>
<tr>
<td>147</td>
<td>0.00155</td>
<td>0.00200</td>
</tr>
<tr>
<td>196</td>
<td>0.00208</td>
<td>0.00277</td>
</tr>
<tr>
<td>245</td>
<td>0.00249</td>
<td>0.00349</td>
</tr>
<tr>
<td>294</td>
<td>0.00297</td>
<td>0.00417</td>
</tr>
<tr>
<td>343</td>
<td>-</td>
<td>0.00477</td>
</tr>
<tr>
<td>392</td>
<td>-</td>
<td>0.00545</td>
</tr>
</tbody>
</table>

These values had to be extrapolated for the model as much greater forces would be experienced during impact with the bump. An offset was subtracted from the force such that the deflection was 0 mm for a load of 0 N, and equations fitted to the force-deflection curves. Exponential, power, linear, and quadratic models were all attempted, with the quadratic models having the best fit. The loading curve for the front wheels was \( F = 1.3E7 \times \delta^2 + 4.4E5 \times \delta \), which had an \( r^2 \) value of 0.9992. For the rear wheels, the equation was \( F = 3.8E6 \times \delta^2 + 4.3E4 \times \delta \), with \( r^2 = 0.9996 \). The force-deflection curves for the front and rear wheels (Table 2-7) were extrapolated using these equations.
Table 2-7  Force-deflection loading values used for front and rear wheels, from equations $F_{\text{front}} = 1.3 \times 10^7 \times \delta^2 + 4.4 \times 10^5 \times \delta$ and $F_{\text{rear}} = 3.8 \times 10^6 \times \delta^2 + 4.3 \times 10^4 \times \delta$.

<table>
<thead>
<tr>
<th>Deflection of wheel (m)</th>
<th>Front wheel load (N)</th>
<th>Rear wheel load (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>0.005</td>
<td>545</td>
<td>310</td>
</tr>
<tr>
<td>0.01</td>
<td>1740</td>
<td>810</td>
</tr>
<tr>
<td>0.015</td>
<td>3585</td>
<td>1500</td>
</tr>
<tr>
<td>0.02</td>
<td>6080</td>
<td>2380</td>
</tr>
<tr>
<td>0.025</td>
<td>9225</td>
<td>3450</td>
</tr>
<tr>
<td>0.03</td>
<td>13020</td>
<td>4710</td>
</tr>
<tr>
<td>0.035</td>
<td>-</td>
<td>6160</td>
</tr>
<tr>
<td>0.04</td>
<td>-</td>
<td>7800</td>
</tr>
<tr>
<td>0.045</td>
<td>-</td>
<td>9630</td>
</tr>
<tr>
<td>0.05</td>
<td>-</td>
<td>11650</td>
</tr>
</tbody>
</table>

The unloading force was defined as a ratio of the loading force. For the rear wheels, this was determined experimentally by releasing each wheel from a height (15 – 30 cm), and recording the height of consecutive bounces using motion capture. Drops were repeated 3 times for each wheel. The unloading/loading ratio was calculated for all bounces above 2 cm. Data below this height was discarded due to random errors in the motion capture and wheel alignment having comparatively greater effect at lower heights. The mean unloading/loading ratio for the rear wheels was 0.810 (σ = 0.027).

For the casters, the unloading/loading ratio was calculated using the kinematics of the entire wheelchair rolling down a slope, with the casters impacting the bump. The ratio was defined as the distance the wheelchair rolled back up the slope after impact (for the cases where the wheelchair was stopped by the bump), divided by the initial release distance. This different method was employed for the front wheels to take into account the properties of the entire caster assembly [43], and also because the casters had too much lateral movement when attempting the drop test to accurately calculate the rebound height. The mean unloading/loading ratio for the casters was 0.294 (σ = 0.145). This was less accurate than the rear wheel drop test, likely due to variations in configuration affecting the wheelchair dynamics. However, attempts at measuring the caster unloading forces using other methods (wheel drop test, using impact times to calculate the impulse from motion capture data) were even less accurate.
2.2. Model validation

The rigid body model of the wheelchair was validated by comparing it to real wheelchair tips and falls. Validation tests were completed for both static stability (ISO 7176-1) and dynamic stability (tip classifications when wheeling downhill over small bumps). Each type of validation test was completed for nine different seat and backrest configurations (Table 2-8). Stoppers were placed on the gas springs actuating the seat elevation and backrest to standardize the configurations and reduce frame flex.

Table 2-8  Wheelchair seat and backrest configurations used for validation tests. Seat angles ranged from 16.1° below horizontal to 13.6° above horizontal, and back angles ranged from vertical to a recline of 34.7°.

<table>
<thead>
<tr>
<th>Number</th>
<th>Seat angle</th>
<th>Backrest angle (from vertical)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>16.1° below horizontal</td>
<td>-1.0°</td>
</tr>
<tr>
<td>2</td>
<td></td>
<td>17.4°</td>
</tr>
<tr>
<td>3</td>
<td></td>
<td>34.7°</td>
</tr>
<tr>
<td>4</td>
<td>1.3° below horizontal</td>
<td>-1.0°</td>
</tr>
<tr>
<td>5</td>
<td></td>
<td>14.6°</td>
</tr>
<tr>
<td>6</td>
<td></td>
<td>29.0°</td>
</tr>
<tr>
<td>7</td>
<td>13.6° above horizontal</td>
<td>-1.0°</td>
</tr>
<tr>
<td>8</td>
<td></td>
<td>6.1°</td>
</tr>
<tr>
<td>9</td>
<td></td>
<td>17.6°</td>
</tr>
</tbody>
</table>

On completion of the physical experiments, an iterative process was used to calibrate aspects of the model. The impact of a broad range of variables were tested, including the mass, centre of mass position, and inertia of each segment, user positioning on the wheelchair, loading and unloading material properties of the wheels, and front and rear axial friction. Quantitative and qualitative comparisons were conducted to match the simulations to the experimental results, and each variable was optimized to increase model accuracy.
2.2.1. **Motion capture systems**

Kinematics of the physical wheelchair were captured using motion capture systems (Qualisys, Sweden and Vicon, UK). Optical motion capture is a common method of recording movement in 3D space, using reflective markers to denote points of interest [44]. Markers are placed on joints, with at least 3 markers on each segment to define its position. Each marker must always be in the view of at least two calibrated, infrared cameras, with additional cameras increasing accuracy and reducing marker swap.

For the experimental tests, a set of 26 markers was used (Figure 2-5). These defined the angles of the seat and backrest, relative positioning of the dummy, as well as the linear and angular position, velocity, and acceleration of the wheelchair. Additional markers (n=5) were also placed on the ground slope to give a reference for alignment and tip angles.

![Set of 26 markers used for recording wheelchair with 3D optical motion capture (mirrored on opposite side).](image2-5)
2.2.2. Static stability validation

The wheelchair static stability was tested as per ISO 7176-1: Wheelchairs -- Part 1: Determination of static stability [16]. This involved placing the wheelchair with locked wheels and dummy on a platform, and increasing the slope of the platform until the wheelchair started tipping. A block was placed at the bottom of the platform to prevent it from rolling off, and the motion capture system was used to determine when the uphill wheels started lifting off and the angle of the slope at that time. The stability was tested for nine configurations (Table 2-8), with each trial repeated three times in both forwards and backwards directions.

2.2.3. Dynamic validation

The model was dynamically validated by rolling the wheelchair down a ramp with a bump at the end (Figure 2-6), and comparing the simulation and experimental tip classifications (forwards tip, backwards tip, rolled over bump, or stopped by bump). Similar methods have been previously used for determining the effect of changes in seat position [20], caster diameter [45], or footrest elevation on dynamic stability [46]. During testing, the dummy was securely strapped to the wheelchair to minimise any relative movement between the dummy and chair.

Motion capture was used to record the relative position of the physical wheelchair and test dummy, which could then be used to calculate velocity and acceleration. This comparison was completed for nine different wheelchair configurations (Table 2-8), two ramp slopes (4.8° and 7.8°), three different bump heights (1.3 cm, 1.9 cm, and 3.2 cm), and at least 4 different speeds (up to 5.3 km/hr) for each variable combination. The speed was changed by releasing the wheelchair from varied distances from the bump. In total, 189 valid trials were completed.
2.2.4. Model calibration

Model accuracy was increased by calibrating less precise aspects of the model to match the experimental results. For the static results, the model was optimized by minimizing the RMSE between all experimental and simulation tip angles by making small variations to the CoG of each segment, as well as the position of the user relative to both the top and base of the wheelchair backrest. Each variable was changed by up to 3 cm in all directions, with a total of 34 variations of the static simulations performed. The optimized simulations had a RMSE of 1.82° for forward stability and 1.06° for backward stability.

The dynamic simulations were optimized by maximizing the number of tip classifications matching the experimental results through small changes to the inertia of segments, position of the user, loading and unloading material properties of the wheels, and axial friction. The segment CoG positions weren’t specifically calibrated for dynamic stability, as they were already optimized for the static model. Comparisons were made to an experimental subset of 66 trials, which were selected as the trials most likely to have discrepancies between the simulation and experiment. Inertia was individually varied from...
50% to 150% of the original values, wheel material loading force varied from 20% to 200% of the original force, wheel unloading from 10% to 100% of the loading force, axial friction from 5% to 250% of original, and user position by up to ±2.5 cm for the distance between both the top and base of the backrest and the user. A total of 172 variations of the dynamic simulation were tested, with the best simulation having 51/66 simulations matching the subset of experimental tip classifications (resulting in 168/189 simulations correct for the full set).
Chapter 3. Defining the stability limits of a manual wheelchair with adjustable seat and backrest

*Peer-reviewed conference paper accepted and presented at:*

RESNA 2017 Annual Conference, New Orleans, LA, USA

*Abstract*

Throughout the day, wheelchair users undertake a variety of mobility related activities of daily living. Adjustable “on the fly” seating changes allow users to adapt their wheelchair configuration to suit these different tasks. The objective of this study was to assess changes to wheelchair stability and maneuverability when adjustments are made to the wheelchair seat dump, backrest angle, rear axle position, and user position. This was performed by creating, validating, and testing a rigid body dynamic simulation of a wheelchair when positioned facing up or down slopes. The stability of the wheelchair was most affected by the position of the rear axle, but adjustments to the backrest and seat angles enabled relevant stability effects that could be used when wheeling in the community. For instance, adjustments to the backrest angle were shown to facilitate wheelchair stability on slopes over 20° steeper when compared to situations where the backrest remained in a fixed position. These findings provide support for the future use of adjustable seating in manual wheelchairs.
DEFINING THE STABILITY LIMITS OF A MANUAL WHEELCHAIR WITH ADJUSTABLE SEAT AND BACKREST

Louise Thomas\(^1,3\), Dr. Carolyn Sparrey\(^1,3\), and Dr. Jaimie Borisoff\(^2,3\)

\(^1\)Simon Fraser University. \(^2\)British Columbia Institute of Technology. \(^3\)International Collaboration on Repair Discoveries (ICORD). Vancouver, BC, Canada

BACKGROUND

Over 60% of active manual wheelchair users experienced falls due to instability over a three year period (Chen et al., 2011). Such incidents occasionally resulted in traumatic brain injury or bone fractures (Opalek et al., 2009). Clearly wheelchair stability is important, but it is not the only design consideration.

Manual wheelchairs must also be easy to push, i.e. maneuverable. This comes at the cost of reduced stability (Brubaker, 1986; Tomlinson, 2000). Maneuverability is improved by increasing the load on the rear wheels (normally done by moving the rear axle position forward). In contrast, the wheelchair is more stable when the load is distributed between the front and rear wheels. A compromise is found between the two objectives, with the optimal configuration dependant on the specific use case.

Adjustable "on the fly" or dynamic seating allows users to change their wheelchair seat configuration throughout the day (Borisoff and McPhail, 2011). Such changes have been identified by RESNA as important for health reasons, easing transfers, improving reach, and enhancing independence (Arva et al., 2009). Dynamic seating changes may move the centre of gravity of the system, affecting the maneuverability and stability of the wheelchair. However, to date little quantitative research has been conducted on the extent of these effects.

PURPOSE

This study aimed to determine the effects of seat dump, backrest angle, rear axle position, and user position (i.e. offset between a user’s hips and the backrest) on wheelchair stability and maneuverability, and to identify optimal seat configurations for sloped environments.

METHODS

Wheelchair simulation

To evaluate the stability of a wheelchair with a range of seat and backrest configurations, a rigid body dynamic model of a manual wheelchair was developed using MADYMO software (TASS International, Netherlands). A 250lb ISO test dummy was modelled, which is the design limit of many manual wheelchairs.

The geometry of the simulation was created using a CAD model of an ultralight manual wheelchair with dynamic seating (Elevation™, PDG Mobility, Canada) and validated using physical measurements of the same wheelchair. The wheelchair had 25” diameter wheels, 5” casters, a seat depth of 16”, and a seat width of 16”. The 250-lb test dummy was modeled to meet ISO 7176-11 standards.

Experimental testing

The wheelchair model was validated by comparing the stability of the simulation to that of a physical wheelchair. Static stability was tested in accordance with ISO 7176.1. An engine hoist was used to lift a platform with a block fixed at the bottom to prevent the wheelchair from rolling down (Figure 1). A 3D motion capture system (Qualisys, Sweden) determined both the time at which the wheelchair started tipping, and the corresponding angle of the ramp.

A full-factorial array of three seat and three backrest positions was tested. Configurations ranged from a seat angle of -13° and vertical
Figure 1: Static stability test setup, showing ramp lifted into a slope with engine hoist and wheelchair stopped from rolling with a block.

backrest (Figure 2b) to a seat angle of 16° and backrest reclined 35° (Figure 2c). Each configuration was tested three times for both forwards and backwards stability. In all cases, there was less than 2° difference between the simulated and experimental tip results.

Analysis

The simulations were run for full-factorial combinations of five backrest angles (-5° to 35°), seat dumps (-10° to 20°), rear axle positions (0cm to 20cm forward from the backrest), and offset distances between the user and the backrest (0cm to 8cm). The reaction forces on each wheel and the tip angles were recorded. The ratio of load on the rear wheels were calculated using MATLAB (Mathworks Inc., USA). This metric related the centre of gravity (CoG) of the wheelchair-user system to its performance, with a higher ratio indicating a more maneuverable but less stable configuration.

RESULTS

The backward stability of the wheelchair increased when the rear axles were positioned further back, when the backrest was more upright, and for greater user offsets (Figure 3). These changes moved the CoG of the system forward relative to the rear wheels. The magnitude of stability changes due to the backrest angle was also dependent on all other configuration variables. No other parameter had significant dependencies.

The angle of the backrest had the greatest effect on stability when the rear axle was moved forward, there was no user offset, and the seat was fully lowered. For a rear axle position of 10cm and no offset, a stability change of over 20° could be achieved just by changing the backrest (left middle panel, Figure 3).

In general, a lower rear seat corresponded to a small increase in stability, with greater
effects for more extreme seat angles. For certain more stable configurations, the effect of seat dump on stability was reversed.

On smaller ramp slopes the seat angle had little effect on wheelchair performance, while back angle had a greater effect (Figure 4). However, on steeper slopes and at more extreme settings, seat dump became a greater factor. While facing downhill the chair was most stable (less maneuverable) with the seat fully lowered and backrest reclined. When facing uphill, the stability was predominately affected by backrest angle. On steep uphill slopes (1:6), the wheelchair became unstable for any backrest angle greater than 20°.

**DISCUSSION**

A wheelchair should be stable enough so that it does not tip over in the user’s environment, but any more stability than necessary may impact performance and maneuverability (Tomlinson, 2000). Therefore, the load ratio on the rear wheels should be as high as possible without causing instability, (which always occurs at a ratio of 1).

For a fixed frame seating configuration, the stability and maneuverability are changed by the positioning of the rear axles and the posture of the user. Though the rear axle positioning has the greater effect of these two variables, our results show that user positioning is not inconsequential. Therefore, user posture should be considered when initially configuring a wheelchair and when

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**Figure 3:** Rear stability changes for different seat and backrest angles. Panels are grouped horizontally by rear axle position (5cm, 10cm, 15cm), and vertically by the user offset (0cm, 5cm).

**Figure 4:** Load ratios on the rear wheels for different seat and backrest configurations when the rear axle position is held constant at 10cm and there is no user offset. Each panel shows the wheelchair on a different slope. A higher ratio represents a greater percentage of the load on the drive wheels, indicating better maneuverability. However, a load ratio approaching 1 indicates instability.
instructing a new user. For example, if a user naturally slouches in a way that their lower back is 1 cm offset from the base of the backrest, the rear axles could be moved forward about 0.4 cm to maintain the stability and maneuverability of the wheelchair system (keeping all other variables constant). In addition, if a user moves forward in their seat the wheelchair will become more stable but less efficient for wheeling performance.

For wheelchairs with the capability, on-the-fly changes to seat and backrest configurations allow the wheelchair to be more tippable, and therefore more maneuverable, for a set initial configuration. For a fixed rear axle position and user offset, our simulations showed changes to the seat and backrest positions enable the wheelchair stability to vary by up to 22°. Backrest position has the greatest effect on stability and maneuverability; however, seat dump angle also affects the performance on steeper slopes.

Using dynamic changes to the wheelchair configuration, it is also possible to maintain the same front/rear wheel load distribution, and therefore maneuverability, of the wheelchair when it is on a slope. Figure 4 shows that when a wheelchair is set up to be stable on level ground, a rear load ratio of 0.75 (as used by Tomlinson, 2000) can be maintained for any slope between +9.5° and -9.5° by adjusting the seat and backrest angles. This is well within the wheelchair ramp standard of 1:12, or 4.8°. The backrest angle, rather than the seat dump, is the main enabler for this range.

Maintaining an optimal rear wheel load ratio could improve wheeling capabilities and safety in the community. For instance, when wheeling uphill, a backrest adjusted forward would provide support to users leaning into the slope; similarly, when traveling downhill, a backrest adjusted to more recline would provide the user with balanced trunk support and wheeling stability (Borisoff and McPhail, 2011; Hong et al., 2011), and obviate the need to be in a “wheelie”.

Limitations

A major limitation of the model is that it only looked at static stability, which does not completely reflect real world wheelchair use. As the user was modelled on an ISO dummy, it will also only represent a certain percentage of the wheelchair user population.

CONCLUSION

Rear axle position has the greatest effect on wheelchair stability and maneuverability. The backrest angle was the next most influential factor, and had significant dependencies on each of the other variables. By adjusting back and seat angles, stability changes of over 20° can be achieved.

ACKNOWLEDGMENTS

The authors would like to thank SFU students Tanuj Singla and Garrett Kryt for respectively helping with the creation and validation of the wheelchair model.

REFERENCES


Chapter 4. Quantifying the effects of “on the fly” seating configuration changes on manual wheelchair stability

Peer-reviewed conference paper accepted and presented at:

39th Annual International Conference of the IEEE Engineering in Medicine and Biology Society, Jeju Island, South Korea

Abstract

In general, manual wheelchairs are designed with a fixed frame, which is not optimal for every situation. Adjustable “on the fly” seating allow users to spontaneously adapt their wheelchair configuration to suit different tasks. These changes move the center of gravity (CoG) of the system, altering the wheelchair stability and maneuverability. To assess these changes, a computer simulation of a manual wheelchair was created with adjustable seat, backrest, rear axle position and user position, and validated with experimental testing. The stability of the wheelchair was most affected by the position of the rear axle, but adjustments to the backrest and seat angles also result in stability improvements that could be used when wheeling in the community. These findings describe the most influential parameters for wheelchair stability and maneuverability, as well as provide quantitative guidelines for the use of manual wheelchairs with on the fly adjustable seats.
Quantifying the Effects of On-the-Fly Changes of Seating Configuration on the Stability of a Manual Wheelchair

Louise Thomas, Student Member, EMBS, Dr. Jaimie Borisoff, and Dr. Carolyn J. Sparrey

Abstract— In general, manual wheelchairs are designed with a fixed frame, which is not optimal for every situation. Adjustable on the fly seating allow users to rapidly adapt their wheelchair configuration to suit different tasks. These changes move the center of gravity (CoG) of the system, altering the wheelchair stability and maneuverability. To assess these changes, a computer simulation of a manual wheelchair was created with adjustable seat, backrest, rear axle position and user position, and validated with experimental testing. The stability of the wheelchair was most affected by the position of the rear axle, but adjustments to the backrest and seat angles also result in stability improvements that could be used when wheeling in the community. These findings describe the most influential parameters for wheelchair stability and maneuverability, as well as provide quantitative guidelines for the use of manual wheelchairs with on the fly adjustable seats.

I. BACKGROUND

The majority of manual wheelchairs have a fixed frame, which does not allow for spontaneous changes in configuration after the initial setup [1]. However, user-initiated changes to seating (e.g. power wheelchairs with seat elevators, Elevation™ manual wheelchair) can help accomplish mobility related activities of daily living, such as transfers, participating in social activities, and extending reach [2]. These on the fly or dynamic seating adjustments allow users to change their wheelchair seat configuration throughout the day to better suit different activities [3].

Changes to seating shifts the center of gravity (CoG) of the system, potentially influencing wheelchair stability (defined as the tip angle of the wheelchair) and ease of wheeling (i.e. maneuverability). Stability and maneuverability are individually improved by shifting the system CoG in opposing directions [1], [4]. A trade-off is generally made between the two performance metrics, with the configuration of a typical fixed-frame wheelchair usually optimized for level ground wheeling. However, this is sub-optimal for situations such as traveling up or down slopes. Dynamic seating adjustments can lessen these compromises and improve task specific stability and maneuverability.

The backrest position, in particular, was identified as important and a target for on the fly adjustability to improve wheelchair use [3], [7], although its effects have not been fully studied. Other parameters may also be important, yet few studies have taken a holistic approach to examining multiple wheelchair configuration parameters at once.

II. PURPOSE

The aim of this study was to quantify the effects of seat angle, backrest angle, user position (i.e. “offset” between a user’s hips and the backrest), user mass, and rear axle position (Figure 1) on the stability and maneuverability of an ultralight manual wheelchair. Maneuverability was defined by the front/rear weight distribution of the wheelchair system where a greater percentage of weight on the rear wheels indicated the wheelchair was easier to push but less stable. A result of 100% signified a backwards tip, and 0% a forwards tip. The resulting equations enabled us to study the relationships between each of the six variables, including the relative effect of the ground slope angle.

III. METHODS

Wheelchair stability and maneuverability were evaluated using rigid body dynamic simulation (MADYMO TASS International, Netherlands). A range of ISO test dummies (25kg to 125kg, with increments of 25kg) represented the wheelchair user.

A. Wheelchair model development

The simulation was created using a CAD model of an ultralight manual wheelchair with dynamic seating (early

Figure 1: Wheelchair configuration changes used in simulation.
model Elevation, PDG Mobility, Vancouver, BC). The wheelchair had 24” diameter wheels, 5” casters, a seat depth of 16”, and a seat width of 16”. The frame, including the seat and backrest, was 7.62kg (Table I), with center of gravity (CoG) positioned 20cm behind the front axles and 37.3cm above the ground. The rear wheels were 1.80kg each, and the casters 0.38kg. Each wheel CoG was positioned at the axle.

<table>
<thead>
<tr>
<th>TABLE I. WHEELCHAIR MASS AND INERTIA</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mass (kg)</td>
</tr>
<tr>
<td>Front wheels (x2)</td>
</tr>
<tr>
<td>Rear wheels (x2)</td>
</tr>
<tr>
<td>Wheelchair frame</td>
</tr>
</tbody>
</table>

B. Experimental model validation

The wheelchair mass and horizontal position of the CoG were validated using scales under each of the four wheels. The vertical position of the CoG was calculated from the tipping point of wheelchair.

A 113kg test dummy was used for validation. The mass of the torso was 62.87kg, thigh 41.57kg, and each leg 4.11kg. The CoG of each component (Table II) was calculated using a pivot and scale, with measurements taken from the outermost point of the dummy hip when in a seated position.

<table>
<thead>
<tr>
<th>TABLE II. DUMMY MASS CoG</th>
</tr>
</thead>
<tbody>
<tr>
<td>Horizontal CoG (cm)</td>
</tr>
<tr>
<td>Torso</td>
</tr>
<tr>
<td>Thigh</td>
</tr>
<tr>
<td>Legs</td>
</tr>
</tbody>
</table>

The full dynamic model of the wheelchair was validated by comparing the stability of the simulation to that of the physical wheelchair. Static stability tests (Figure 2), as defined in ISO 7176.1, compared the angles at which the wheelchair tipped over for different configurations.

A 3D motion capture system (Qualisys, Sweden) was used to determine when the uphill wheels started lifting off and the angle of the ramp. The wheelchair marker set was comprised of 26 markers, with those in Figure 3 mirrored on the opposite side. Additional markers were placed on the ramp and ground to determine reference planes.

<table>
<thead>
<tr>
<th>Equation 1</th>
</tr>
</thead>
<tbody>
<tr>
<td>Backward tip angle = \frac{38.89 \pm 0.0824 \cdot S - (0.00531 \cdot S + 0.00092 \cdot M + 0.407)B - 1.14 \cdot R + 0.536 \cdot U + (0.00127 \cdot M + 0.191)M}{B}</td>
</tr>
</tbody>
</table>

where B is the backrest angle in degrees, S is the seat angle (degrees), R is the rear axle position (cm), U is the user offset (cm), and M is the user mass (kg). The equation correlation coefficient is R^2 = 0.953, and the RMSE is 1.27°.
Figure 4. Experimental and simulation derived tip angles showed excellent agreement for all tested wheelchair configurations for both forwards (a) and backwards (b) tips. Tip angle errors were greatest when the backrest was in its most upright position.

Figure 5. Interaction plots show slope angle affected the impact of most of the wheelchair configuration parameters except user offset on maneuverability. The effects of rear axle position and user offset on maneuverability was consistent throughout the trials. Values for each parameter ranged from seat angle (-10° to 20°), backrest angle (-5° to 35°), rear axle position (6cm to 20cm), user offset (0cm to 6cm), user mass (25kg to 125kg), and slope (-15° to 15°). Each panel shows the effect of the x-axis parameter on the rear wheel load ratio, with colored lines indicating the upper (blue) and lower (red) limits of the row parameters. The difference in slope between the blue and red lines indicates the interaction between the row and column variables, and parameter combinations with no interaction effects are shown by the faded, dotted lines.
The distribution of ground reaction force between the front and rear wheels was calculated as a metric of wheelchair stability and maneuverability. The slope of the ground was the greatest factor for determining the front/rear distribution of the weight, and had significant interaction effects with each of the other parameters apart from user offset (Eq. 2, Figure 5). There were also interaction effects for the backrest angle × seat angle and backrest × user mass.

\[
\text{Rear wheel load (\%) = } 40.4 + 0.302 \cdot S + \\
(0.00438 \cdot M + 0.420)B + 2.51 \cdot R - 1.29 \cdot U + \\
(-0.00160 \cdot M + 0.0932)M + (-0.0568 \cdot G - \\
0.0211 \cdot S + 0.00705 \cdot M + 0.0942 \cdot R + 0.711)G
\]

where \( G \) is the slope of the ground in degrees. The equation correlation is \( R^2 = 0.938 \) and RMSE = 5.40%

V. DISCUSSION

The stability and maneuverability of a wheelchair are dependent on a number of parameters, with some fixed during use (e.g., the position of the rear axles, and user mass), some situational or environmental (e.g., the ground slope angle), and some potentially adjustable by the user (e.g., the seat angle, backrest angle, and user offset). Adjustable parameters enable the wheelchair to adapt to situational variables.

A wheelchair should be stable enough to avoid tipping, but more stability than necessary reduces the performance and maneuverability of the wheelchair [4]. The rear axle position is an important parameter for optimizing initial wheelchair configuration. The wheelchair becomes less stable but more maneuverable as the rear axles are moved forward. On level ground, each 1cm shift forward of the rear axle increases the load on the rear wheels by 2.51% of the total system weight. On standard uphill 1:12 ramps (4.8° slope), the effect increases to 2.96% of the system weight. These changes equate to a 1.14° difference in rear stability for each 1cm change in rear axle position.

The user mass is positively related to the backward tip angle of the wheelchair, such that heavier users are more stable. Assuming body proportions similar to ISO wheelchair test dummies, each 10 kg increase in user mass approximately corresponds to a 3 to 4.5° increase in stability. The user mass also increases the effect of changing the backrest angle. Consequently, lighter users with adjustable backrests would need a greater angular adjustment to produce the same stability and maneuverability changes.

Dynamic seating changes, to the backrest in particular, are thought to enable the wheelchair to be more maneuverable on level ground, while also retaining the required stability for wheelchair on slopes [3], [7]. The results presented here confirm those assessments. For each degree change in backrest angle, the front/rear weight distribution changes by 0.86% of the system weight for heavier (100 kg) users, or 0.64% for lighter (50 kg) users. For uphill wheeling, rear stability can be increased by adjusting the backrest forward, with each degree backrest change corresponding to a 0.38-0.63° increase in stability. Therefore, for a backrest with 30° of adjustability, stability changes of up to 18.9° can occur. For traveling downhill, a reclined backrest would provide the user with balanced trunk support and negate the need for the user to perform a wheelie.

Changes to the seat height (by changing the seat angle) also have significant effects on stability and maneuverability. On level ground, each degree of seat depression increased the load on the rear wheels by 0.302% of the system. These changes can be used to negate the effects of user movements; for example, if the user changes their position 2 cm forward the same weight distribution can be maintained by lowering the seat by 5°. The effects of seat changes were more pronounced on downhill slopes, and had less of an effect when wheeling uphill.

VI. CONCLUSION

Simulated models for an ultralight manual wheelchair showed that the rear axle position and angle of the backrest were the most influential terms for wheelchair stability and maneuverability. Stability increases of up to 1.14° can be gained for each 1cm shift backwards of the rear axle, and up to 0.63° of stability can be gained by shifting the backrest 1° forward. The axle position should be configured to enable maximal maneuverability without tipping, and dynamic seating changes, particularly to the backrest, can therefore be used to increase task specific stability.

REFERENCES


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5.1. Abstract

Background

For people who use manual wheelchairs, tips and falls can result in serious injuries including bone fractures, concussions, and traumatic brain injury. We aimed to characterize how changes to manual wheelchair configuration affected dynamic tip probability while rolling down a slope with a small block at the end.

Methods

Rigid body dynamic models of a manual wheelchair and test dummy were created using multi-body software (Madymo, TASS International, Livonia, MI), and validated with 189 experiments. Dynamic stability was assessed for a range of seat angles (0 to 20° below horizontal), backrest angles (0 to 20°), rear wheel positions (0 to 20 cm from base of backrest), ground slopes (0 to 15°), bump heights (0 to 4 cm), wheelchair speeds (0 to 20 km/hr), user masses (50 to 115 kg), and user positions (0 to 10 cm from base of backrest). The tip classifications (forward tip, backward tip, rolled over bump, or stopped by bump) were investigated using a nominal logistic regression analysis.
Results

Faster wheelchair speeds significantly increased the probability of tipping either forward or backward rather than stopping, but also increased the probability of rolling over the bump (p<0.001). When the rear wheels were positioned forward, they increased the risk of a backward tip compared to all other outcomes (p<0.001), but also reduced the probability of being stopped by the bump (p<0.001 compared to forward tip, p<0.02 compared to rolling over). Reclining the backrest reduced the probability of a forward tip compared to all other outcomes (p<0.001), and lowering the seat increased the probability of either rolling over the bump or tipping backwards rather than tipping forward (p<0.001). In general, the wheelchair rolled over bumps <1.5cm, and forwards tipping was avoided by reducing the speed to 1 km/hr.

Conclusions

The probability of forward tipping, corresponding to the greatest risk of injury, was significantly reduced for decreased speeds, smaller bumps, and a reclined backrest. On-the-fly adjustments to the seat and backrest can reduce the probability of tipping and/or increase the probability of rolling over a bump. For wheelchairs with dynamic seat adjustability, when travelling downhill the seat should be lowered as far as possible to increase the likelihood of safely rolling over a bump.

Keywords

Wheelchair stability; mobility devices; rigid body dynamics; simulation; optical motion capture
5.2. Background

It is estimated that approximately 1% of the population in developed countries require the use of a wheelchair [47], [48]. Each year, 3.3% of people who use wheelchairs in the United States are involved in serious accidents [5], sometimes resulting in traumatic brain injury, bone fractures, and concussions [6]. For active manual wheelchair users, the risk is even higher. Over a three year period from January 2006 to December 2008, 60.7% of people using manual wheelchairs (n=56) reported tipping and falling at least once [7]. In the developed world, that equates to over 1.5 million manual wheelchair tips and falls every year.

The risk of a wheelchair tipping is related to its stability. Manual wheelchair static stability is defined by ISO 7176-1: 2014 as the angle at which a wheelchair and user tip over at rest [16]. However, there are currently no standards for determining manual wheelchair dynamic stability, that is, the risk of tipping while moving. Previous studies have considered manual wheelchair dynamic stability as the maximum speed that causes the wheelchair to stop rather than tip when rolling down a slope with a 5cm bump at the end (while varying seat position and caster diameter) [20], [45]. Yet this fails to consider a range of obstacles that wheelchair users encounter, some of which they would be able to safely roll over. The lack of more comprehensive dynamic stability studies is likely due to the difficulties of experimentally controlling variables such as wheeling speed in a safe environment, and the considerable number of variables that affect the stability of a wheelchair in use. Such difficulties can be minimized by integrating computer simulations, validated with controlled experiments.

Rigid body dynamics are commonly used for biomechanical analyses of injuries [49] and falls [50], and are characterized by equations relating the kinematics of a system to the corresponding kinetic forces [36]. A key simplifying assumption, as suggested by the name, is the absence of deformation. This reduces the degrees of freedom, enabling problems to be solved without needing to calculate the stresses and strains in each segment. Compared to finite element analysis, rigid body dynamic simulations are therefore much more efficient and computational inexpensive for analyzing large motions of bodies, making it an ideal method of studying wheelchair dynamics [41].
Our aim was to determine how fixed and spontaneous changes to a manual wheelchair configuration can affect the dynamic stability of the wheelchair rolling down a slope with a small bump at the end; a wheelchair skill that poses well-known safety concerns [51], [52]. Currently most manual wheelchairs are designed with a fixed frame [15], but more recent innovative designs allow users to adjust the seat height and backrest angle ‘on-the-fly’ to suit their purposes [31]. These changes affect static stability by changing the centre of gravity of the system [53]. However, these changes are also likely to affect the inertia of the system and the resulting dynamic stability. The purpose of this study was to determine the effects of on-the-fly wheelchair configuration adjustments (seat angle and backrest angle), fixed wheelchair configuration changes (rear wheel axle position), user variables (user mass, user positioning), and usage conditions (wheelchair velocity, slope of the ground, and bump height), on the dynamic tip probability of a wheelchair when moving down a slope.

5.3. Methods

5.3.1. Simulation

To quantify the effects of wheelchair configuration changes (Figure 5-1a) on downhill stability, a rigid body dynamic model of a wheelchair (Figure 5-1b, 16”x16”, first generation Elevation™ model with 24” rear wheels and 5” casters, PDG Mobility, Vancouver, BC) and ISO standard test dummy were developed using MADYMO software (TASS International, Livonia, MI) and placed on a sloped ramp with a small obstacle at the end. The model was used to simulate a manual wheelchair and user rolling down a slope, over a small bump. The initial velocity of the wheelchair was assigned to the chair center of mass when the wheelchair front axles were 10 cm from the bump. The chair was then released to freely roll down the incline and impact the bump.

The wheelchair model was defined by seven components: the seat, backrest, front wheels (x2), rear wheels (x2), and frame. The point mass and inertia of each of these components were taken from a CAD model provided by PDG Mobility (Table 5-1). The mass distribution of the CAD model had been previously validated by comparing the
tipping angle to that of the physical wheelchair for both forward and backward static stability [53]. The initial dummy measurements were taken from a CAD model of a 250lb test dummy, the same one used for the experimental validation. When varying the user mass, segment masses and CoGs were calculated from ISO 7176-11 [54]. The dummy was rigidly attached to the chair in the simulations to prevent relative motion between the dummy and the chair.

Figure 5-1  Diagram of wheelchair model. Variations were made to the wheelchair seat angle, backrest angle, rear axle position and user offset (a), as well as user mass, wheelchair speed, ground slope, and bump height in the simulations. The Madymo model is shown on the right (b).

Table 5-1  Mass and inertia for all wheelchair and dummy components included in model.

<table>
<thead>
<tr>
<th>Component</th>
<th>Mass (kg)</th>
<th>Inertia: $l_{xx}$, $l_{yy}$, $l_{zz}$, $l_{xy}$, $l_{yz}$, $l_{xz}$ (kg.m$^2$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Front wheels (x2)</td>
<td>0.38</td>
<td>0.0005, 0.0009, 0.0005, 0, 0, 0</td>
</tr>
<tr>
<td>Rear wheels (x2)</td>
<td>1.80</td>
<td>0.0670, 0.1323, 0.0670, 0.0023, 0, 0</td>
</tr>
<tr>
<td>Seat (inc. gas springs)</td>
<td>3.21</td>
<td>0.0645, 0.0529, 0.0892, 0, 0, -0.0044</td>
</tr>
<tr>
<td>Backrest</td>
<td>1.24</td>
<td>0.0435, 0.0253, 0.0242, 0, 0, -0.0016</td>
</tr>
<tr>
<td>Wheelchair frame</td>
<td>3.19</td>
<td>0.1328, 0.1187, 0.2024, 0, 0, -0.0117</td>
</tr>
<tr>
<td><strong>Total wheelchair mass</strong></td>
<td>12.00</td>
<td></td>
</tr>
<tr>
<td>Torso</td>
<td>62.80</td>
<td>0.9439, 0.6674, 1.3138, 0, 0, 0.0730</td>
</tr>
<tr>
<td>Thigh</td>
<td>42.16</td>
<td>0.9682, 0.5219, 1.2659, 0, 0, 0.0702</td>
</tr>
<tr>
<td>Legs (x2)</td>
<td>4.16</td>
<td>0.0182, 0.1022, 0.0871, 0, 0, 0.0163</td>
</tr>
<tr>
<td><strong>Total dummy mass</strong></td>
<td>113.28</td>
<td></td>
</tr>
</tbody>
</table>
The loading characteristics of the rear wheels and casters, which define the compression response during contact, were calculated by measuring the static deflection of each wheel under masses ranging from 0 – 40kg, and fitting a curve to the results. The unloading curve was defined as a percentage of the loading curve. For the rear wheels, this was calculated by measuring the reduction in bounce height of the wheels when they were dropped from heights of 15-30 cm, which was recorded and analyzed using motion capture. The mean unloading/loading ratio for the rear wheels was 0.810 (σ = 0.027). For the casters, the assembly was measured as a whole since the housing also has a significant effect on contact characteristics [43]. For the cases where the wheelchair was stopped by the bump during the experimental testing, the unloading percentage was calculated using the average distance the wheelchair rolled back up the slope after impact with the bump. Using this method, the mean unloading/loading ratio for the caster housing was 0.294 (σ = 0.145). The axial friction in the wheels were found experimentally by rotating each of the wheels and recording the deceleration using motion capture. The process was repeated three times for each wheel, with the frictional torque calculated from the wheels’ inertias and the resulting angular decelerations. The front wheels had a mean frictional torque of 0.000918 N/m, and the rear wheels 0.00263 N/m.

A sensitivity analysis was performed to determine the accuracy and sensitivity of various model inputs, including the inertia of each segment, wheel loading and unloading characteristics, axial frictions, and offsets between the user and the wheelchair backrest. Each parameter was altered independently at least 5 times for a set of simulations (66 trials), and evaluated by the number of simulation outcomes matching the experimental results. Additional simulations were run with variations to caster diameter (4”, 5” and 6”). These were separate from the rest of the sensitivity analysis as the wheelchair caster diameter was known, but changes to that diameter (if different casters were used) would likely have a significant impact on the probability of rolling over. For these simulations, all other wheelchair configuration variables were held constant (seat angle 10°, backrest angle 10°, rear wheels 10cm from the base of the backrest, a slope of 4.8°, user mass of 75kg, and no offset between the user and backrest), and the bump height was increased in increments of 1mm until the wheelchair no longer rolled over the bump. This procedure was followed for 3 different speeds (1, 3, and 5 km/h).
5.3.2. Experimental Validation

The model was validated by comparing simulations of the user and wheelchair rolling down a slope and into a bump to the kinematics of the physical wheelchair and test dummy, which was captured using 3D motion capture (Vicon, Oxford, UK). The dummy was strapped to the chair during testing to minimize relative motion between the dummy and the wheelchair, and padding was placed at the end of the ramp to minimize damage when forward tipping (Figure 5-2).

The wheelchair was tested for a full-factorial combination of nine seat and backrest configurations (Table 5-2), two ramp angles (4.8° [55] and 7.8°), three bump heights (1.3 cm, 1.9 cm, and 3.2 cm), and at least four speeds (up to 5.3 km/hr). This resulted in a total of 189 trials. Speed was varied by changing the release distance from the bump to the front wheels. Wheelchair kinematic behaviour was classified into four categories; rolled over bump, stopped by bump, tipped forwards, or tipped backwards. These classifications were used to compare the simulations to the physical experimental results.

**Figure 5-2** Experimental setup for testing wheelchair downhill stability.
Table 5-2  Wheelchair seat and backrest configurations used for validation tests. Seat angles ranged from 16.1° below horizontal to 13.6° above horizontal, and back angles ranged from vertical to a recline of 34.7°.

<table>
<thead>
<tr>
<th>Configuration type</th>
<th>Seat angle</th>
<th>Backrest angle</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>16.1° below horizontal</td>
<td>-1.0°</td>
</tr>
<tr>
<td>2</td>
<td>17.4°</td>
<td>17.4°</td>
</tr>
<tr>
<td>3</td>
<td>34.7°</td>
<td>34.7°</td>
</tr>
<tr>
<td>4</td>
<td>1.3° below horizontal</td>
<td>-1.0°</td>
</tr>
<tr>
<td>5</td>
<td>14.6°</td>
<td>14.6°</td>
</tr>
<tr>
<td>6</td>
<td>29.0°</td>
<td>29.0°</td>
</tr>
<tr>
<td>7</td>
<td>13.6° above horizontal</td>
<td>-1.0°</td>
</tr>
<tr>
<td>8</td>
<td>6.1°</td>
<td>6.1°</td>
</tr>
<tr>
<td>9</td>
<td>17.6°</td>
<td>17.6°</td>
</tr>
</tbody>
</table>

5.3.3. Analysis

Due to the number of variables, a Latin Hypercube experimental design [21], [22] was used to run 2000 variations of the validated model. The independent variables were the seat angle (0 to 20° below horizontal), backrest angle (0 to 20° from vertical), rear wheel position (0 to 20cm from base of backrest), slope of the ground (0 to 15°), bump height (0 to 4cm), and speed of the wheelchair (0 to 20 km/hr), user mass (50 to 115kg), and user offset from base of backrest (0 to 10cm). The geometry of the dummy model was constant for all user masses, and the CoG of the torso, thigh, and leg sections changed according to the wheelchair dummy standards [18]. The inertia values were scaled by the change in mass of each segment, and transformed using parallel axis theorem for changes in CoG locations. The observed dependent variable was the tip condition of the chair after impact with the bump. The final position of the wheelchair after impact with the bump was characterized as tipped forward, tipped backward, rolled over or stopped. A nominal logistic regression analysis was performed on the tip classifications using JMP software to determine the effects of the independent wheelchair configuration and user variables on the resulting tip behaviour (v13, SAS Institute, NC, USA). P-values less than 0.05 were considered significant, with results grouped by p < 0.001, p < 0.02 and p < 0.05.
5.4. Results

5.4.1. Simulation sensitivity analyses

The wheel unloading curve for the front casters had the greatest impact on model accuracy (Table 5-3). Rear wheel friction had an increased effect because, for the sensitivity analysis, speed was controlled by releasing the wheelchair from varied distances up the slope (the same as the experiment) and so axial friction affected impact speed. However, for the final simulations, an initial velocity was assigned to the wheelchair directly before hitting the bump, thus mitigating the effect of axial friction. User positioning also had a considerable effect on model sensitivity, highlighting the need to consider posture and user movement when configuring manual wheelchairs. For each inch increase in caster diameter, the maximum bump height that the wheelchair could successfully roll over increased by 2-3 mm (Table 5-4). For situations where the wheelchair could not roll over the bump, results differed depending on speed: for higher bumps, the wheelchair stopped when travelling at slower speeds (≤ 3 km/h), tipped forward when travelling at higher speeds (≥ 5 km/h). The effect of caster diameter on dynamic stability had been previously studied [45], and was not included in the main model as it is well known that larger diameter casters assist in rolling over higher bumps.

Table 5-3 Sensitivity of wheelchair model to set parameter changes.

<table>
<thead>
<tr>
<th>Parameter variation</th>
<th>Percentage change in correct simulations</th>
</tr>
</thead>
<tbody>
<tr>
<td>Torso inertia</td>
<td>50-150% of original</td>
</tr>
<tr>
<td>Thigh inertia</td>
<td>50-150% of original</td>
</tr>
<tr>
<td>Wheel unloading characteristics</td>
<td>50-150% of original</td>
</tr>
<tr>
<td>Wheel loading characteristics</td>
<td>50-150% of original</td>
</tr>
<tr>
<td>Rear wheel friction</td>
<td>50-150% of original</td>
</tr>
<tr>
<td>Caster wheel friction</td>
<td>50-150% of original</td>
</tr>
<tr>
<td>Offset between user and base of backrest</td>
<td>±1.5 cm from original</td>
</tr>
<tr>
<td>Offset between user and top of backrest</td>
<td>±1.5 cm from original</td>
</tr>
</tbody>
</table>
Table 5-4  Maximum bump height that the wheelchair rolled over for different caster diameters and speeds.

<table>
<thead>
<tr>
<th>Speed</th>
<th>Caster diameter</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>4 in</td>
</tr>
<tr>
<td>1 km/h</td>
<td>1.2 cm</td>
</tr>
<tr>
<td>3 km/h</td>
<td>1.7 cm</td>
</tr>
<tr>
<td>5 km/h</td>
<td>2.1 cm</td>
</tr>
</tbody>
</table>

5.4.2. Validation with experiments

Of the 189 validation simulations performed, 168 (89%) achieved the same tip classification as the experimental results (Table 5-5 and Table 5-6). The most common occurrence was rolling over the bump (84 out of 189 experimental trials), while tipping backwards was least likely to occur (Table 5-5). Backwards tipping was also the least accurately modelled case, with only 64.3% of simulations showing a backwards tip correctly. The simulations were most accurate for low bumps (1.27 cm) and least accurate when the bump height was 1.91 cm (Table 5-6). The majority of trials rolled over the low bump, and were stopped or tipped forward for the high (3.18 cm) bump. The tip outcomes were more variable for the mid-sized bump.

Table 5-5  Experimental vs. simulation confusion matrix. Shaded cells indicate misclassified trials. Rolling over the bump was the most common scenario, followed by being stopped by the bump.

<table>
<thead>
<tr>
<th>Experimental result</th>
<th>Simulation Result</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Forward tip</td>
</tr>
<tr>
<td>Forward tip</td>
<td>28</td>
</tr>
<tr>
<td>Backward tip</td>
<td>-</td>
</tr>
<tr>
<td>Rolled over</td>
<td>3</td>
</tr>
<tr>
<td>Stopped</td>
<td>-</td>
</tr>
</tbody>
</table>
Table 5-6  Comparison of simulation and experimental results, grouped by slope and bump height. For 189 trials, 88.9% of the simulations gave the same results as the experiment.

<table>
<thead>
<tr>
<th>Slope angle</th>
<th>Bump Height</th>
<th>Sims correct</th>
<th>Sims incorrect</th>
<th>Discrepancies Simulations</th>
<th>Discrepancies Experiments</th>
<th>Percentage correct</th>
</tr>
</thead>
<tbody>
<tr>
<td>7.8° 1.3 cm</td>
<td>24</td>
<td>1</td>
<td>Rolled</td>
<td>Backwards tip</td>
<td></td>
<td>96.0</td>
</tr>
<tr>
<td>7.8° 1.9 cm</td>
<td>22</td>
<td>5</td>
<td>3x forward tip</td>
<td>Rolled</td>
<td>Rolled</td>
<td>81.5</td>
</tr>
<tr>
<td>7.8° 3.2 cm</td>
<td>24</td>
<td>3</td>
<td>Stopped</td>
<td>Backward tip</td>
<td>Backward tip</td>
<td>88.9</td>
</tr>
<tr>
<td>4.8° 1.3 cm</td>
<td>33</td>
<td>1</td>
<td>Stopped</td>
<td>Forward tip</td>
<td>Forward tip</td>
<td>97.1</td>
</tr>
<tr>
<td>4.8° 1.9 cm</td>
<td>33</td>
<td>6</td>
<td>4x rolled</td>
<td>Stopped</td>
<td>Forward tip</td>
<td>84.6</td>
</tr>
<tr>
<td>4.8° 3.2 cm</td>
<td>32</td>
<td>5</td>
<td>3x stopped</td>
<td>Backward tip</td>
<td>Forward tip</td>
<td>86.5</td>
</tr>
</tbody>
</table>

At higher speeds, the front of the wheelchair often became airborne on impact with the bump (Figure 5-3). In some cases, this assisted in rolling over the bump, but also increased the probability of a backwards tip. Backwards tipping generally occurred when the wheelchair launched over the bump and the casters did not come down after clearing the bump. With the large test dummy, flex was observed in the wheelchair frame on impact with the bump, particularly to the backrest. For higher bumps, the wheelchair rolled over the bump using a rocking motion that popped the castors up (Figure 5-4).

Figure 5-3  Experimental sequence of events for wheelchair rolling over a medium bump (1.91 cm) at 3.92 km/h. (1) wheelchair released on slope, (2) casters impact bump, (3) the momentum of the wheelchair causes the casters to launch over bump, (4) rear wheels impact bump while casters are still in the air, (5) wheelchair continues rolling down slope.
5.4.3. Multinomial logistic model

The multinomial logistic parameter estimations (Table 5-7) showed bump height and speed were the most influential parameters on tip outcomes; rear wheel position and backrest angle had the greatest effect of the wheelchair configuration variables. Speed had a significant effect on all tip classifications, and the backrest angle had a significant effect (p<0.001) on all comparisons apart from ‘rolled vs stop’. Lowering the seat made the wheelchair significantly more likely to roll over the bump or tip backwards rather than tipping forwards.

The results of the logistic analysis, considering only linear terms, had a generalized R² value of 0.908 and a misclassification rate of 10.2% (Table 5-8). The majority of simulations (1093 of 2000) rolled over the bump, and rolling over was accurately predicted by the logistic model 94.9% of the time. Backwards tips were the most likely behaviour to be misclassified, with 54.0% of the simulations that tipped backwards misclassified as rolling over. Being stopped by the bump was the least likely scenario, occurring for 7.55% of simulations with a model prediction accuracy of 92.1%. With interaction terms included in the analysis, the generalized R² value increased to 0.942 and the misclassification rate was reduced to 7.6%. The most significant interaction effects were speed*bump height, rear wheel position*bump height, and speed*rear wheel position (Table 5-9).

![Figure 5-4](image)

Experimental sequence of events for wheelchair rolling over a high bump (3.18 cm) at 2.59 km/h. (1) wheelchair released to roll down slope, (2) casters impact bump and rear wheels lift, (3) the rear wheels return to the ground, but the momentum causes the casters to lift, (4) casters clear bump, (5) the rear wheels follow, also clearing the bump.
Table 5-7  Multinomial logistic parameter estimations, with standard errors in brackets. Bump height and wheelchair speed were the most influential parameters, with the rear wheel position and backrest angle having the greatest effect of the parameters directly relating to wheelchair configuration.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Forward tip vs Stop</th>
<th>Backward tip vs Stop</th>
<th>Rolled vs Stop</th>
<th>Backward vs Forward tip</th>
<th>Rolled vs Forward tip</th>
<th>Rolled vs Backward tip</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bump height (cm)</td>
<td>-0.127 (0.292)</td>
<td>-6.088*** (0.467)</td>
<td>-7.612*** (0.473)</td>
<td>-5.962*** (0.439)</td>
<td>-7.486*** (0.448)</td>
<td>-1.524*** (0.148)</td>
</tr>
<tr>
<td>Speed (km/hr)</td>
<td>2.311*** (0.235)</td>
<td>2.684*** (0.244)</td>
<td>2.851*** (0.244)</td>
<td>0.373*** (0.041)</td>
<td>0.540*** (0.040)</td>
<td>0.167*** (0.022)</td>
</tr>
<tr>
<td>Rear wheel position (cm)</td>
<td>0.170*** (0.038)</td>
<td>0.547*** (0.049)</td>
<td>0.128** (0.042)</td>
<td>0.377*** (0.035)</td>
<td>-0.042 (0.024)</td>
<td>-0.419*** (0.032)</td>
</tr>
<tr>
<td>Backrest angle (°)</td>
<td>-0.119*** (0.035)</td>
<td>0.258*** (0.042)</td>
<td>0.042 (0.038)</td>
<td>0.377*** (0.031)</td>
<td>0.160*** (0.026)</td>
<td>-0.216*** (0.022)</td>
</tr>
<tr>
<td>Slope (°)</td>
<td>0.532*** (0.065)</td>
<td>0.439*** (0.068)</td>
<td>0.493*** (0.067)</td>
<td>-0.094** (0.034)</td>
<td>-0.039 (0.030)</td>
<td>0.054* (0.025)</td>
</tr>
<tr>
<td>User offset (cm)</td>
<td>0.006 (0.073)</td>
<td>-0.375*** (0.084)</td>
<td>-0.170* (0.080)</td>
<td>-0.382*** (0.054)</td>
<td>-0.176*** (0.047)</td>
<td>0.205*** (0.039)</td>
</tr>
<tr>
<td>Seat angle (°)</td>
<td>-0.059 (0.035)</td>
<td>0.086 (0.041)</td>
<td>0.029 (0.039)</td>
<td>0.145*** (0.026)</td>
<td>0.087*** (0.023)</td>
<td>-0.058** (0.019)</td>
</tr>
<tr>
<td>User mass (kg)</td>
<td>0.025** (0.011)</td>
<td>-0.010 (0.012)</td>
<td>0.005 (0.012)</td>
<td>-0.035*** (0.008)</td>
<td>-0.020** (0.007)</td>
<td>0.015** (0.006)</td>
</tr>
</tbody>
</table>

*p<0.05 (lighter grey cells), **p<0.02 (darker grey cells), ***p<0.001 (black cells with white writing)

Table 5-8  Confusion matrix for the logit model showing a 10.2% misclassification rate when comparing the predicted result from the multinomial logistic analysis to the simulation results. Shaded cells indicate misclassified trials.

<table>
<thead>
<tr>
<th>Simulation result</th>
<th>Predicted Logit Model Result</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Forward tip</td>
</tr>
<tr>
<td>Forward tip</td>
<td>506</td>
</tr>
<tr>
<td>Backward tip</td>
<td>11</td>
</tr>
<tr>
<td>Rolled over</td>
<td>14</td>
</tr>
<tr>
<td>Stopped</td>
<td>8</td>
</tr>
</tbody>
</table>
Table 5-9 Interaction effect p-values for dynamic model. Significant interaction effects with p<0.001 were found for speed*bump height, rear wheel position*bump height, speed*rear wheel position, speed*slope, backrest angle*rear wheel position, slope*bump height, and user offset*speed.

<table>
<thead>
<tr>
<th></th>
<th>Bump height</th>
<th>Speed</th>
<th>Wheel pos.</th>
<th>Back angle</th>
<th>Slope</th>
<th>User offset</th>
<th>Seat angle</th>
<th>User mass</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bump height</td>
<td>&lt;0.0001</td>
<td>&lt;0.001</td>
<td>0.4644</td>
<td>0.0005</td>
<td>0.6938</td>
<td>0.0849</td>
<td>0.066</td>
<td></td>
</tr>
<tr>
<td>Speed</td>
<td>&lt;0.0001</td>
<td>&lt;0.001</td>
<td>0.4743</td>
<td>0.0001</td>
<td>0.0009</td>
<td>0.2798</td>
<td>0.2597</td>
<td></td>
</tr>
<tr>
<td>Wheel position</td>
<td>&lt;0.0001</td>
<td>&lt;0.001</td>
<td>0.0003</td>
<td>0.0289</td>
<td>0.6383</td>
<td>0.0435</td>
<td>0.0581</td>
<td></td>
</tr>
<tr>
<td>Backrest angle</td>
<td>0.4644</td>
<td>0.4743</td>
<td>0.0003</td>
<td>0.0112</td>
<td>0.7064</td>
<td>0.339</td>
<td>0.047</td>
<td></td>
</tr>
<tr>
<td>Slope</td>
<td>0.0005</td>
<td>0.0001</td>
<td>0.0289</td>
<td>0.0112</td>
<td>0.0934</td>
<td>0.6612</td>
<td>0.2141</td>
<td></td>
</tr>
<tr>
<td>User offset</td>
<td>0.6938</td>
<td>0.0009</td>
<td>0.6383</td>
<td>0.7064</td>
<td>0.0934</td>
<td>0.0864</td>
<td>0.0051</td>
<td></td>
</tr>
<tr>
<td>Seat angle</td>
<td>0.0849</td>
<td>0.2798</td>
<td>0.0435</td>
<td>0.339</td>
<td>0.6612</td>
<td>0.0864</td>
<td>0.2391</td>
<td></td>
</tr>
<tr>
<td>User mass</td>
<td>0.066</td>
<td>0.2597</td>
<td>0.0581</td>
<td>0.047</td>
<td>0.2141</td>
<td>0.0051</td>
<td>0.2391</td>
<td></td>
</tr>
</tbody>
</table>

Black cells with white writing = p<0.001, darker grey cells = p<0.02, and lighter grey = p<0.05.

To explore the effects of on-the-fly adjustability on downhill stability, the expected wheelchair tip classifications from the logit model were plotted for different backrest angles, seat angles, speeds, and bump heights (Figure 5-5). Rear wheel position was held constant at 10cm, slope was set to 4.8 degrees (equivalent to 1:12, a wheelchair standard for maximum ramp inclines), user mass set to 75 kg, and the user was positioned with no offset to the backrest. The plots show that bumps of 1.5cm or less are unlikely to be an issue for manual wheelchairs to roll over, and forwards tipping over higher bumps can be avoided by reducing speed to 1km/hr. Bumps of 2.5cm and greater could generally not be rolled over regardless of variable configurations (except at higher speeds). For speeds of 1 km/h and 3 km/h, lowering the seat height moved the expected outcomes of forward tipping or stopping to the safer results of stopping or rolling over. Similar results are shown for backrest recline, where a reclined backrest increases the likelihood of stopping rather than tipping forward and, for bumps <2 cm, increases the probability of rolling over the bump instead of stopping. However, under greater backrest angle conditions, backwards tips are also possible.
Figure 5-5  Expected wheelchair behaviour after rolling into/over a bump with respect to backrest angle and seat angle. Panels are grouped by speed and bump height.
5.5. Discussion

Manual wheelchairs are an invaluable mobility aid for those that require them, but can pose a risk of tipping when traveling on sloped and uneven surfaces. Of manual wheelchair users that have experienced a fall, it is reported that 46.3% of falls were in the forward direction [56], which is also the tip direction most likely to result in a serious injury [18]. The top three self-reported causes of wheelchair related accidents are inexperience, uneven surfaces, and obstacles [7]. This study explored the stability of a manual wheelchair when wheeling down a slope and into a small bump using a combination of experiments and simulations. A comprehensive map of the effects of wheelchair configuration, user position, and user mass on tip risk when wheeling downhill was determined. Bump height, wheeling speed and rear wheel position were the most significant determinants of tipping probability, while on-the-fly adjustments to the seat height and backrest angle could also favorably change the outcome.

While standards exist for static stability [16], there are currently no standards for manual wheelchair dynamic stability. Previous studies considered dynamic stability rolling down a slope with a large (5cm) bump at the bottom [20], [45], [46], where the outcome was either a stop or forwards tip. One such study showed that by moving the horizontal position of the seat (and therefore CoG) forward, the speed required to cause a forward tip decreases [20]. This agrees with our results, which show that forward movement of the CoG (by reducing the backrest angle or increasing user offset) increases the risk of a forward tip (Table 5-7).

A forward tip is the worst case scenario, and most likely to result in injuries requiring medical attention [18]. The parameters that had the greatest effect on forward tip probability were bump height, speed, and rear wheel position. As the bump height increased, the speed required to roll over (assuming no torso movement) also increased. However, increasing speed also increased the risk of tipping rather than stopping. For lower bumps (≤2cm), speed could be used to assist in overcoming obstacles, but this
increases the risk of causing greater injury if a tip does occur. These results agree with others’ work and highlight the importance of training wheelchair users to effectively navigate obstacles during downhill wheeling, including by adjusting their wheeling speed for different obstacles [51]. Lowering the seat significantly increased the probability of rolling over the bump, and reduced the risk of a forward tip. It is therefore recommended to lower the seat as far as possible, if the wheelchair includes this function, for downhill wheeling.

When wheeling downhill, the ideal outcome is for the wheelchair to roll over the bump. This occurred for 95% of simulations with a bump lower than 1cm and backrest angle less than 20 degrees. However, if rolling over is not possible, it is much better for the wheelchair to be stopped by the bump rather than tip. In general, encountering a bump at 1 km/h (slow speed) allowed the user to safely stop without tipping. On level ground, comfortable propulsion speeds range from 3.7 km/h [57] to 4.6 km/h [58], with downhill wheeling sometimes faster. Common obstacles encountered when wheeling downhill include potholes, rocks, and differences in pavement height, most of which are unlikely to be more than 2cm in height. Wheelchair users can overcome higher obstacles such as curbs using torso rotation and controlled wheelies [27]. A similar type of movement was shown in Figure 5-4, where the wheelchair pitched back and forth over the high bump. User movements (such as balancing in a wheelie) could be used in addition to configuration changes and speed to further improve downhill stability over bumps.

Reclining the backrest increased the probability of rolling over the bump or stopping rather than tipping forward. This did increase the risk of a backwards tip, but this was the least common outcome (5.8% of experiments and 10.6% of final simulations), was only an issue at very high backrest angles typically not used during active wheeling, and has been shown to be less dangerous than a tip forward [18]. The angle of the backrest can be the difference between a forward tip, being stopped by the bump, rolling over, or tipping backward (Figure 5-5). A reclined backrest assists in maneuvering over bumps, but once the angle is more than 20 degrees there becomes a risk of tipping backward. This is similar to the static stability of the wheelchair, where a more reclined backrest enables the wheelchair to be more maneuverable, but less stable [53]. For wheelchairs
without adjustable backrests, generally the user will have to perform a wheelie to go down steep inclines [51], which many users find unsafe or unable to perform [59]; reclining the backrest may negate the need to do this. Nonetheless, users with fixed frames may also benefit from knowing the quantified effects of backrest and seat angle on dynamic downhill stability, as it could assist in selecting the correct configuration for daily usage conditions. Depending on individual stability requirements, adjusted results from this study could be used to create guidelines to inform users and therapists of customized stability limits and maneuverability changes resulting from different wheelchair configurations.

User positioning has been previously shown to have a significant effect on stability [60]. When the user’s pelvis was positioned at an offset from the backrest, the probability of tipping backward was significantly reduced in comparison to all other behaviours. However, the probability of tipping forward rather than rolling over was also increased. For users that sit with their hips forward from the base of the seat, configuring the wheelchair with the rear wheels further forward can permanently reverse the ensuing stability effects, or a reclined backrest could be used to temporarily adjust the stability as needed. As suggested by the Wheelchair Skills Training Program Manual, users should therefore be encouraged to reposition themselves as far back in the wheelchair as possible during downhill wheeling [51] to reduce the risk of a forward tip.

In general, configuration changes that made the wheelchair more likely to roll over the bump (lowering the seat, reclining the backrest, moving the rear wheels forward) did so by shifting the system CoG towards the rear axles. On level ground, backward shifts in the CoG position also increase maneuverability [61]. The position of the rear wheels had the greatest effect on tip response at slower speeds and when the bump was between 1.5 and 2.5 cm. For these cases, the outcome was less predictable and the position of the rear wheels could be the deciding factor of whether the wheelchair tipped or rolled over. Moving the rear wheels further forward made the chair more likely to tip backwards; interestingly, it also slightly increased the probability of rolling over the bump or tipping forwards rather than being stopped.
Rolling over probability was likely increased due to shifting the CoG towards the rear wheels, which reduced the load on the front wheels, making it easier for them to clear the bump. The increase in forward tipping probability may be owing to the weight of the rear wheels shifting the CoG forwards in relation to the front wheels. The effect of wheel position on dynamic rolling stability highlights the need for therapists and industry professionals to properly configure the wheelchair for a particular user. These results relate to previous research on manual wheelchair static stability, which showed that forward movements of the rear wheels reduced stability, but increased maneuverability for a straight trajectory (defined as minimizing rolling resistance) [53]. It also suggests an opportunity for future designs offering a rear axle (or CoG) ‘shift on the fly’ adjustment capability that could significantly improve wheeling stability on slopes.

When modelling the dynamics of a manual wheelchair, it is also important to take resistive forces into account [14]. For situations where the user is pushing the chair (i.e. most dynamic cases apart from wheeling downhill), reducing resistance is important for minimizing the risk of upper limb overuse and strain injuries [23]–[26]. Therefore, to fully explore the maneuverability of a manual wheelchair, the ease of pushing should also be considered. Increasing the load on the rear wheels reduces rolling resistance for straight trajectories [13], such as the modelled case of wheeling downhill, but does so at the cost of reducing rear stability [53], [61]. Furthermore, this increase in rear wheel loading corresponds to an increase in resistive forces due to turning [14]. Dynamic wheelchair performance is likely a balance between stability, rolling resistance, and turning resistance, with the optimal configuration dependent on task specific requirements. Thus, the ability to change wheelchair configurations ‘on the fly’ to emphasize different performance advantages may be beneficial to wheelchair users.
5.5.1. **Strengths and limitations**

Computational models are an efficient method for studying wheelchair dynamics, however they are limited by model input accuracy [41]. The use of passive dummy models is a particular limitation, as it disregards any active movements of the user. For the case of rolling down a slope this is not a major issue as users are advised to maintain their weight towards the rear of the wheelchair when descending [51]. However, when navigating obstacles and for other situations where the user actively changes their position, future models will need to be modified to simulate user activity. Since the mass of the user represents the majority of the system mass, dummy stature is another limitation. The ISO dummies used represent the average stature of a wheelchair user [54], but individual variations may affect model accuracy by changing the mass distribution and therefore the inertial characteristics and centre of mass of the user.

The modeling of the wheels is another point of potential inaccuracy in the model, as rigid body models are unable to fully capture the dynamics of collisions [62]. This is demonstrated by the increased sensitivity of the model to the wheel unloading characteristics (Table 5-3). Since some deformation occurs on impact with the bump, finite-element methods would improve the accuracy of the tire contact calculations. Including tire deformation would also allow the rolling resistance of the wheelchair to be more accurately modelled. However, using finite element analysis in the model would greatly increase computational time and limit the number of simulations that could feasibly be run.

The measured physical properties of the wheelchair were another possible source of error in the model. In particular, the accuracy of the wheel contact characteristics and the axial friction were limited by the methods used to measure them. Since the loading of the wheels were measured statically, they would not precisely match the dynamic loading characteristics during a collision. Measuring the dynamic loading of the wheels was outside the scope of this study. Using an unloaded axial friction load was also a limitation, but provided a reasonable approximation. Estimating friction coefficients from the deceleration of the wheels resulted in a less accurate model than using the friction loads from the unloaded wheel.
5.6. Conclusion

A combination of skills training and dynamic wheelchair adjustability could greatly improve user safety when wheeling over obstacles. The most significant factors for downhill wheeling stability were bump height, speed, and rear wheel position. On-the-fly adjustments to the seat and backrest could be used in certain situations to reduce the probability of tipping and/or increase the probability of rolling over a bump. The quantified downhill rolling stability results could also be used to guide the configuration of fixed-frame wheelchairs to define operating limits. For wheelchairs with dynamic seat and backrest adjustability, when travelling downhill the seat should be lowered as far as possible to increase the likelihood of safely rolling over a bump. Reclining the backrest may also help in overcoming obstacles, but should be adjusted with caution as reclining far will also increase the probability of a backwards tip.
Chapter 6. Discussion and conclusions

6.1. Manual wheelchair stability and maneuverability

Manual wheelchairs need to be stable enough to not tip over, but increases in stability reduce maneuverability. Both metrics are affected by the system CoG, which can be altered by changing wheelchair configurations. Experimental and computational methods were used to quantify the effects of possible on-the-fly wheelchair configuration adjustments (seat angle and backrest angle), fixed wheelchair configuration changes (rear wheel axle position), and user variables (user mass and user positioning) on the static stability and straight motion maneuverability of a manual wheelchair. The effects of these variables, as well as usage conditions (wheelchair velocity, slope of the ground, and bump height), on the dynamic tip probability of a wheelchair when moving down a slope were also explored.

For an ultralight manual wheelchair, the rear axle position was the most influential term for wheelchair stability and maneuverability. Rear stability (defined as the angle at which tipping occurred when facing uphill) was increased by shifting the rear wheels backward, with stability increases of up to $1.14^\circ$ gained for each 1cm shift of the rear axles. Maneuverability was defined by the load percentage on the rear wheels, with a higher rear wheel load ratio corresponding to a reduction in rolling resistance. This definition of maneuverability is applicable for wheeling in a straight trajectory; for turning maneuverability, tire scrub would also need to be considered [14]. Conversely, straight motion maneuverability was increased by shifting the rear axles forward. Forward shifts to the rear axles also increased the likelihood of rolling over a bump when wheeling downhill.
For a fixed rear axle position and user offset, dynamic changes to the seat and backrest positions enabled stability changes of up to 22°. These configuration changes also made it possible to maintain the same front/rear wheel load distribution, and therefore wheelchair maneuverability, on slopes ranging from +9.5° to -9.5°. These slopes are greater than the 4.8° standard for accessible design [55], but plausible for wheelchair users to encounter in substandard environments. For each degree change in backrest angle, the front/rear weight distribution changed by 0.86% of the system weight for heavier (100 kg) users, and 0.64% for lighter (50 kg) users. For uphill wheeling, rear stability was increased by adjusting the backrest forward, with each degree backrest change corresponding to a 0.38-0.63° increase in stability. For a more maneuverable (and therefore tippy) initial configuration, dynamic seating changes, particularly to the backrest, can therefore be used to increase stability as needed.

The most significant factors for downhill wheeling stability were bump height and speed, with rear wheel position and backrest angle having the greatest effect out of the wheelchair configuration parameters. For wheelchairs with dynamic seat adjustability, when travelling downhill the seat should be lowered as far as possible to increase the likelihood of safely rolling over a bump and reduce the risk of a forward tip. Reclining the backrest may also help in overcoming obstacles and reduce the risk of a forward tip, but should be adjusted with caution as reclining too far will also increase the probability of a backwards tip. In general, encountering a bump at 1 km/h (slow speed) allowed the user to safely stop without tipping.

The downhill rolling stability results reinforce wheelchair configurations required for improving forward static stability, but contradict the backward static stability results [53], [61]. Stability and maneuverability are highly dependant on usage circumstances. Static backward stability is increased by shifting the system CoG forward, but this decreases straight-line maneuverability, forward static stability, and the ability of the wheelchair to roll over obstacles. For wheeling downhill and in straight trajectories on level ground, to improve safety and ease of wheeling, the chair should be configured to increase forward stability. However, for wheeling uphill the wheelchair should likely be configured to maximize backward stability.
Maneuverability was defined by reducing rolling resistance, as indicated by ratio of load on the rear wheels. This definition only relates to straight trajectories, such as the modelled case of wheeling down a slope. However, during turns, tire scrub introduces an additional resistive force [14]. As the load on the rear wheels increases, straight trajectory maneuverability increases, but turning maneuverability decreases [14]. Therefore, both the optimal maneuverability and stability depend on wheelchair usage conditions. For future simulations, tire scrub should be considered when modeling the wheelchair maneuverability in situations involving turns, which may reduce compromises between the two objectives.

6.2. Significance

This study quantifies the specific effects of wheelchair configuration, user parameters and environmental variables on stability and maneuverability. The results highlight the magnitude of the effect of each variable on stability and provide guidance for numerous stakeholders, including users, therapists, equipment designers, and manufacturers. An increased knowledge of the factors affecting wheelchair performance could help clinicians and occupational therapists when configuring wheelchairs for individual patients, subsequently improving the quality of life for manual wheelchair users and reducing tip and fall risk. The dynamic stability results also highlight the need to consider wheelchair stability and mobility over obstacles when prescribing and configuring wheelchairs, and training wheelchair users. Wheelchairs with dynamic seating functionality would not only allow users to adapt to changes in environment (e.g. wheeling on slopes), but also to changes in individual stability and maneuverability requirements. Key outcomes of this study include:

1. On-the-fly adjustments to the seat and backrest were shown to have a significant effect on static stability, maneuverability, and downhill dynamic tip probability of an ultralight manual wheelchair. For wheelchairs with this capability, configuring the chair for maximal rolling efficiency on level ground (the most common dynamic activity) no longer needs to compromise downhill rolling stability, since the stability can be increased as needed for slopes.
2. When wheeling downhill, both lowering the seat and reclining the backrest increased the probability of rolling over a bump, and decreased the probability of tipping forward. On-the-fly seating configuration changes could therefore improve wheelchair safety when travelling downhill, and negate the need for performing wheelies.

3. The substantial effect of rear wheel position on both static and dynamic stability highlights the importance of properly configuring wheelchairs for each particular user. It also suggests an opportunity for the future development of a wheelchair with on-the-fly rear axle adjustment capabilities, which could significantly improve wheeling stability on slopes.

The results indicate that on-the-fly adjustable seating has the potential to greatly improve wheelchair performance and situational specific stability, though currently the technology is not widely used. Wheelchairs such as PDG Mobility’s Elevation model are more expensive than most ultralight wheelchairs [63], which creates a barrier to purchasing. One potential solution for this, at least in countries such as Canada with substantial public healthcare funding, would be to further investigate the potential reduction in hospital spending resulting from the ability to alter wheelchair stability and manoeuvrability (e.g. overuse injuries and tipping); healthcare savings may justify more expensive mobility solutions. For wheelchair users without dynamically adjustable seating, the results from this study can still provide value by quantifying the relative effects of different configuration parameters. Simulations could be used to define stability bounds for a fixed configuration wheelchair, assisting both users and therapists in selecting the correct wheelchair configuration for individual stability and maneuverability requirements.

6.3. Model challenges and limitations

The model accuracy is partially limited by the accuracy of the wheel contact properties. Rigid body models do not simulate contact-related deformation [62], and negating wheel deformation causes the absence of some dynamic resistive forces. To accurately assess turning maneuverability, tire scrub also needs to be accounted for [14].
These issues could be addressed using a finite element tire model, but at the cost of increasing computational time. Also, since the force-displacement curves for the tires were statically determined, the loading forces for the tires may be lacking accuracy for dynamic collisions. Tire contact characteristics would also be affected by tire inflation pressure.

Calibrating the dynamic simulations was difficult due to the number of variables affecting the tip probability, with many of the variables having interaction effects. Testing the full factorial effects of every input in the model was unfeasible due to time and data storage constraints. Optimization methods were attempted, such as implementing a genetic algorithm and using black-box optimization software (OASIS, Empower Operations, Vancouver), but these were unable to converge on an optimal solution. Part of the reason why the optimization was unsuccessful may have been due to the difficulty of assigning weightings to the various inputs, as different units were used for many of the variables. Instead, model calibration was completed by individually varying the most imprecise inputs, and selecting the simulation with the most trials matching the experimental results.

6.4. Future research suggestions

The manual wheelchair stability and maneuverability changes explored in this study are specific to the cases of being stationary and when wheeling downhill. Yet wheelchair use is much broader than these situations. To increase the applicability of the research, future studies could explore the stability and maneuverability of manual wheelchairs in other use cases, such as performing turns and wheeling uphill. For wheelchair use involving turning, tire scrub should also be considered when evaluating maneuverability [14], which may require the deformation of the wheels to be modelled. The research presented also neglects the forces that user movements impart on the wheelchair (which are minimal for static and downhill cases).

Push-rim forces and user movements cause the fore-aft load distribution to change throughout each propulsion cycle, and can also cause the total weight of the system to vary from 80-110% of the static load [64]. Together these changes affect the overall
stability of the wheelchair [19]. To get an accurate representation of wheelchair dynamics during use, future research could include real user kinematic and kinetic data instead of a passive dummy. Multi-body simulations could then be used to explore the effects of user motion and push-rim forces on manual wheelchair stability in different environments.
References


[34] J. A. Inskip *et al.*, “Dynamic wheelchair seating positions impact


