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Yaw Rate and Linear Velocity Stabilized Manual Wheelchair

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Abstract— **We present the development of a prototype novel low-power, inexpensive stability control system for manual wheelchairs. Manual wheelchairs, while providing the ability to maneuver in relatively small indoor spaces, have a high center of gravity making them prone to tipping. Additionally, they can easily slide on sloped surfaces and can even spin and tip when attempting to turn or brake too quickly. When used on ramps and in outdoor environments where the surface is rarely perfectly flat (slopes greater than 1:20 (5%) are common), wheelchair users can easily encounter potentially dangerous situations. The design and evaluation of an accident prevention system for independent manual wheelchair users that increases independence by enabling mobility with greater confidence and safety is described. The system does not limit a wheelchair user's ability to manually brake, rather, if the system detects that the wheelchair is out of control, braking force will be added by the system to either one or both wheels. The prototype utilized inexpensive bicycle technologies for the wheel brake and electrical power generator assemblies. Custom servos were designed along with custom electronics and firmware in the prototype to evaluate performance. The goal of the project was to derive specifications for a control and actuation system that utilizes inexpensive bicycle components in this cost-sensitive application. The design is detailed and the final specifications provided.**

I. Introduction

There were more than 80,000 emergency room visits in 2008 due to wheelchair falls [1,10]. Falls can cause significant physical [10], and potentially emotional damage to an individual leading to a restricted lifestyle [12,13]. We designed a system to assist manual wheelchair users in maintaining control of their wheelchair and remaining upright.

Manual wheelchairs are both propelled and slowed by the hands gripping the hand rims and either pushing or holding them (some users may also use their feet). Wheelchair brakes are not designed to slow the moving wheelchair, but are intended to hold the chair still once it is stopped. Turning is accomplished by having one wheel move faster than the other wheel. Once a wheelchair is moving, it takes significant hand force to slow it down. That force must be applied gradually to keep the wheels from locking and skidding. The proposed system will not limit a wheelchair user's ability to brake, rather, if the system detects that the wheelchair is out of control, additional braking force will be added by the system to either one or both wheels.

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For example, if a wheelchair user is on a sidewalk that crosses a driveway, there may be a sudden side slope that causes the wheelchair to turn sharply downward. If the wheelchair starts to turn too quickly, the system will begin applying braking pressure to one brake to slow the turning and allow the wheelchair user to regain control of the wheelchair. On any ramp or slope, going forwards or backwards, if the wheelchair begins to move too quickly, the system will brake until the wheelchair is within a controllable speed range. By sensing and reacting to the wheel chair's linear speed and wheel rotational velocity simultaneously, a closed-loop control system is created. This also allows the system to detect and avoid hazardous situations such as skidding. The system must be inexpensive to succeed in the market and a cost goal of $75 or less was set. The electronic control module will weigh less than 3 lbs and will be designed to easily affix to most manual wheelchair frames. The braking elements, repurposed where possible from the bicycle industry, will also be designed for flexibility and easily install on a wide variety of wheelchairs.

This system was specifically designed for manual wheelchairs as opposed to power wheelchairs. Manual wheelchairs provide significant advantages for wheelchair users who are able and prefer to use them. Manual wheelchairs are relatively easy to transport as they can be folded and weigh significantly less than power wheelchairs. Most power wheelchairs require a special transport vehicle. Manual wheelchairs also require less maintenance and are less expensive.

Recent research observed that 3.86 million Americans require wheelchairs and about 70% of those use manual wheelchairs [14]. The number of wheelchair users has been increasing annually by an average annual rate of 5.9% a year [18]. Wheelchairs provide freedom, allowing users be more independent and have increased access to work, educational, and social opportunities as well as reducing dependence on others [16, 19, 20, 22]. Loss of independent mobility may necessitate a move to assisted living [15]. Independent wheelchair users have substantially better outcomes than non-wheelchair users and those who are dependent on others for use [19, 20, 22, 23]. Limited mobility affects social participation [23] and has been shown to result in social isolation, anxiety, and depression [24].

There is limited information regarding the psychological attributes of wheelchair users; however, confidence is recognized as affecting wheelchair use [23]. Rushton observed that every wheelchair user and healthcare professional reported that physical barriers challenged an individual's confidence with more confidence being required to overcome outdoor environmental barriers. Wheelchair users have limited control in the community environment and

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continually have to face new situations that are difficult to assess for safety. This ambiguity can result in both avoidance of attempts or attempting unsafe situations. One physical therapist reported that a single fall can significantly deter a wheelchair in regards to what they are willing to attempt.

Falls are the leading cause of injury-related emergency department visits in the United States and the leading cause of traumatic brain injury [2]. Wheelchair users are at increased risk of falling due to strength, range of motion, and seizure activity [4-6] as well as hardware design or failure [7,8] and user or caregiver behavior [4,6]. Falls result in significant morbidity and mortality [9] for wheelchair users as well as affecting function, activity, independence, and quality of life [3]. The fear of falling causes many people to restrict their activities [12,13].

In the Unites States, Data from the National Electronic Injury Surveillance System detail substantial increases in the number of emergency department visits due to wheelchair-related injuries [11]. There were an estimated 102,300 emergency department treated wheelchair-related injuries in 2003, twice the number from 1991 [11]. The leading cause of these injuries was tips and falls, which were responsible for more than 65% of the injuries [11]. The most frequent injuries diagnoses were fractures, contusions, and lacerations [11]. While most patients were treated and released, 17.2% required a hospital admission [11]. In a wheelchair user with a spinal cord injury, a fracture usually requires a 4 to 8 week hospital stay with significant time on bed rest, leading to loss of strength and possible blood clots [3]. Additionally, there is approximately one wheelchair-related death per week; most are due to falls [3].

There are many causes of wheelchair-related falls including transfers, ramps, propelling over uneven terrain, reaching for objects, and maneuvering curbs or stairs [3]. It is difficult with current data collection systems to quantify the exact cause of many falls. Transfer injuries, which have resulted in successful wheelchair modifications, have been found to account for less than 17% of falls [25]. Kirby et al [26] conducted a study of non-institutionalized users of manual wheelchairs to describe their tips and falls. Tips and falls were common, with 57.4% of users reporting a complete tip or fall and 66% a partial tip. The tips and falls were more than twice as likely to occur outdoors. Ramps, curbcuts, and doorways were common locations for the incidents.

Interventions to prevent wheelchair-related falls include user training, modification of buildings and outdoor environments to be more wheelchair friendly, and wheelchair modifications [3, 7]. No one strategy will be effective at preventing all types of falls, and while environmental modification may be the most effective strategy, it would also be the most costly and is probably the least likely to occur. Passive solutions that do not depend on user education have been more effective at preventing injuries, as a result the preferred focus is moving towards systems changes [29-31].

As a result of the recognition for the need for system changes, improvements to wheelchair safety are underway. There is, however, no system currently available for independent, manual wheelchair users to passively detect the wheelchair going out of control and provide automatic stability assistance. Manual wheelchairs are used by more than half of the wheelchair users in the United States [14] and are preferred for a variety of reasons, including their lightness, maneuverability, and ability to be transported in a car [15]. Manual wheelchair users commonly experience tips and falls (forwards and backwards) that could be prevented with improved technology.

II. Prototype Design and Evaluation

The dynamics of a manual chair were initially bounded using an instrumented chair in a series of tests. A wireless data sensor system; a manually operated, instrumented braking system; and computer data logging system were developed to collect real time data under safe conditions. Multiple trials were collected on an adjustable ramp designed specifically for testing wheelchairs. This data was then used in a simulation. Using this simulation, real-time estimation and control algorithms were designed and tested. A prototype chair was developed in the following four tasks:

A. Develop hardware including sensors to detect the state of the wheelchair, develop a brake actuation system, and develop a power harvesting collection system.

In order to correctly estimate the state of the wheelchair and provide appropriate control signals, hardware was created comprising of sensors, power conditioning, a processor, and various interfaces. This hardware was used to collect real-world data to verify simulations, test control loops, and prove feasibility of the system. The signal processing and control hardware contained a 3-axis gyroscope (ST Micro L3GD20), a 3-axis accelerometer (ST Micro LIS3DH), and a 3-axis Magnetometer (HMC5883L). The data from these sensors, along with data from external sensors (i.e. braking force sensors) were transmitted via an attached Bluetooth radio at up to 800 Hz and logged by a remote laptop. The sensor electronics fit on a single PCB in a volume smaller than 1”x1”x0.5” and weighing under 20 grams. Before the closed loop braking system was developed, a mathematical model of reaction of the wheelchair to braking forces was developed and the physical parameters were determined to create requirements for the prototype system design. A test braking system was developed so that the tester could safely operate each wheel brake independently by manually pulling on brake levers. The amount of braking force applied was sensed using strain-gauge sensors and captured by instrumentation. The braking data was synchronized to the sensor data allowing the parameters of the model to be tested and verified. Additionally, using a process called system identification, the underlying physical parameters were determined which inform the control system on how much braking force is required for a desired reaction.

B. Develop firmware to analyze sensor data, determine if instability exists, and activate the brakes to safely bring the wheelchair under control.

The braking control system relies on knowing the state of the wheelchair in order to determine the appropriate braking force. For this project, the state variables included the current wheelchair velocity, turn-rate, sensor errors, and orientation (i.e. tilt). The true values of the state variables are estimated using data fusion techniques in a Kalman Filter.

Using the estimated states and the mathematical model of the wheelchair, a control system was designed to keep the wheelchair user in safe operating conditions (Fig. 1). The final output from the control system was the amount of braking force to be applied to the left and right wheels. These values are determined from a combined value of two pseudo independent control systems—one that controls the wheelchair linear velocity and one that controls the wheelchair rotation rate. The rotation rate controller determines the differential braking force which must be applied and the linear velocity controller determines the braking force which must be applied to both wheels simultaneously. This control system was tuned using physical parameters of the wheelchair and measured results run through the simulation program. The control system was designed to avoid disruptive braking force that could cause the wheelchair to skid or result in an uncomfortable feel.

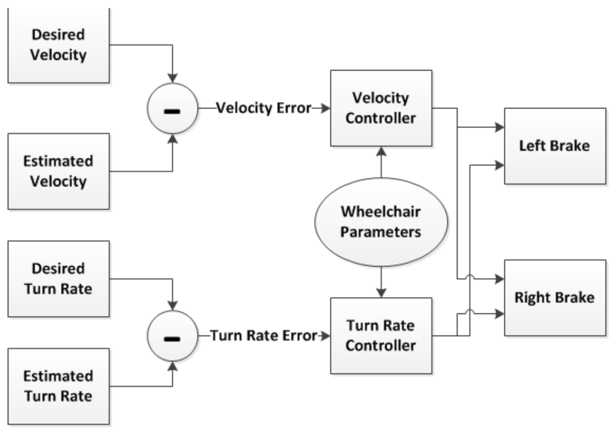

Figure 1: Simplified function diagram of the braking control system.

C. Construct a prototype sensor system and install in existing manual wheelchair

To construct a prototype sensor system, a PCB was designed and built. Firmware was written to allow for high speed and efficient data transfers between the sensor and processor while accurately maintaining a low-latency timestamp with 1 microsecond resolution. To decrease the processor overhead and latency, a real-time state machine architecture was used in lieu of an operating system. By using a Bluetooth 2.1 EDR compatible wireless radio, the sensor data could be read by nearly all modern computing systems such as desktops, laptops, or mobile device compatible with Bluetooth. The range and robustness of the wireless data transfer system was tested to ensure reliable communications could take place through the entire testing area at the required data rates.

The sensor system was installed on the side of the wheelchair in a location which was rigid with respect to the wheelchair frame. The system was powered with a rechargeable lithium polymer battery.

Relatively inexpensive Sturmey-Archer X-FDD bicycle hubs were fitted to provide braking force and electricity (Fig. 2). The drum brakes were activated with a bowden cable, connected to an electric gear motor. The cables are flexible, and they were routed to the motors and attached to a fixture that was fastened to the frame of the wheelchair below the seat. The brake hubs were equipped with replacement fittings to securely mount to the wheelchairs quick-release mounting hardware. The cables were routed through a fixture housing two load cells, to provide measurement of the applied braking force (Fig. 1). The measurement of cable tension allows for accurate measurement of braking response relative to motor input, and was used to model the braking system. A key outcome of the project was determining the force and latency requirements for controlling these types of bicycle hubs in the wheelchair control application.

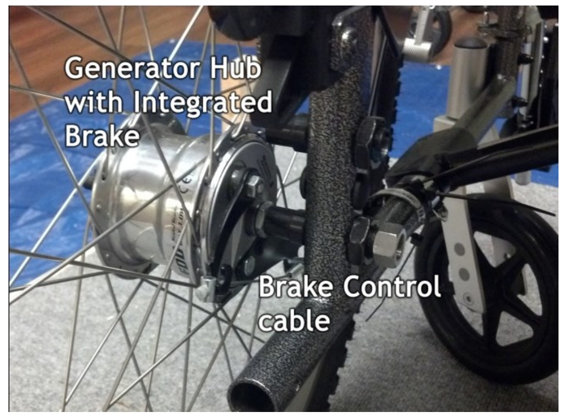

Figure 2: Hub with drum brakes and integral generator.

D. Evaluate the system for feasibility and next-generation prototype specifications by conducting laboratory tests for generating power, detecting out of control motion, and safely braking

To allow for repeatable testing over a large variety of conditions, an adjustable wheelchair ramp was constructed. This 16 foot long ramp could be adjusted in height to allow for grades between 5% and up to 20%. Using a split level design, the ramp could also be configured to allow for a change in grade mid-way down the ramp.

As previously described, the wheelchair was fitted with the sensor and processing hardware and an RN-41 Bluetooth radio. Braking force was measured using two transcend BSS-250 Force sensors. A test matrix was designed to test the various aspects of the system, 4 trails for each test were performed and logged using the logging system previously described. The tests were all performed with varying grades, surfaces, and wheel types to ensure system robustness. Data from each test were used to create and validate the mathematical simulation.

Power draw of the electronics while the radio was inactive was measured to be under 10mA @ 5V (0.05 Watts). During high rate data transfers with an active radio, power draw increased to 30mA @ 5V (0.15 Watts) over a 1s average. While in sleep mode, the system draws under 0.1mA @ 5V and can be activated automatically upon detection of movement by the accelerometer. A 64 gram lithium polymer battery with a size of roughly 8cm x 3cm x 1cm is capable of holding 12 watt/hours of power and is available for a retail cost under $5. As a result of the lower power draw of the electronics, they could be continually active without recharging for roughly 240 hours. By putting the electronics to sleep while the user is not in motion, a single charge would last much longer. While the system is in use, the wheel hub generators recharge, allowing continual use without plugging the system into an electrical outlet.

III. Results

The requirements and specifications, as seen in the table below fully define a next generation system. Further improvements are likely achievable to improve power consumption, and latency and mass of the brakes. Through the data collected during testing, we have demonstrated how the critical wheelchair states can be estimated accurately and can be controlled to stay within safe bounds.

Braking Force (Max):	50 N
Braking Resolution:	0.25 N
Braking Latency:	100 mS
Power Draw (Inactive Mode):	0.01 mW
Power Draw (Sensing Mode):	30 mW
Power Draw (Active Braking Mode):	1000 mW

IV. Conclusion

The project aimed to create an inexpensive yaw rate and linear velocity stabilized manual wheelchair. The key enabling mechanical technologies of the wheel brakes and power generator were borrowed from bicycles where mass production keeps prices low and quality high. The primary objective of this research was to specify the actuator requirements for a stability control system to effectively utilize the bicycle braking system and state the electrical power requirements for a functioning system. Bicycle components are suitable in this application and power requirements are modest and likely improvable.

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