Veterans Administration

Journal of Rehabilitation Research
and Development Vol. 24 No. 2
Pages 87–98

An adaptable optimal controller for electric wheelchairs*

KEVIN E. BROWN, M.S.; RAFAEL M. IÑIGO, Ph.D.; BARRY W. JOHNSON, Ph.D.
Department of Electrical Engineering, School of Engineering and Applied Science, University of Virginia, Charlottesville, Virginia 22901

Abstract—A microcomputer-based optimal control system for electric wheelchairs is presented. As the weight of a wheelchair is generally small compared to that of a user, the user's actual weight can have a tremendous impact on the dynamics of the wheelchair. The velocity feedback controller presented here was designed to give an optimal response for all users by incorporating a measure of the individual user's weight into the control algorithm. This allows optimal control of the motor velocities while assuring the constant stability of the system. Control of the system is managed by a modified proportional-integral-derivative (PID) controller, and the adaptability is handled by use of variable-structure control. The factors and considerations involved in such a system are identified, as are the advantages and disadvantages of the particular control strategy used. Experimental results are presented.

INTRODUCTION

Of the many external variables affecting the control of an electric wheelchair, the weight of the individual user was found to have the greatest impact (1). In spite of this fact, electric wheelchair manufacturers, if they account for the user's weight at all, design only for users of average weight. This leads to a degradation of system response for users of other than average weight. In addition, automatic velocity control is seldom incorporated into commercial wheelchair controllers.

A Z80-based velocity feedback controller has been designed that can give an optimal response for users of all weights. A software algorithm was implemented that incorporates a measure of the user's weight into the determination of the coefficients for the control parameters.

The theoretical basis for this work was developed by Johnson (1,2). His goal was the design of a self-adaptive controller that could be used in an automatic guidance system for electric wheelchairs. He designed a control system that used a modified PID control scheme to calculate and combine the control parameters, and a variable structure controller to determine the appropriate coefficients for those parameters. However, practical problems prevented implementation.

The adaptable controller presented here is a modified version of the controller designed by Johnson. The designed system was implemented in an electric wheelchair and its response tested in the field. A number of factors were identified as being pertinent to the control of the system, and were tested thoroughly to determine their optimal condition.

THEORETICAL WORK LEADING TO THIS PROJECT

As stated above, the general philosophy upon which this work is based was developed by Johnson (1). Johnson outlines a controller that will, theoretically, adapt itself to variations in load inertia and

*This work was supported by NIHR Grant #GOO–83–00072 and performed at the Rehabilitation Engineering Center, P.O. Box 3368, University Station, Charlottesville, VA 22903.

damping so as to maintain system performance to within 10 percent of the optimal. This controller is composed of two self-reliant elements that are combined interdependently. The first element is the actual control algorithm used to perform the control, and the second is the adaptive algorithm that determines the exact control parameter coefficients that should be used.

For the actual task of control, a modified proportional-integral-derivative (PID) controller was selected. The modified PID controller differs from a conventional PID controller in that it has an integral term dependent on the error signal combined with derivative and proportional terms which depend only on the output.

The concept used to make the system adaptive is variable structure control (VSC). Basically, the idea is to have a finite set of control equations differing only in the parameter coefficients; each equation gives acceptable system performance under a different range of loading conditions. The range of loading conditions associated with a particular control equation is referred to as a *region*. As conditions encountered by the wheelchair vary from those of one region to another, the system selects the new control equation from memory, thus preserving acceptable system performance.

Johnson has shown (1) that the number of regions needed for electric wheelchair control can be limited to four while maintaining system performance to within 10 percent of optimal. These regions are a function of two parameters: the load inertia and the load damping. When the four regions are mapped into a friction-inertia plane, it can be seen that variations in load damping have little effect on region specification, so that the primary factor involved is variation of the load inertia.

The actual implementation of the self-adaptive controller was attempted by Schwab (3,4). He devised a bench-test setup configured to provide known loading conditions, and this system afforded data which demonstrated the controller's optimal response to a range of loading conditions for a given region. However, the test setup also exposed the

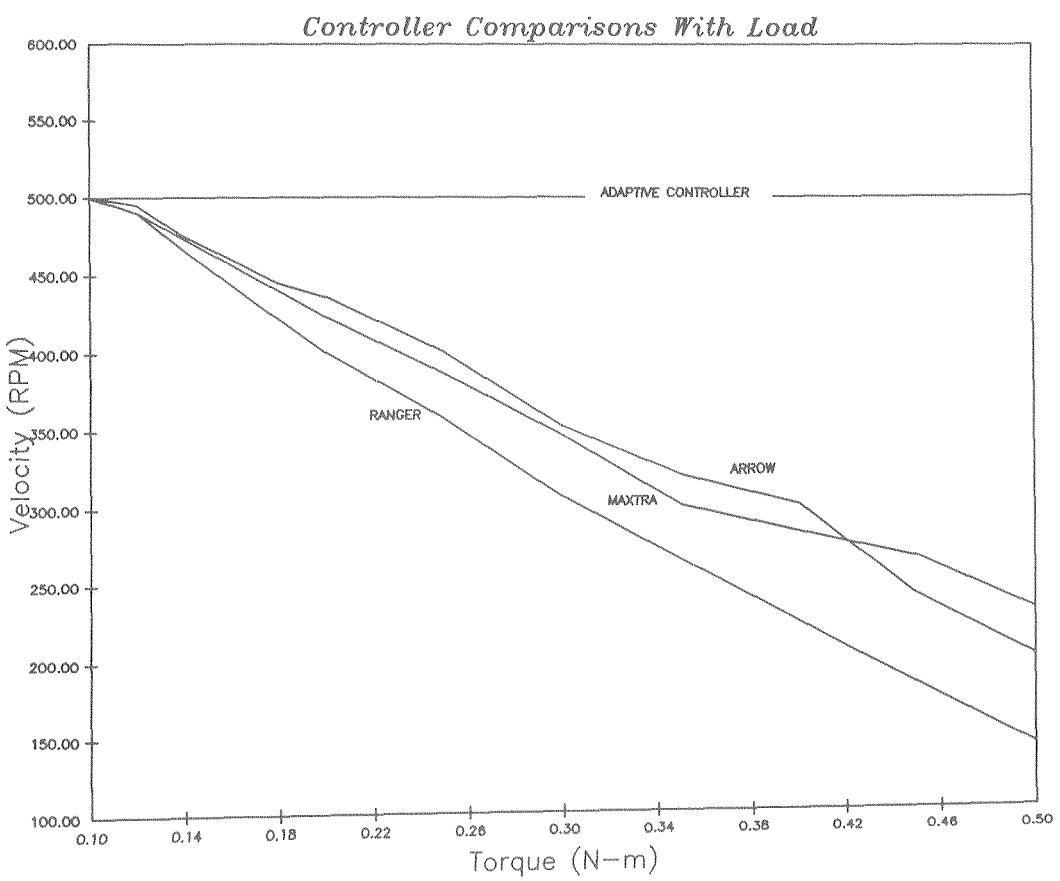

Figure 1.
Comparison of adaptable controller with three Invacare controllers.

practical difficulties of implementing crucial self-adaptive control algorithms under the constraints of a real-time system. In his tests of the individual controller regions, Schwab showed that each region maintained adequate performance for a wide range of loading conditions and that the value of the load damping had little effect. These results engendered the idea of using the variable structure controller in an electric wheelchair, but making it manually adaptable rather than self-adaptive. As the exact region of operation is primarily dependent on the load inertia, which in turn is proportional to the user's weight, the proper region of operation can be estimated and set in software with the assurance of adequate system performance.

The bench testing also included a comparison of the velocity response of the adaptable controller with some commercially available "controllers" for electric wheelchairs. The test was performed by starting at a given speed with minimum torque and increasing the torque without changing the reference signal. Results are shown in **Figure 1** (3), from which it is evident that the adaptable controller offers superior performance.

SYSTEM SOFTWARE

Control Strategy Implementation

The modified PID controller used in this control system required four equations, one for each region and each containing three terms. These three terms are each of the three control parameters (proportional, integral, and derivative terms) modified by a calculated coefficient.

The first of these terms, the proportional term, is taken directly from the actual motor velocity. The second term is the integral term. At each sample time the desired motor velocity is subtracted from the actual and this difference is summed with all past sample errors, forming a quantized error integral q. The third term, being the derivative of the actual velocity, represents the acceleration of the motor. It is obtained by calculating the difference between the current velocity and the last sampled velocity and dividing by the elapsed time since that last sample, i.e., the sampling time.

The parameter coefficients for each of the control equations were derived in (1). The final control equations used for the complete control of the wheelchair were:

$$REGION\ 1:\ e_a = -(0.81738 \times \omega_l + 0.01325 \times \omega_{l_p} + 0.34594 \times q) \quad [1]$$

$$REGION\ 2:\ e_a = -(6.81882 \times \omega_l + 0.05636 \times \omega_{l_p} + 0.85837 \times q) \quad [2]$$

$$REGION\ 3:\ e_a = -(-0.99209 \times \omega_l - 0.00042 \times \omega_{l_p} + 0.18803 \times q) \quad [3]$$

$$REGION\ 4:\ e_a = -(2.72040 \times \omega_l + 0.02719 \times \omega_{l_p} + 0.51111 \times q) \quad [4]$$

Programming Outline

The software developed for the adaptable controller was coded partly in Fortran–80 and partly in Z–80 assembly language. The primary control routines were written in Fortran, and all input and output operations were handled by assembly language. The programming involved will be described here in two ways. First, the general flow of the programming will be outlined, giving only the basic flow without reference to specific data transfers. The data flow through the control process will then be discussed, including the flow through both the hardware and the software.

The basic programming flow is shown in **Figure 2**. The memory, timing circuits, and CPU registers are initialized prior to the accession of the control

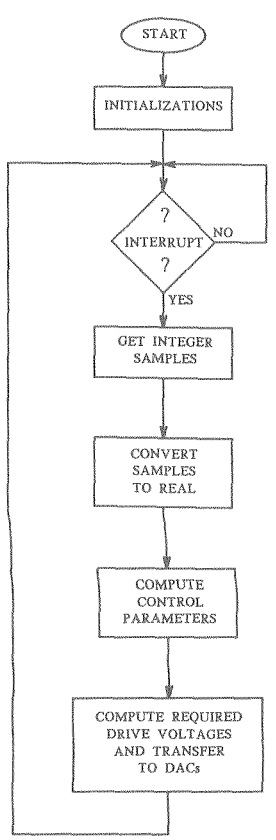

Figure 2.
Programming flow diagram for the adaptable controller.

loop. Upon receipt of a timing interrupt, integer samples are taken of the left and right actual motor velocities and of the desired speed and direction of the wheelchair. These samples are converted to real numbers, the desired motor velocities are determined, the three control parameters are calculated for each motor, and the necessary output motor drive voltages are determined and output. The process is repeated indefinitely beginning with the wait for a timing interrupt.

The data flow involved in this procedure in shown in **Figure 3**. This chart is a combination of both hardware and software elements of the system, with each software box representing a subroutine or a small group of closely related subroutines. The data first enter the system through the hardware inputs (the hardware steps of the process are represented by double boxes), and are converted to real numbers. The control parameters are calculated and used in the appropriate region equation to determine the output, and this value is then converted into a bit pattern for output to the motor system.

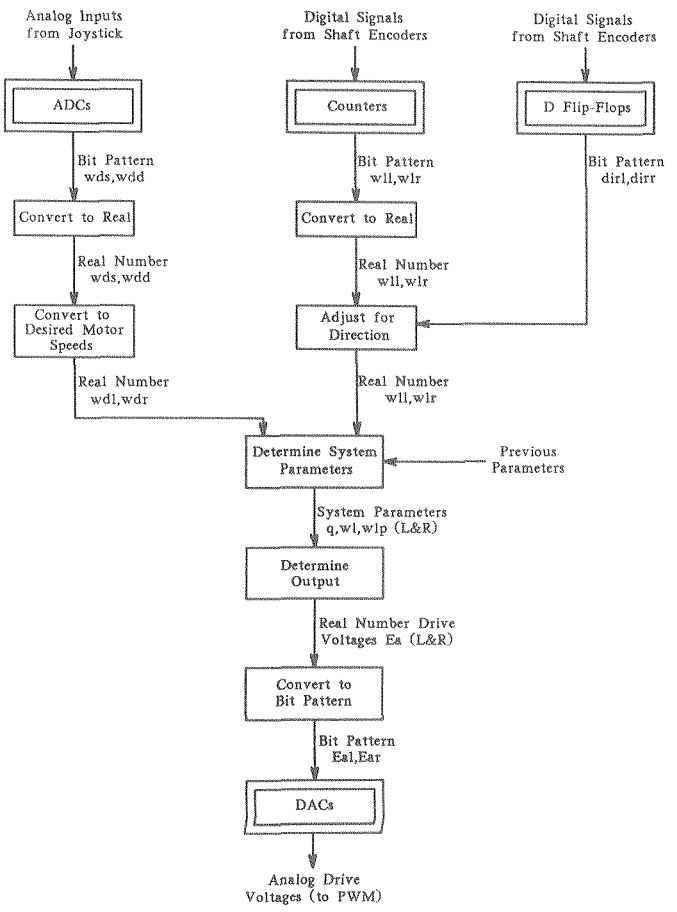

Figure 3.
Data flow diagram for the adaptable controller.

SYSTEM HARDWARE

A Zilog Z80 processor/STD bus system was used as the basis of the control system hardware. The entire electrical system is made up of four distinct subdivisions (see **Figure 4**). The primary section is the STD bus rack (microcomputer), and it contains all of the digital processing components and the input/output circuitry connecting it to the other three sections. The other sections are: the joystick, to provide the desired signals for the controller; the incremental shaft encoders, to provide information on the actual wheelchair status; and the pulse-width-modulator (PWM) power supply, which converts the controller output into a variable direct current voltage to drive the motors.

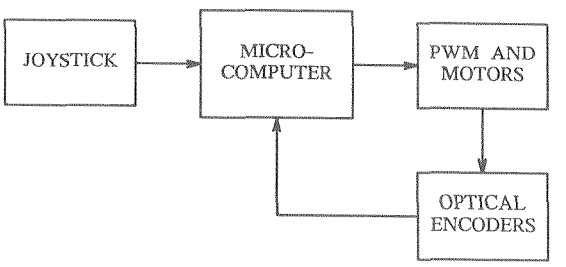

Figure 4.
System subdivisions.

A Prolog STD bus configuration was used in a motherboard rack to contain and interface the individual boards used in this system (**Figure 5**). The processing system was completely contained in three STD bus cards: the processor card, the power supply card, and the input/output card.

The processor card is a CPU–2+ Z80B processor, counter/timer card purchased from Computer Dynamics, Inc. It contains a 6-MHz Z80B microprocessor and a Zilog Z80B Counter/Timer Chip (CTC) with four independent 16-bit channels. The six memory sockets included on the CPU card proved sufficient to contain all of the RAM and ROM necessary for operation of the system.

The power supply board was required to convert the 24 volts of the wheelchair battery pack into the voltages necessary to power the STD rack, the attached boards, the joystick, and the optical encoders. The STD bus configuration specifies that pins 55 and 56 carry +12 and −12 volts, respectively. However, components on the input/output card required ±15 volts for proper operation. There-

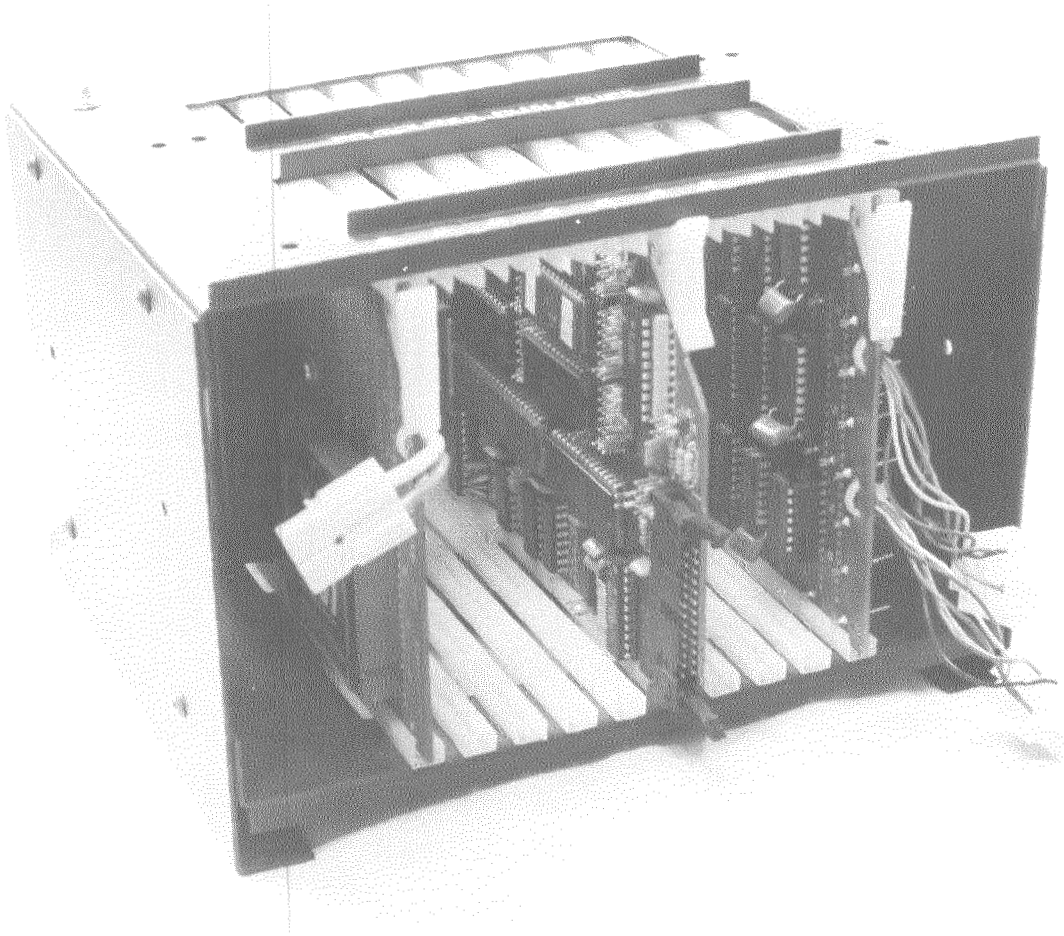

Figure 5.
Controller processing system.

fore, a slight modification was made to the STD bus configuration to replace the +12 volt line with +15 volts and the −12 volt line with −15 volts. The constructed power supply board generates all of the required voltages through the use of two purchased voltage converters. Both voltage converters take 24 volts as input.

The third and final board required for system operation is the input/output board. This board contains all of the special circuitry needed for the input of the joystick signals and the motor velocities, and for the output of the calculated motor voltages. The contents and functions of this board are described in detail in a later section.

Joystick

The joystick selected for use in this system contains the common double brush-type potentiometer with the two brushes oriented in perpendicular planes. The structure of this joystick allows the individual potentiometers to be powered by +5 volts which produces signals which range between 0 and 5 volts. Since the required input of the analog-to-digital converters is also in the 0 to 5 volt range, this eliminated the need for any signal conditioning between the two circuits.

Optical Shaft Encoders

Datametrics K15 Series Modular Encoders were used to gauge the actual motor velocities. This encoder is shown in **Figure 6** and consists of a slotted disc/hub assembly which mounts to the motor shaft, enclosed by a photohead assembly mounted onto the motor housing and sealed to prevent soiling. Each encoder produces two frequency signals with a phase quadrature of approximately 90 degrees. This allows for motor velocity information to be obtained from one channel because the frequency

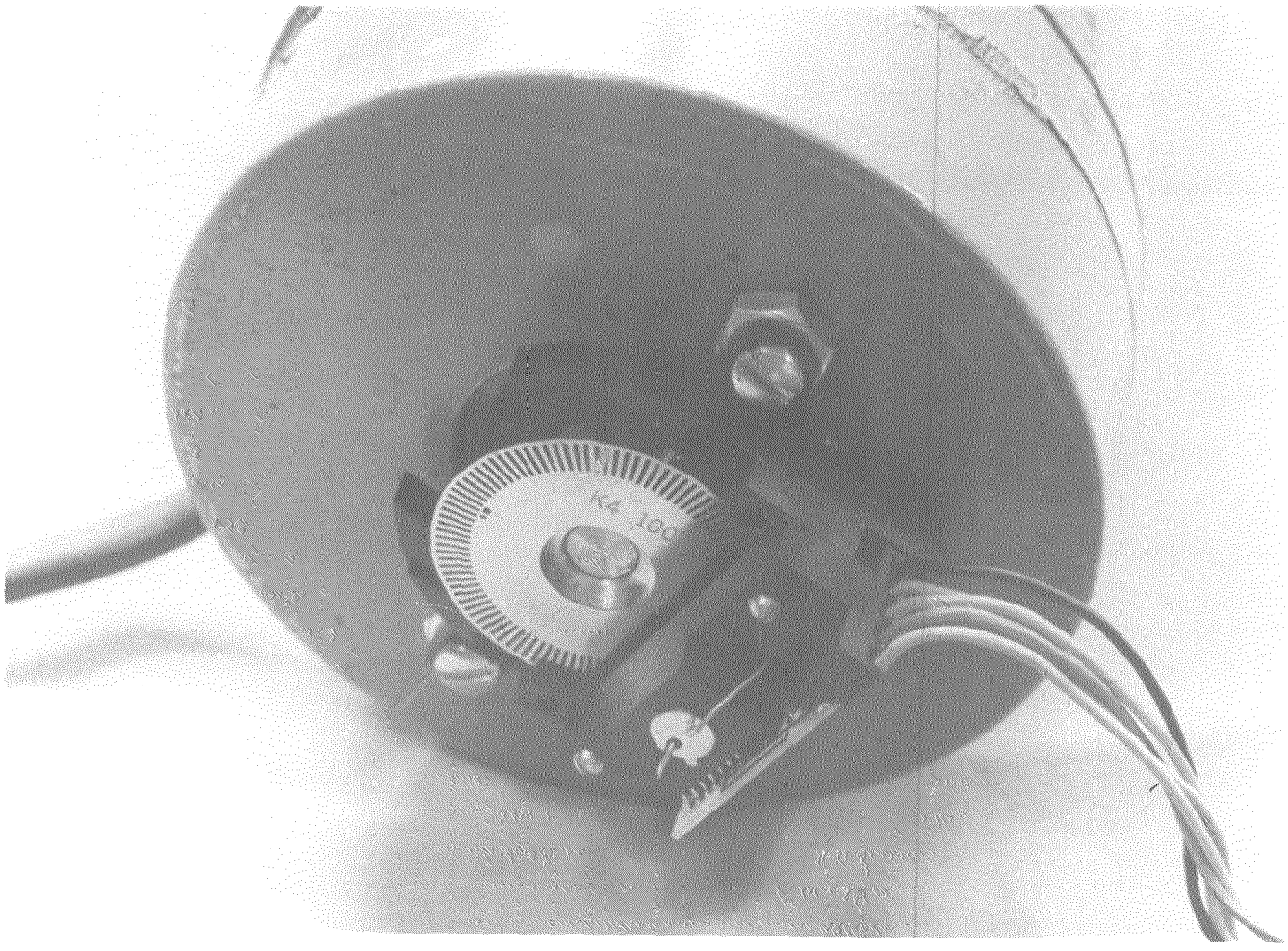

Figure 6.
Optical shaft encoder.

of the output waveform is proportional to the motor speed, while motor directional information can be extracted from the phase difference between the two channels.

Pulse Width Modulator

The output of the microcomputer system is a voltage in the −6 to +6 volt range for each motor. For efficient performance of the motors, this signal needs to be converted into a pulse-width-modulated (PWM) power signal. In order to simplify the task of implementing the adaptable controller, the motor drive of the existing Invacare Maxtra controller was used. Since the joystick assembly of the Maxtra controller generates the motor command signals, the joystick inputs of the motor drive can be directly replaced by the adaptable control device.

Control System Input/Output

The input and output circuitry of the computer system is the heart of the hardware developed for the adaptable controller. The I/O board is made up of four distinct circuits. As shown in **Figure 7**, three circuits are for inputting the desired and actual system status and the fourth is for outputting the corrected motor signals to the PWM and motors. In the following paragraphs, each of these circuits will be analyzed in detail.

Joystick Input. As previously pointed out, the output of the joystick is directly applicable to the input of the A/D converters, so no signal conditioning is required. The circuit is therefore made up primarily of the A/D converters. The operation of the circuit begins with a "start conversion" signal from the CPU, which causes an analog sample to be taken and converted. When conversion is complete (as signaled by the ADC's) the CPU reads the digital sample.

Motor Velocity Input. The basic premise of op-

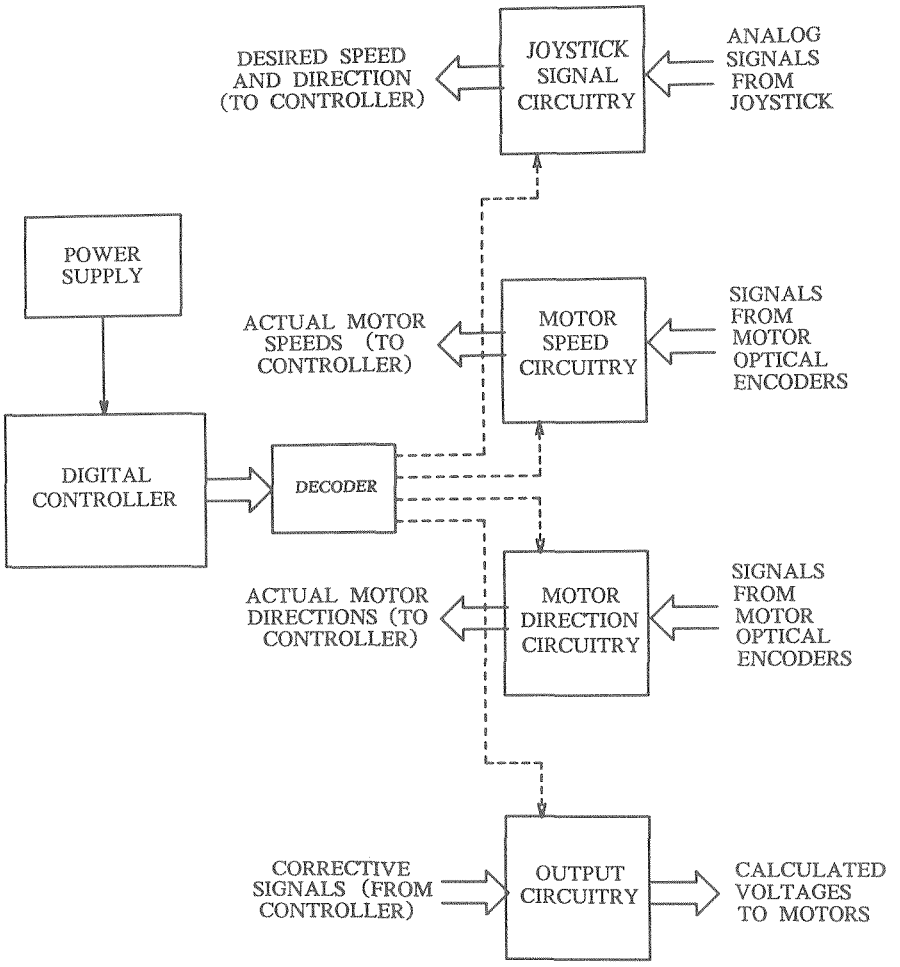

Figure 7.
Processing system block diagram.

eration of the motor velocity input circuitry is the counting of pulses from one of the output channels of each optical shaft encoder. A signal from the CPU during the timing interrupt service routine regularly latches (every 55 msec) the counts contained in the counters. A second signal then clears and restarts the counters. This leaves the motor velocity counts stored in latches until the control program reads them.

Motor Direction Input. Since the motor velocity input provides only the speed at which the motors are turning, additional circuitry is needed to determine their direction of rotation. The circuit devised for this job is very simple, using only two D flip-flops. Each shaft encoder produces two square-wave signals with a phase relationship dependent upon the motor direction. With one of these signals driving the D input of a D flip-flop and the other driving the clock, the output Q will be either a logical 1 or a logical 0 depending on the motor direction; i.e., depending on which input first becomes active because its phase signal is leading.

Motor Signal Output. To produce an analog output voltage corresponding to a calculated digital signal, use is made of digital-to-analog converters (DACs) feeding into operational-amplifier (op-amp) circuits. The output circuitry first receives an eight-bit digital signal from the data bus for each motor. Each digital signal is then converted into an equivalent analog voltage, and amplified and level shifted by simple op-amp circuits to obtain the required output voltage range.

Data Collection Hardware

For the successful collection and processing of performance data for evaluation purposes, some additional hardware had to be added to the system. (This hardware is not necessary for actual wheel-

chair use.) First, more random access memory (RAM) had to be included to supply the system with enough available memory in which to store the data, and then an interface had to be provided between the wheelchair system and the outside world in order to retrieve that data.

An adequate amount of memory was obtained through the addition of a byte wide memory card, and to transfer the collected data out of the wheelchair system, a serial I/O setup was installed which could be easily disconnected from the receiving system during the actual collection of data.

TESTING AND RESULTS

Throughout the course of construction and testing of the adaptable controller, several factors were identified as being important to the control of the wheelchair. The major factors consist of the joystick decoding, the sampling time, and the integral state error. Also presented are the analysis techniques used to study the system performance (specifically data plots and stability analyses).

Joystick Signal Decoding

The joystick relates two signals which will be referred to as the desired wheelchair speed (ω_{ds}) and direction (ω_{dd}). These signals are shown graphically in **Figure 8**. Before any specific mathematical expressions can be derived to convert these signals into desired motor velocities, two decisions must be made regarding the behavior of the wheelchair in relation to the joystick position. The first decision involves the steering of the wheelchair and identifies the direction of motion the wheelchair should take as the joystick is placed upon each of the possible radial lines. The second decision relates the speed of the wheelchair with the angular displacement of the joystick from the vertical along each of these radial lines.

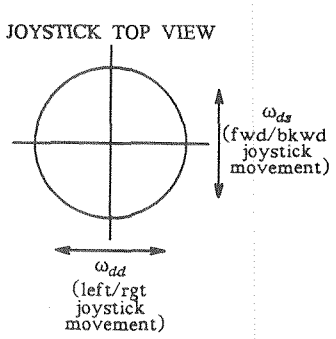

Figure 8.
Joystick signals.

Two techniques were studied with regard to the wheelchair steering. Shown in **Figure 9**, they differ mainly in the reverse direction. The first technique, **Figure 9A**, gives true direction steering for reverse motion. A dead zone is needed to eliminate any abrupt changes in the signals, and two separate sets of decoding equations are required—the final result is an excessive amount of software. The chosen technique is shown in **Figure 9B**. This allows for smooth transitions throughout the circumference of the joystick and therefore limits the amount of calculations required.

Relating the wheelchair speed to the angle of joystick displacement proved to be the most involved aspect of the decoding problem. At first a simple linear relation was devised, making the wheelchair speed proportional to the joystick displacement and the individual motor speeds a linear combination of this displacement and the joystick signal ω_{dd}. This relation proved to have too little resolution at lower speeds, making tight maneuvering difficult.

The wheelchair speed was then related by a square law to the joystick displacement. This slightly improved the maneuverability of the wheelchair, but the response was still too jerky at low speeds, due to limited joystick resolution. So, an entirely new scheme was developed for the wheelchair speed decoding which encompassed the concept of differential wheel velocities. In this scheme the joystick variables ω_{ds} and ω_{dd} are transformed into a desired average speed and a wheel speed differential, respectively. They are then mapped into the left and right desired motor velocities. This scheme combines all of the desirable aspects previously mentioned, and proved to be the most successful joystick decoding.

Data Collection Analysis

The goal behind the data collection was to gain an understanding of the effects of certain variables in the control algorithm. This was accomplished by plotting sets of system response data. Each set studied the effects of one variable by testing the step response of the system for a range of values of that variable while holding all other things constant. Three variables in all were tested in this manner;

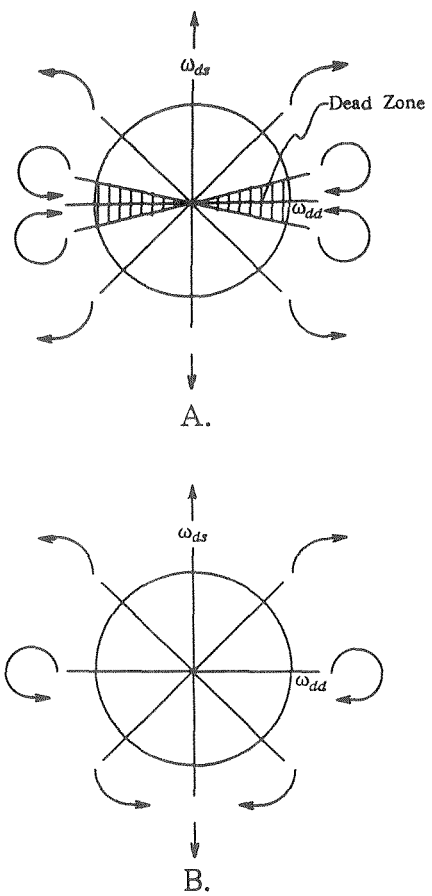

Figure 9.
Two wheelchair steering techniques.

the effects of varying the region number was the original objective, and this soon led to tests using differing loads and terrains. The effect of the sampling time was also studied, and these tests are also presented.

Region and Weight Variation. The step response of the system was tested at 55 msec interrupts for all four regions for each of four user weights: 40, 125, 180, and 260 pounds (18, 57, 82, and 118 kilograms). The regions in order of increasing user weight are 3, 1, 4, and then 2.

The left motor responses for a 180-pound (82 kg) user are shown in **Figure 10**, which indicates that the best response occurs in region 4. Region 3, designed for the lightest of users, has a large peak overshoot (~26 percent) and a long settling time. Region 1 shows an improved response over Region 3, yet still has a peak overshoot large enough to be felt by the user. The region "expecting" the heaviest of users, Region 2, shows continued overshoot in the response. In view of the other region responses, one would expect this region to be noticeably overdamped, yet from a logical point of view this response is anticipated. The control coefficients are adjusted to calculate a response for a much heavier user. That calculated output, when received by a system much lighter than expected, will cause the wheelchair to overreact and each overreaction is "corrected" by another signal again calculated for a heavier user. This oscillation continues about any non-zero reference signal. To the user, a rapid oscillation such as this is perceived as a vibration not unlike running on rough terrain.

Tests identical to these were run with users of several other weights. The best responses for two of these are shown in **Figure 11**. The best response represents coincidence of user weight and region number, thus validating the use of equations [1] to [4] for the respective regions.

Terrain Variation. The system was tested for its response and speed control while moving both up and down a ramp of the type used by handicapped persons for constant region 4 operation with a 180 pound (82 kg) test subject and a 55 msec sampling time, (see **Figure 12**). Recall that this is the proper region of operation for these conditions.

The ramp used held a fairly constant slope of about 4.3 degrees and connected to a length of level ground at each end. This was desired as it gave the wheelchair time to get up to a normal level-ground speed before hitting the ramp, and time to recover back to that speed after completing it.

The results of these tests show that the controller maintained the desired speed quite well. The up-ramp test results are shown in **Figure 12A**. Here the wheelchair speed first increases to the desired value, then dips slightly as the ramp is encountered. The desired speed is then quickly recovered and maintained up the ramp, spiking high only slightly when level ground is again reached. In the down-ramp test, **Figure 12B**, a very similar response is shown. The desired speed is closely maintained except for small fluctuations corresponding to the beginning and end of the ramp.

Overall, the control system responded very rapidly to changes in the terrain, with any sudden disturbances being quickly detected and corrected for. Although this test was not performed using a commercial "controller," the bench test described in Section II indicates that their performance would not match that of the adaptable controller.

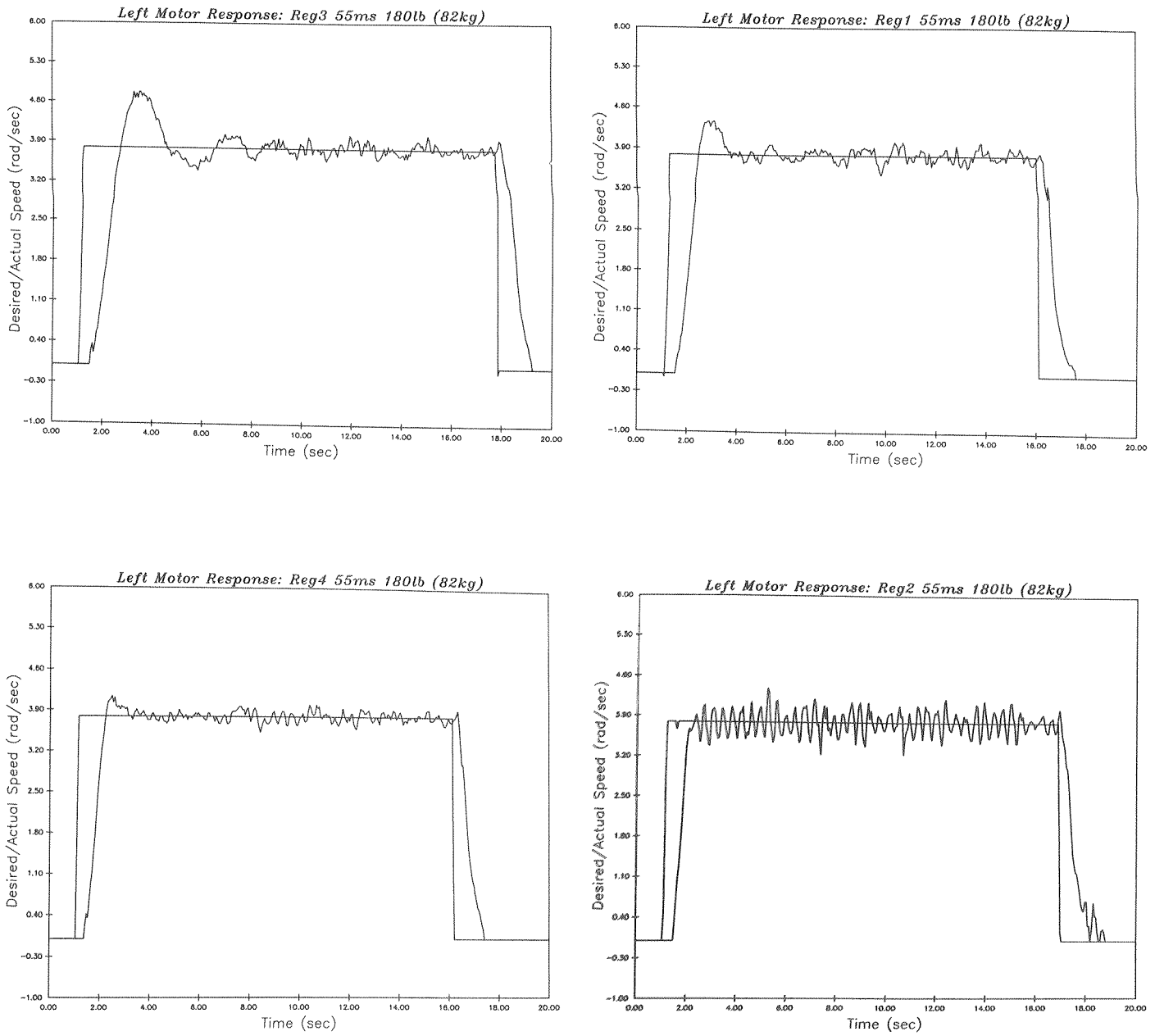

Figure 10.
Step response for a 180-pound user.

Sampling Time. The maximum sampling time guaranteeing the stability of the system is 55.5 msec (3). This sampling time was used during the development of the wheelchair installed adaptable controller. When the system was complete and all real-time calculations were minimized, the system timing was removed to allow a free running measurement of the processing time. It resulted in a run-through time of 9.8 msec. This afforded quite a bit of leeway in the exact timing of the system, so tests were run to determine if a decreased sampling time would secure an improved system response. Three different sampling times were examined in a variety of tests: 15, 37, and 55 milliseconds.

Two approaches employed to study the sampling time were the previously mentioned collection and plotting of response data, and a stability analysis of the system. The results of the data plot analyses indicated that 55 msec is a better sampling time than either 15 or 37 msec. The results of a stability analysis of the control system showed that as the sampling time is decreased, the complex conjugate

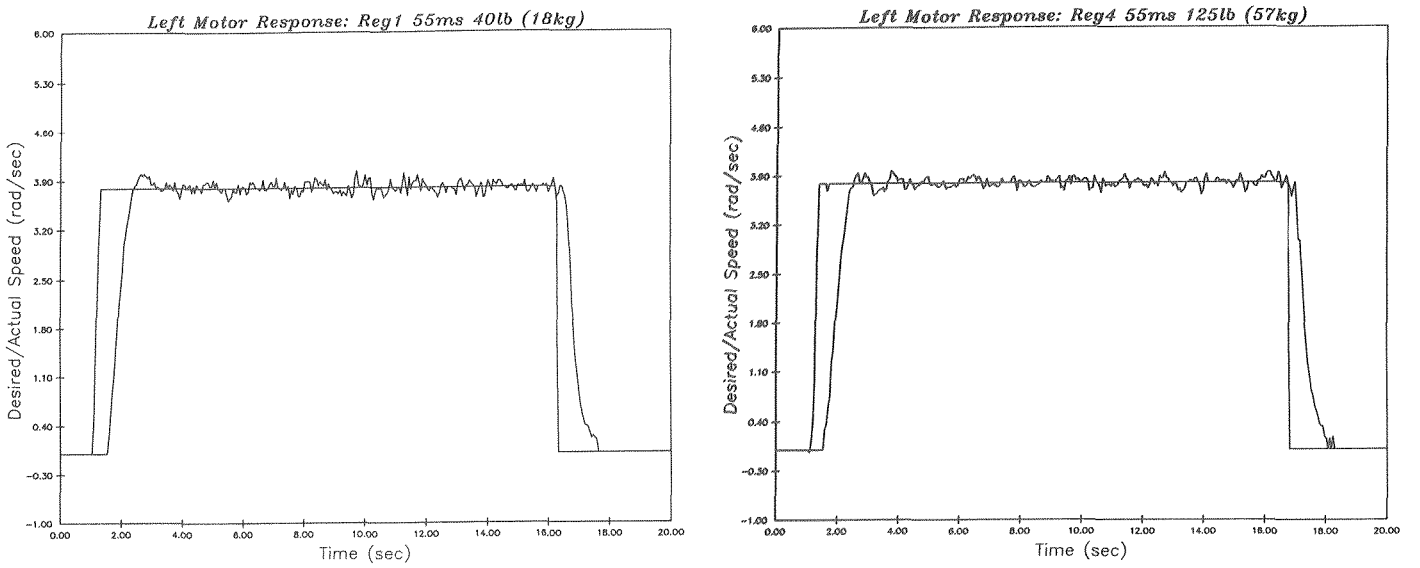

Figure 11.
Best step responses for 40-pound and 125-pound users.

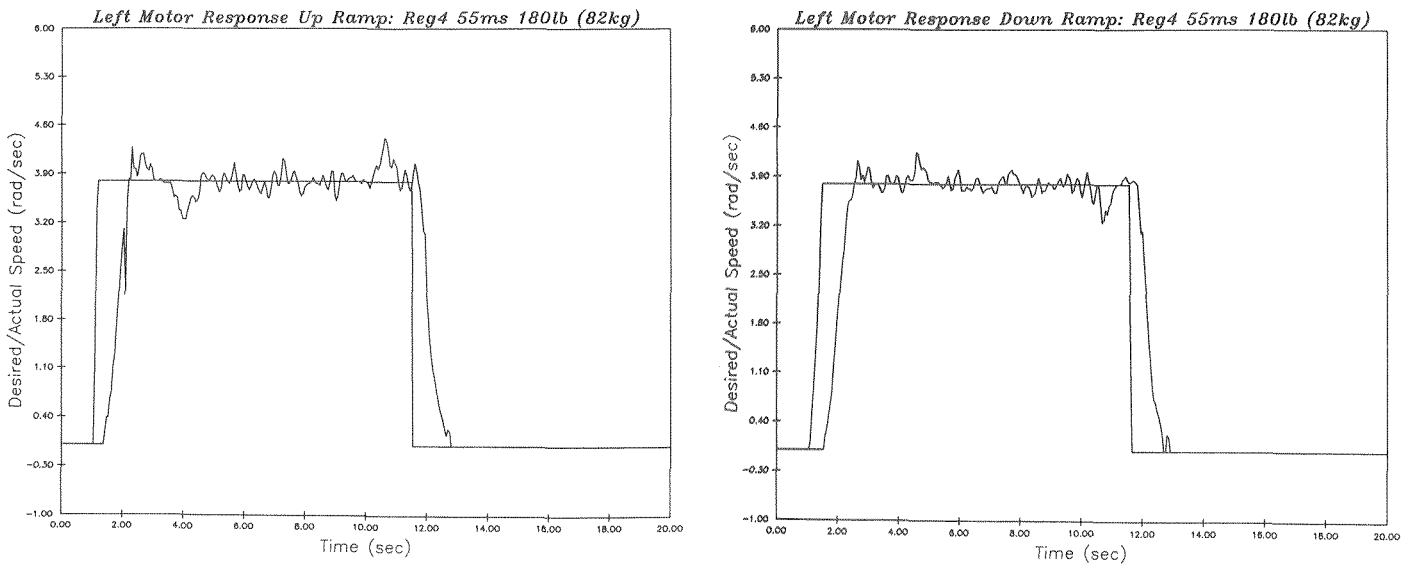

Figure 12.
Ramp test data plots.

pair of poles of the third-order system move outward towards the unit circle. This indicates an increasing degree of oscillation and therefore a decreasing degree of stability. Thus the 55 msec sampling time was chosen as giving the best system response, and for this reason it was used in all other tests in which the sampling time is held constant.

The Integral State Error Problem. The integral state error, one of the three parameters used in the control scheme of the adaptable controller, is fundamentally the sum of all past sampled differences between the actual and desired speed of a motor. When the control system encounters any set of conditions to which it cannot properly respond, this error term builds up. If those conditions persist, it can reach values which, upon removal of those conditions, will require an inordinate number of sampling periods to eliminate. This can cause the wheelchair to continue moving for an abnormal length of time beyond the release of the joystick.

All attempts at limiting the magnitude of the integral term proved to also limit the control response of the wheelchair. No complete solution to the error problem was found, but two modifications were made to the code which eliminated the possibility of build-up under normal wheelchair usage. The first was to limit the maximum speed so that the output signal can never exceed the capability of the motors even under conditions of heavy loading or ramp climbing. The second modification was to add an instruction to the program which sets the integral state error to zero at any time that both the desired and actual velocities are zero. In this way, if the wheelchair gets stuck against a curb and the user tries to move back, the joystick passes through the center position and the error immediately becomes equal to zero.

CONCLUSIONS

The purpose of this work was to develop and implement an adaptable digital controller for an electric wheelchair by modifying a theoretical self-adaptable controller. The hardware and software necessary for such a system was developed, installed in an electric wheelchair, and tested for its response and control.

The results showed that the system had an exceptionally fast response time with near-optimal performance when adjusted correctly, yet the high speed of the response sometimes led to difficulty in maneuvering the wheelchair in an elegant and graceful manner. It was found that the mapping of the joystick space had a sizable effect on the response and control. Many different tactics were employed in the study of these effects, and it was decided that there is a lot of room for further study in this area. The issue of the sampling time was also addressed, and it was found that, for the particular coefficients of the control parameters used, a sampling time of close to 55 msec gave the most stable and effective control.

The only major problem encountered with the control system in this work was the tendency of the integral state to amass extensive amounts of error. This problem was dealt with in a practical way which minimized the possibility of it ever affecting the control of the system.

REFERENCES

1. Johnson BW: Microcomputer-Based Adaptive Controller for an Electric Wheelchair Guidance System. Doctoral Dissertation, University of Virginia, Charlottesville, VA. January 1983.
2. Johnson BW and Aylor JH: Design of an adaptive controller for microcomputer implementation. *IEEE Transactions on Industrial Electronics*, IE-ee(1):28–33, February 1986.
3. Schwab AJ: Implementation of a Microcomputer-Based Adaptive Controller for Electric Wheelchairs. Master's Thesis, University of Virginia, Charlottesville, VA. May 1985.
4. Schwab AJ: A microcomputer-based adaptive controller for electric wheelchairs. *Proceedings of the Eighth Annual Conference on Rehabilitation Engineering*, pp.55–57, pp.24–28, June 1985.