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DESIGN AND IMPLEMENTATION OF A HEAD-MOTION CONTROLLED ELECTRIC WHEELCHAIR FOR ASSISTIVE MOBILITY IN QUADRIPLEGIC PATIENTS

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Abstract
Quadriplegic individuals, who have lost motor control in all four limbs, require specialized assistive technologies to achieve mobility and independence. This paper presents the design and implementation of a head-motion-controlled electric wheelchair, developed to enhance the autonomy of individuals with severe physical disabilities. The system supports two operational modes: manual and automatic, with the latter utilizing head tilt gestures to control movement. A head-mounted inertial measurement unit (IMU) detects head orientation (forward, backward, left, right), and a microcontroller translates these inputs into movement commands. The mechanical structure comprises a standard wheelchair retrofitted with DC motors and a gear reduction system, while motion control is achieved using a custom-built, relay-based H-bridge motor driver. To enhance motion detection accuracy, sensor fusion is performed using a Kalman filter. Experimental evaluation shows that the wheelchair maintains a stable forward velocity of approximately 0.5 m/s, with smooth bidirectional turning and high command recognition reliability. The paper details the system architecture, hardware and software integration, control strategy, and performance assessment. Potential future improvements include emergency stop features via GSM, variable speed control, health monitoring, alternate input methods, and obstacle detection. The results affirm that the proposed system is a viable, cost-effective assistive solution, significantly improving mobility and quality of life for quadriplegic users.

INTRODUCTION

Mobility is a fundamental necessity for independent living, yet many individuals with paralysis or limb loss are unable to operate conventional wheelchairs using their hands [1], [2]. Quadriplegia, often resulting from spinal cord injuries, stroke, or neurological diseases [3], leaves patients unable to use all four limbs and thus fully dependent on caregivers for movement. According to a study by the Christopher & Dana Reeve Foundation, approximately 5.4 million people worldwide live with paralysis, and about 1 in 50 individuals suffers from paralysis due to causes such as stroke (33.7%), spinal cord injury (27%), and multiple sclerosis (18.6%) [4]. These patients face tremendous challenges in performing everyday activities and require assistive solutions to regain mobility. Motorized electric wheelchairs with joystick control exist, but quadriplegic users cannot operate a joystick[5], and advanced models with alternative controls are often prohibitively expensive for families of lower socio-economic status.

Research in assistive technology has explored various human-machine interfaces to control wheelchairs without hand use. Existing approaches include eye-gaze controlled wheelchairs, which use an eye-tracking camera to direct movement [6], tongue-operated wheelchairs that respond to tongue motions with an intraoral magnet and sensor [7], head gesture or tilt-controlled wheelchairs, voice-command driven systems, and even brain–computer interfaces (EEG-based control) [8]. Each of these has limitations: vision-based systems can fail in poor lighting, tongue-based systems are invasive (requiring a magnetic tongue stud) and incompatible with MRI procedures, voice-based control is unusable by mute patients and error-prone in noisy environments, and EEG-based control is highly user-specific and prone to variable brain signal patterns [9]. Despite these challenges, such innovative interfaces aim to grant paralyzed users more autonomy. Recent work has even combined multiple inputs; for example, a hybrid control system allowing both voice commands and head tilts was developed to let quadriplegic patients choose their preferred method of control [10].

This paper presents a simple, low-cost, head-motion-controlled wheelchair designed for quadriplegic individuals. The system interprets natural head tilts forward, backward, left, and right as movement commands, enabling intuitive, hands-free navigation without special training or calibration. Built using a standard wheelchair with added IMU sensors and a microcontroller, the system aims to restore independent indoor mobility. We describe the hardware setup, control algorithm, and test results, and outline potential improvements for enhanced safety and functionality.

Literature Review

Recent advancements in eye-tracking technologies have enhanced hands-free wheelchair control systems. Xu et al. [11] developed a deep learning-based system that uses a monocular camera and attention mechanisms to translate gaze direction into navigation commands. Similarly, Higa et al. [6] proposed an intelligent eye-controlled wheelchair using a one-dimensional CNN and LSTM network to estimate visual intentions. These systems address challenges like sensitivity to lighting and eye fatigue through better gaze detection algorithms and real-time processing.

The Tongue Drive System (TDS) has seen refinements with non-invasive magnetic sensor arrays mounted on custom dental retainers [8]. These detect magnetic field variations caused by tongue movement. A 2022 study evaluated a tongue-operated robot for comfort and control reliability in mobility applications [7], noting it as a viable option for individuals with severe physical impairments. While still somewhat intrusive, modern TDS systems reduce discomfort compared to earlier models that required adhesives or piercings.

There is a shift towards hybrid systems that incorporate multiple input modes for redundancy and adaptability. Jiang et al. [12] integrated EEG brain signals with smart sensing for real-time control. Combining speech, gesture, and neural inputs is becoming the norm in assistive mobility tech to ensure inclusivity and reliability.

While recent assistive wheelchair technologies explore AI-based and multi-modal controls, they often require high computational power, internet connectivity [13], or invasive setups limiting practicality and affordability. In contrast, our system offers a low-cost, standalone solution using head-motion detection via an IMU and sensor fusion through a Kalman filter. It avoids complex dependencies, enabling reliable, hands-free control for quadriplegic users, and demonstrates high accuracy, ease of use, and potential for real-world application. In the next section the detailed description of our research is presented.

System Architecture

The proposed head-motion-controlled wheelchair comprises two main components: a head-mounted sensor module[14] and an on-board wheelchair drive system[15]. As shown in Figure 1, an IMU sensor mounted on the user's headgear measures head orientation using a 3-axis accelerometer and gyroscope. This data is sent to an Arduino Mega 2560 microcontroller on the wheelchair, which processes tilt gestures pitch (forward/backward) and roll (left/right) to determine movement commands.

Based on predefined angle thresholds, the microcontroller signals an H-bridge motor driver to activate two rear-wheel DC motors for directional control. The system supports both manual and automatic modes: in manual mode, a switch disengages the motors for caregiver-assisted movement; in automatic mode, the wheelchair responds continuously to head movements for hands-free navigation.

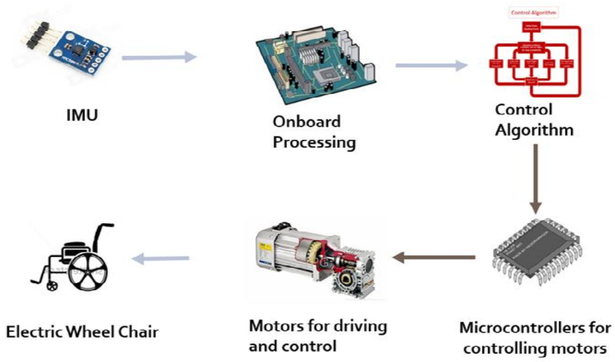

Fig. 1. Functional diagram of the head-motion interface, control logic, and motor driver for the wheelchair.

The system architecture is modular, comprising three main components: the sensing module (IMU on a headset), the processing module (microcontroller)[16], and the actuation module (motor drivers and motors). This separation allows easy upgrades to different sensors or motors can be used without changing the core logic. Currently, wired communication is used between the IMU and microcontroller for reliability, though wireless options like Bluetooth can be added for user comfort. Power is supplied via onboard batteries, delivering regulated 5V for logic and higher voltage for motors. As shown in Fig. 1, tilt data from the IMU is processed by the microcontroller, which sends commands to the motor drivers. A basic safety feature is implemented: tilting the head backward triggers an immediate stop. Additional features like wireless control or hardware cutoffs can be integrated later for enhanced safety.

System Model

The system model includes both mechanical and electronic components. A standard steel-frame manual wheelchair with large rear and small front wheels (Fig. 2) was used as the base, preserving features like brakes, footrests, and posture support. To motorize it, two 24V

Fig. 2. Standard manual wheelchair integrated with motor hardware, preserving full manual functionality.

permanent-magnet DC motors [17] as shown in Fig. 3 were mounted beneath the seat to drive the rear wheels. These motors offer a good balance of torque and speed, achieving around 150–200 RPM under load, resulting in a forward speed of approximately 0.5 m/s. Gears attached to the motor shafts transmit power to the wheels, enabling smooth motion while maintaining affordability and structural reliability.

To motorize the wheelchair, two identical 24V permanent-magnet DC motors (Fig. 3) were installed to drive the rear wheels. These motors offer a balance of torque and speed, enabling smooth motion and the ability to climb slight inclines. With a load speed of 150–200 RPM, they provide a forward velocity of approximately 0.5 m/s. Each motor is gear-coupled to a rear wheel and mounted under the seat, aligned with the axle for efficient power transmission.

Fig. 3. 24V permanent-magnet DC motor driving the wheelchair's rear wheels, with bidirectional control via the H-bridge.

Gear Reduction Assembly: To increase torque at the wheels and effectively utilize motor power, a 1:2 gear reduction mechanism [18] was implemented. A small pinion gear on the motor shaft meshes with a larger gear on the wheel axle, doubling torque and halving rotational speed. This setup ensures the wheelchair can carry the user's weight and overcome rolling resistance. Helical gears were chosen for their smooth, quiet operation, despite slightly reduced efficiency and axial loads. The gears are housed to protect them from dust and retain lubrication. During testing, the assembly allowed smooth starts, stops, and turns, while also preventing free-rolling downhill. Steel helical gears from automotive parts kept costs low.

Fig. 4. Two-gear reduction assembly diagram, with a 2:1 speed reduction and increased torque from the motor to the wheel axle.

The gears are housed to protect them from dust and retain lubrication. Figure 5 shows the gear assembly installed on the wheelchair, highlighting the meshing of metal gears. Grease was applied to minimize friction and wear. During testing, the gear reduction allowed smooth movement without motor stalling and prevented free-rolling downhill by providing resistance when power is off. The steel helical gears, salvaged from automotive parts, kept costs low.

Fig. 5. Installed gear assembly with helical gears, transmitting power from the motor to the wheel and increasing torque for smooth operation.

The motor mounting and drive system were custom fabricated with a metal motor plate attached beneath the wheelchair seat. The plate holds the motors in position, with shafts aligned to the rear wheel axles. Steel brackets welded to the frame provide a sturdy fixture. Figure 6 shows the motor installation, including the motors, part of the battery pack, and wiring to the motor driver. The motors are placed low to avoid interfering with the user's legs or ground clearance, keeping the center of gravity low for improved stability.

Fig. 6. Bottom view of the motor installation. Two DC motors are mounted on a metal plate under the seat, connected to the H-bridge driver for efficient power transfer

Two 12 V lead-acid batteries, connected in series for 24 V, are mounted on the motor plate (Fig. 6). They also power a 5 V regulator for the microcontroller and IMU sensor. While cost-effective, lead-acid batteries add weight; future versions may use lighter lithium-ion batteries.

Electronic Control Unit: The electronic hardware is centered around the Arduino Mega 2560 [19], chosen for its ample I/O pins and memory to interface with sensors and implement the control algorithm. It reads IMU data, executes control logic, and drives the motors via a custom H-bridge made of high-current relays and diodes. The relays, capable of handling up to 10 A, switch the motor direction or halt it. The Arduino outputs control signals to activate the relays, which in turn control the motors. This relay-based H-bridge is cost-effective but slower than solid-state alternatives. The Arduino and relay circuits are mounted on a small wooden board beneath the wheelchair seat, with power supplied by the two batteries. Figure 7 shows the control layout with the Arduino board, IMU, and relays neatly arranged. A master toggle switch is included for system power, and manual mode allows the wheelchair to be pushed freely.

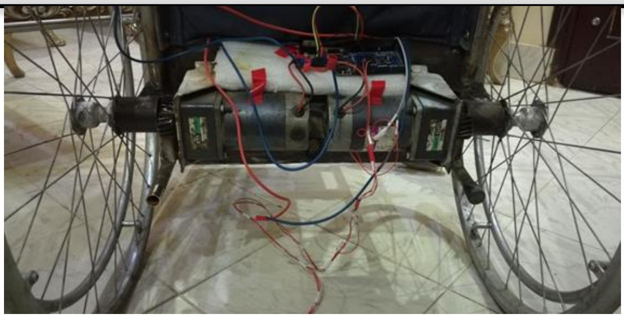

Fig. 7. Electronic control unit with Arduino Mega 2560, IMU sensor, relay circuits, and dual 12 V batteries for power.

Head-Mounted Sensor (IMU): The system uses a GY-87 IMU sensor [20] to detect head movements. This 10-degree-of-freedom sensor includes a 3-axis accelerometer and a 3-axis gyroscope, which are used to determine head orientation. The accelerometer detects gravity-induced tilt, while the gyroscope tracks angular velocity. The sensor data is processed using a Kalman filter to provide a stable tilt estimate. Mounted on a headband, the sensor communicates with the Arduino via I²C. It is sampled at 50 Hz, sufficient for detecting head tilts. Calibration aligns the sensor's zero tilt with the user's resting head position, ensuring intuitive control. The lightweight design makes the sensor unobtrusive and adjustable for different users.

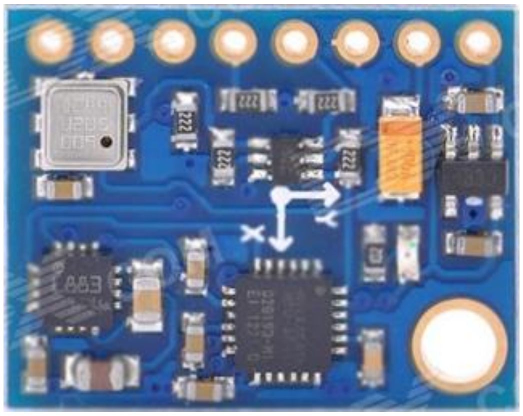

Fig. 8. The GY-87 IMU sensor module (2 cm), featuring a 3-axis accelerometer and gyroscope for real-time head tilt measurement in both directions.

Microcontroller and Sensor Integration: The Arduino Mega interfaces with the IMU using I²C to read data from the accelerometer (ADXL345) and gyroscope (L3G4200D)[21]. We use Arduino libraries to retrieve acceleration and angular velocity readings, applying basic filtering (e.g., moving average for accelerometer data) and scaling for gyroscope data. The sensor sampling is synchronized with the control loop, running at ~50 Hz. Communication over I²C introduces minimal delay, and the loop is timed with a 20 ms delay per cycle. Figure 9 shows the Arduino Mega connected to the GY-87 sensor during testing. Final connections were secured with soldered wires or locking connectors to prevent disconnection during motion. The system captures head motion and translates it into wheelchair movement through a control algorithm.

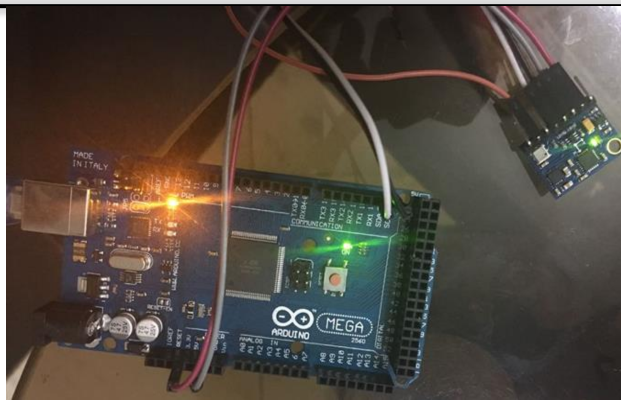

Fig. 9. Arduino Mega connected to the GY-87 IMU sensor (right), reading head tilt data via I²C for motor control decisions.

Control Algorithm and Software Design

The control algorithm on the Arduino interprets IMU data to control the wheelchair's motors. It recognizes four main head gestures:

1. **Forward:** Head tilt forward (chin down).
2. **Stop:** Head tilt backward (chin up).
3. **Turn Right:** Head tilt right.
4. **Turn Left:** Head tilt left.

An implicit command is to continue the current motion when the head returns to neutral, reducing user effort (e.g., tilting forward starts movement, and the wheelchair continues until a stop command is given).

Signal Processing and Tilt Determination:

Raw accelerometer and gyroscope data are fused using a Kalman filter to determine the head's tilt angles. Accelerometer data alone is noisy due to vibrations, while gyroscope integration causes drift. The Kalman filter combines both, providing stable tilt estimates. The algorithm checks these angles against predefined thresholds (18° for forward/backward, 30° for left/right) to avoid accidental triggers. Hysteresis is used to prevent oscillations near the thresholds.

Control Logic:

The system starts in a "rest" state when the head is level. Tilting the head triggers corresponding movements: forward, backward, left, or right. The system checks the tilt thresholds and maintains the motion state until a new command is given, similar to a state machine but with instant checks.

```
loop:
read sensor (get tilt_angle_forward, tilt_angle_side)
if tilt_angle_forward > forward_threshold:
command_forward()
else if tilt_angle_forward < -backward_threshold:
command_stop()
else if tilt_angle_side > right_threshold:
command_turn_right()
else if tilt_angle_side < -left_threshold:
command_turn_left()
else:
// no new command, maintain current state
```

To differentiate a Stop command from returning to neutral, a backward head tilt (about 18° upward) was chosen as an explicit stop command. This avoids confusion with normal posture and ensures safety. After tilting forward to move, the wheelchair continues until the user tilts back to stop. Similarly, for turns, the wheelchair keeps turning until the head is straightened or a new command is given. This scheme is intuitive, with the user nodding forward to go and back to stop, similar to a yes/no gesture.

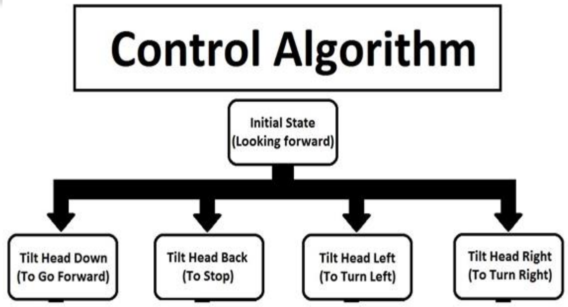

Fig. 10. Control algorithm logic for interpreting head tilt commands. Tilting the head initiates forward motion, stop, or turns, with actions continuing until a stop command is given.

The microcontroller controls the motors based on head tilt commands. For forward motion, both motors move forward. To turn right, only the left motor runs while the right stops, and vice versa for a left turn. Stopping halts both motors, with relays braking them for quicker response. A short delay and debouncing help prevent false triggers from quick or accidental head movements. Commands are confirmed only if the tilt lasts around 200 ms, filtering out brief or unintended motions.

Figure 11 presents the algorithm flowchart. It begins by reading GY-87 IMU data and calculating tilt angles using a Kalman filter. Based on the tilt direction, the system issues a move, stop, or turn command. The loop then repeats. If no new input is detected, the wheelchair maintains its last action. A safety timeout stops the wheelchair if the head stays in a neutral position for too long while moving.

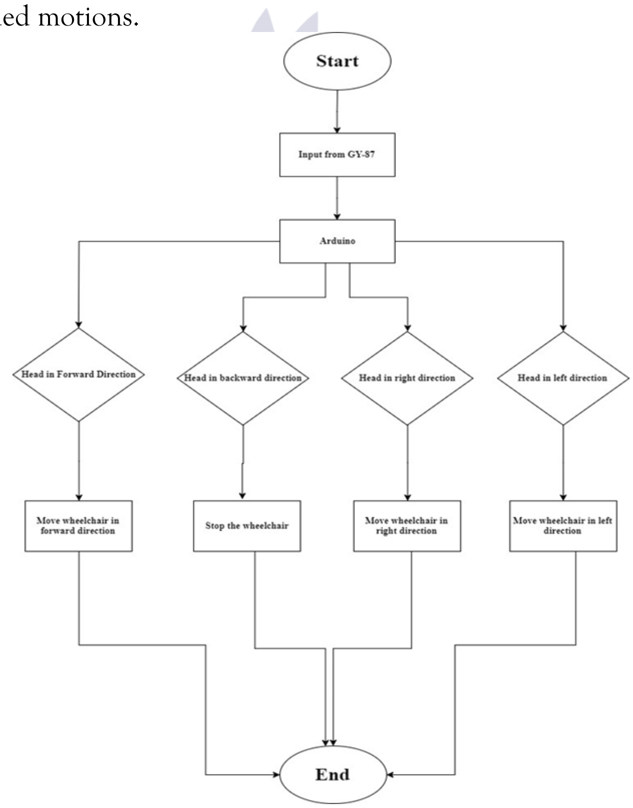

Fig. 11. Flowchart of tilt-based motion control implemented on the microcontroller.

During development, we tested the control software with able-bodied users simulating quadriplegic use (hands still, only head movement) to fine-tune responsiveness. The system handles quick transitions—like from forward to left tilt—smoothly, enabling curved paths. Rapid forward-then-backward tilts result in a brief lurch, like a joystick tap, which was acceptable.

The algorithm is a simple rule-based system enhanced by sensor fusion via a Kalman filter [22], running at ~50 Hz for low-latency response (~0.02 s). Filtering significantly improved accuracy by reducing gyroscope drift and false triggers. After tuning, the system reliably distinguished intentional head movements from noise, as supported by experimental results. The table. 1 shows the hardware names with their specifications used in the system model.

Table 1. Hardware Components and Specifications

Equipment Name	Model/Details
Wheelchair Base	Standard steel-frame manual wheelchair (custom motorized)
Motors	24V permanent-magnet DC motors (2 units, 150–200 RPM)
Gear Assembly	1:2 helical gear reduction (steel, automotive parts)
Motor Driver	Custom relay-based H-bridge
Batteries	Two 12V lead-acid batteries (series for 24V)
Microcontroller	Arduino Mega 2560
Sensor	GY-87 IMU (10-DOF, 3-axis accelerometer + gyroscope)
Wiring	IC communication, soldered/locking connectors

Results

After assembling the hardware and programming the control algorithm, the head-controlled wheelchair was tested indoors on a smooth surface. Key performance metrics included speed, responsiveness, turning ability, and command accuracy. Tests were conducted with a seated user to simulate real conditions, and in some cases, the wheelchair was pushed manually for specific measurements.

Forward Motion Performance:

In forward motion trials, users tilted their head forward to move and backward to stop. Figure 12 shows a typical distance-time plot, where the wheelchair quickly accelerates and then moves at a steady speed of about 0.5 m/s (1.8 km/h) a safe, comfortable indoor pace. The slight curve at the start reflects the acceleration phase, after which the motion becomes linear, indicating constant velocity. Upon receiving a stop command, the wheelchair halts within a short distance, ensuring safety in indoor use.

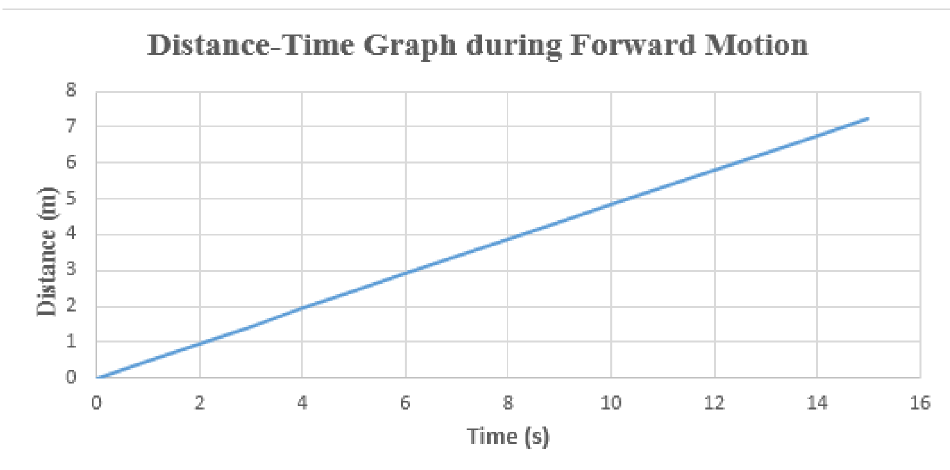

Fig. 12. Distance–time graph showing steady forward speed (~0.5 m/s) after initial acceleration.

We measured the wheelchair's speed over time using smartphone GPS and by differentiating distance data. As shown in Fig. 13, speed quickly reaches ~0.5 m/s within 1–2 seconds and then stays nearly constant until the stop command is issued. A dip around 14–15 s marks deceleration as the user tilts their head back. This consistent cruising speed results from the open-loop control and motor characteristics, with no PWM used—motors run at full voltage. Despite its simplicity, the system provides stable, predictable motion, which users adapted to easily. Future versions could add closed-loop speed control for more flexibility.

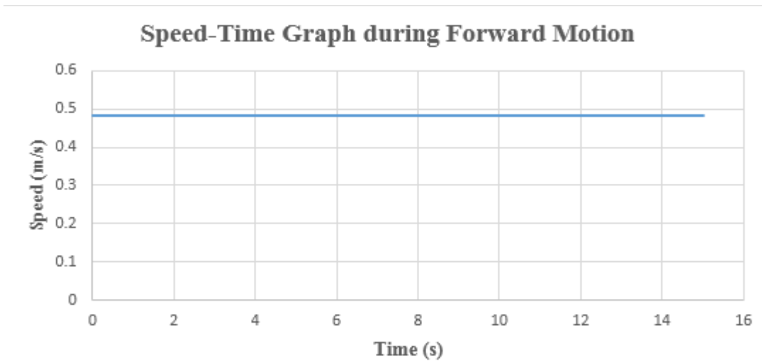

Fig. 13. Speed–time graph showing quick rise to ~0.5 m/s and steady motion until stop

Turning Performance: We tested right and left turns by tilting the head and holding the position. The wheelchair pivots by rotating one wheel while stopping the other. Angular displacement was measured using a protractor, stopwatch, and gyroscope. Figure 14 shows a steady, nearly linear rotation over time, with about 2200° (6 full turns) in 15 seconds equivalent to 0.4 revolutions per second. A full 360° turn took roughly 2.5 seconds, demonstrating a relatively high, constant turning speed suitable for sustained pivoting.

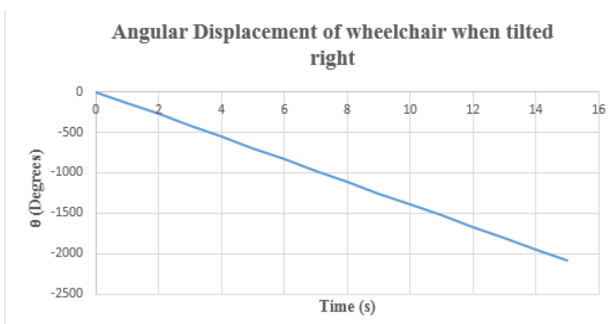

Fig. 14. Angular displacement during right turn showing steady rotation to −2200° in 15 seconds.

From the angular displacement slope, the turning rate is about 144°/s (0.4 rev/s). Figure 15 shows angular velocity fluctuating slightly around −0.4 rev/s during the right turn, stabilizing quickly without uncontrolled acceleration. The negative sign indicates direction. This steady rate allows users to stop turning by centering their head and issuing a stop command. The turning method—one wheel moving, the other stopped—enables in-place rotation with a small turn radius, ideal for tight spaces. Though faster, testers found the speed manageable; future designs could add PWM control or alternate turning methods for safer, slower turns.

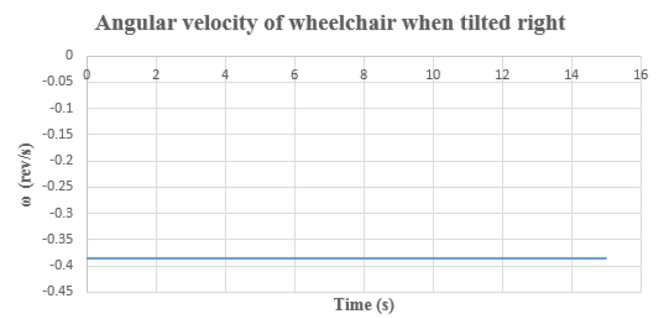

Fig. 15. Angular velocity during right turns steady at about –0.4 rev/s (144°/s).

Analogous tests for left turns show angular displacement over time as a mirror image of right turns. Figure 16 displays rotation in the opposite direction, reaching about +2400° in 15 seconds, confirming a similar turning rate slightly above 0.4 rev/s. This symmetry indicates balanced hardware and control for both directions. The linear plot suggests uniform turning motion.

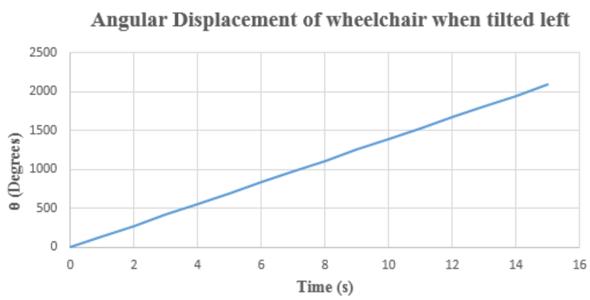

Fig. 16 Left tilt angular displacement rises to +2200–2400° in 15 seconds, showing symmetric turning performance.

During left turns (Fig. 17), the angular velocity was about +0.4 rev/s, similar to right turns. Although the speed seems high in degrees per second, the wheelchair's inertia and friction provide natural damping, resulting in a smooth rotation. Future versions could reduce turning speed by adjusting motor control or using differential wheel control.

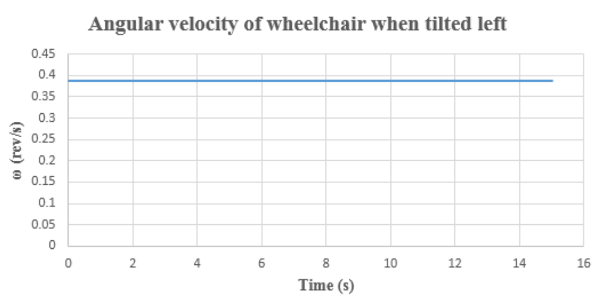

Fig. 17 Wheelchair turns left at about +0.4 rev/s, similar to right turns, demonstrating balanced performance.

System Responsiveness

The measured response delay between head tilt and wheelchair motion was approximately 0.1-0.12 seconds (100-120 ms), accounting for sensor latency, computation, and actuation. Although subjectively the response felt immediate, analysis showed that once the tilt angle crossed the threshold (e.g., ~20° forward), the motor activated within ~0.11 s.

Delays remained consistent across different tilt angles, with minimal variation. The primary contributor to this delay was the relay activation time and the motor overcoming static friction. The control algorithm itself responded within one loop cycle (~0.02 s) after threshold crossing.

From a user perspective, this short delay is barely noticeable and does not impair control. In fact, a slight delay helps prevent unintended activation due to small or accidental head movements. Figure 18 summarizes the delay across various tilt angles.

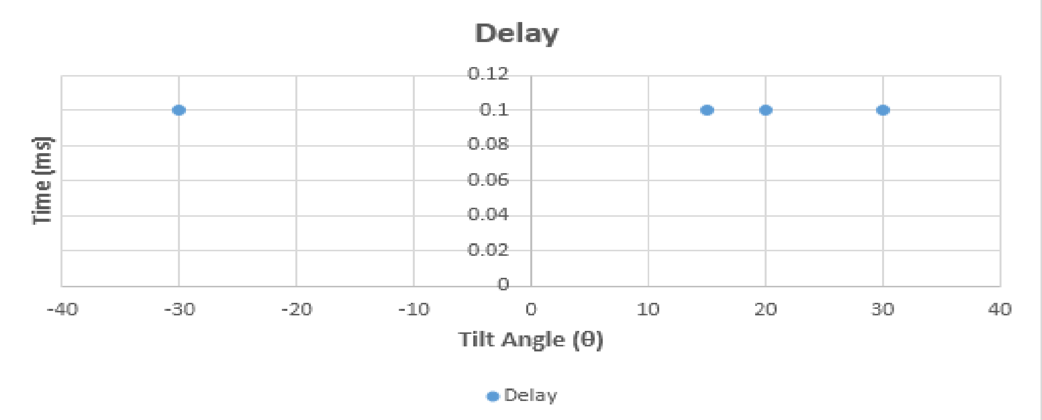

Fig. 18. Control delay (~0.1 s) vs. head tilt angle (θ), showing fast response once the threshold is crossed.

Reliability (Command Recognition and Fail-Safe)

Each command (forward, stop, left, right) was tested 50 times under varying conditions to assess recognition accuracy. A failure was defined as either missing a valid head tilt or triggering movement without sufficient tilt.

Approximately 80% of forward command trials exhibited zero failures. Most other sets showed only 1–2 minor errors, typically due to borderline or unclear head tilts. Figure 19 presents a histogram of failures per 50 commands, with the highest frequency at zero failures.

Overall recognition success ranged from 95–100%. No unintended movements occurred in a stationary state, and the Kalman filter effectively mitigated false triggers caused by motion or bumps. Encouraging clear and deliberate head tilts further enhances system reliability, confirming its suitability for practical use.

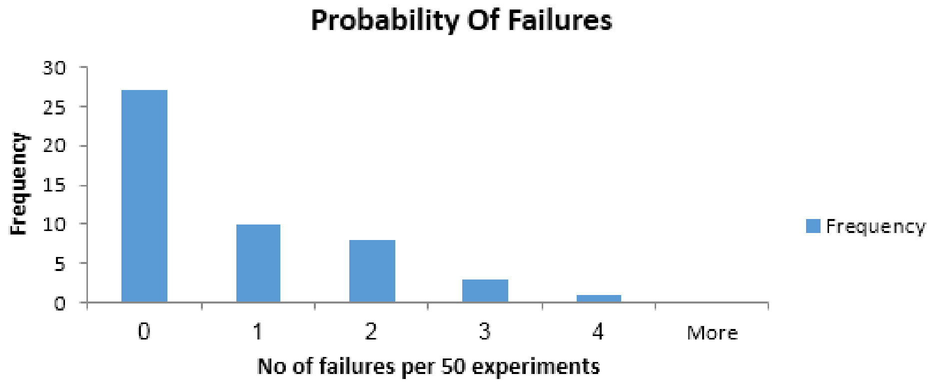

Fig. 19. Most command sets had 0–2 failures out of 50, demonstrating a high success rate for the head-motion control system

It is worth noting that these experiments were done in a controlled indoor environment. Factors such as uneven terrain, sudden movements of the user's body (not just head), or electromagnetic interference could potentially affect performance. However, the use of a physical relay-based control (as opposed to sensitive electronics) and the simplicity of the sensor likely make the system robust to moderate environmental disturbances. We also assessed the physical strain on the user: since the user only needs to tilt their head by a moderate angle for a short duration to issue a command, the system is comfortable to use. It effectively reduces the physical effort compared to, say, continuously pushing a manual wheelchair or even continuously blowing into a sip-and-puff device. Users reported that the head motions felt natural after a few practice runs – akin to nodding or leaning to indicate where they want to go.

Conclusion

This work presents a head-motion controlled wheelchair system designed for individuals with quadriplegia and severe motor impairments. An IMU sensor mounted on the user's head detects intentional tilts and translates them into movement commands via an Arduino-controlled, motorized wheelchair retrofitted with a relay-based H-bridge drive.

Using a Kalman filter to fuse accelerometer and gyroscope data ensures stable, real-time head orientation tracking. The system supports four core commands – forward, stop, turn left, and turn right – with low latency and high accuracy, allowing users to return to a neutral head position after issuing a command.

Testing confirmed continuous, error-free operation, with a forward speed of ~0.5 m/s, a turning rate of ~0.4 rev/s, and a reaction time of ~0.1 s. The system performed reliably in indoor environments, with high command recognition and minimal false triggers.

This low-cost, easily reproducible solution offers practical mobility and greater independence for users. Future enhancements include obstacle detection, emergency stop, multimodal controls, and smart home integration.

The proposed system is a promising step toward accessible assistive mobility, combining embedded systems and rehabilitation engineering to improve quality of life for users with severe disabilities.

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