GaitWay: Development and Testing of a Novel Ambulation Device for Gait Rehabilitation

COLLEGE OF ENGINEERING
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VIRGINIA TECH

Leah Thomas, BS[1]; Andrew Schroeder, MD[1,2], Jenna Altaii, BS[1], Robert Stone, Ph.D[1], Arnold Salzberg, MD, FACS[2]
Chris Arena, Ph.D[1]

[1]Virginia Polytechnic Institute and State University, [2]Department of Surgery, Carilion Clinic - Virginia Tech Carilion School of Medicine

CARILION CLINIC
VTC Virginia Tech Carilion School of Medicine

INTRODUCTION

- 7-12% of the U.S. population experiences significant ambulation limitations [1,2].
- Immobility contributes to poor patient outcomes, with daily bed rest causing 1.3–3% decline in muscle strength in healthy individuals [5].
- Across multiple patient populations, early ambulation has been shown to decrease length of hospital stays and post-operative risks [6,7,8,9].
- Ambulation training requires significant resources and qualified staff, which are often limited.
- Several existing devices offer ambulation training but have limitations, such as dependence on personnel, lacking range of assistance, lack of biofeedback, or cannot be used supine.

USER NEEDS & DESIGN INPUTS

User Need	Design Inputs	Source
Device promotes activation of muscle groups in the same pattern as walking	The device shall activate key lower-limb muscle groups in a similar pattern to normal gait cycles, measured by EMG	SME
Device improves muscle strength	Device shall increase lower-limb muscle strength (measured as peak torque) by the expected strength increase to walk from PT	SME
Device includes features that focus on cardio	Device shall elevate/sustain user's heart rate to 50-85% of age-predicted max heart rate for a min of 10 minutes per session	[13]
Device collects objective quantitative data of patient progress	Sensor data will be collected for 30 mins of treatment and compared to proper gait cycle to calculate "deviation score"	SME
Device provides range from fully passive to fully active	Device may be operated with an adjustable assistive force proportional to the force exerted by patients	SME

DEVICE DESIGN

Symbol	Description
W_l	Known weight of the user's leg.
F_a, F_h, F_d, F_p	Force sensor readings from the anterior (a), heel (h), distal (d), and proximal (p) foot regions.
$F_{u(x,y)}$	Estimated user-applied force at position (x, y), computed from W_l and foot sensor data.
pos_x, pos_y	Estimated user-applied force at position (x, y), computed from W_l and foot sensor data.
$[E_u]$	User-specific force-position map generated from F_u and position data.
$[E_e]$	Normalized expected force-position map, based on user height and weight.
$A_{(x,y)}$	Assistive multiplier at position (x, y), proportional to E_e / E_u
$F_{f(x,y)}$	Final calculated force applied by the motor at position (x, y).

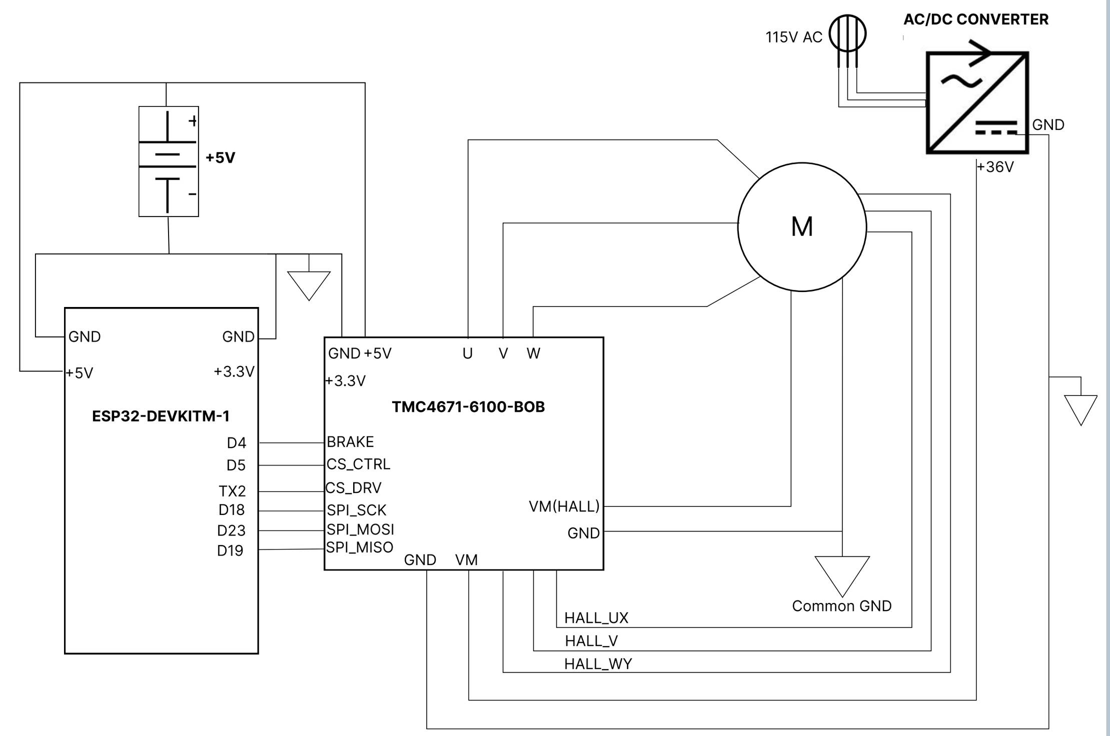

Fig. 1: Circuit Diagram

$$[E_u] = \begin{bmatrix} pos_{x1} & pos_{y1} & Fu_{x1} & Fu_{y1} \\ pos_{x2} & pos_{y2} & Fu_{x2} & Fu_{y2} \\ \dots & \dots & \dots & \dots \\ pos_{xn} & pos_{yn} & Fu_{xn} & Fu_{yn} \end{bmatrix} \quad [E_e] = \begin{bmatrix} pos_{x1} & pos_{y1} & Fe_{x1} & Fe_{y1} \\ pos_{x2} & pos_{y2} & Fe_{x2} & Fe_{y2} \\ \dots & \dots & \dots & \dots \\ pos_{xn} & pos_{yn} & Fe_{xn} & Fe_{yn} \end{bmatrix} \quad A_{(x,y)} = \begin{matrix} Fe_{xn} / Fu_{xn} \\ Fe_{yn} / Fu_{yn} \end{matrix}$$

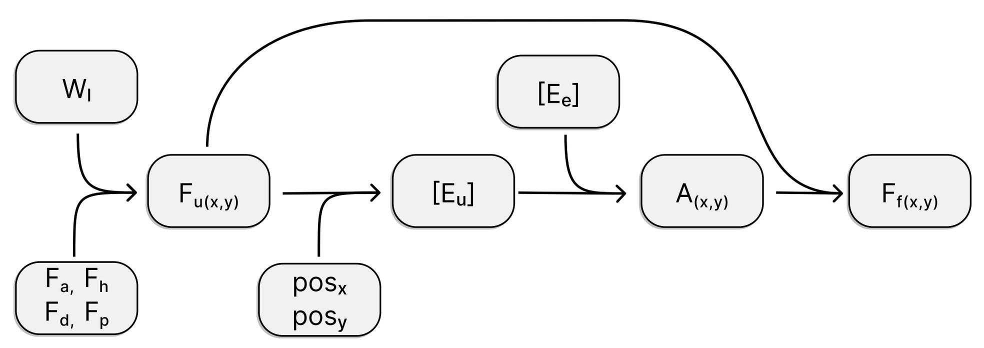

Fig. 2: System Architecture

Vertical Force Formula

Used to calculate force needed for vertical component motor response to heel force sensor data

$$F_f = W_l + A(W_l - F_h)$$

The GaitWay provides a range of exercises, biofeedback, and customized gait training for patients with limited mobility.

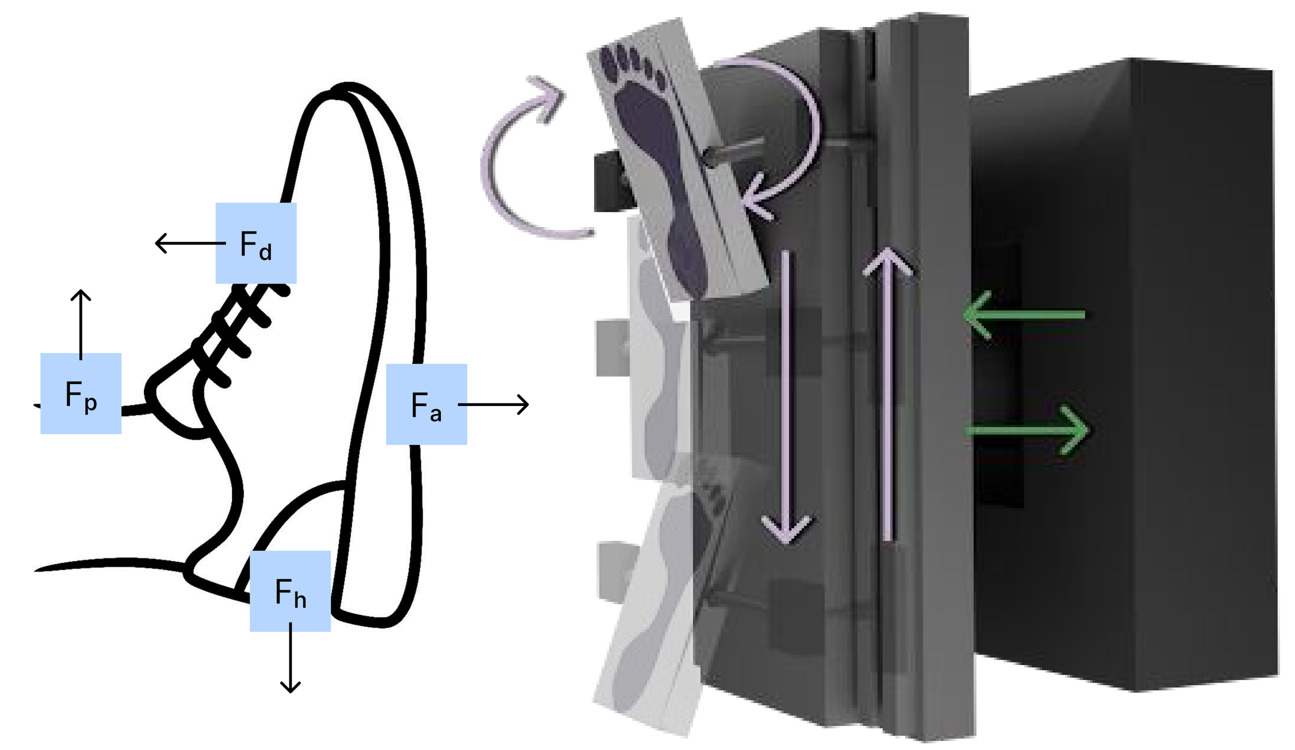

Fig. 3: Force Sensor Placement and Proposed Device Model

EXPERIMENTAL SETUP

- Prototype was vertically mounted
- Motor driven belt pulley system used to lift plate along a linear rail.
- Incremental weights were affixed to the plate, and displacement over time was recorded using a high-speed camera
- Velocity and acceleration were calculated from displacement data using second-order polynomial regression.
- Motor current was measured using a current clamp, capturing both average and peak current during dynamic loading conditions.
- Tests were conducted under varying torque digital motor control value (DMCV) parameters and different weight loads to observe effects on motor current draw, duration of lift, and mechanical output.
- Collected data (motor current, velocity, and acceleration) analyzed under varied conditions

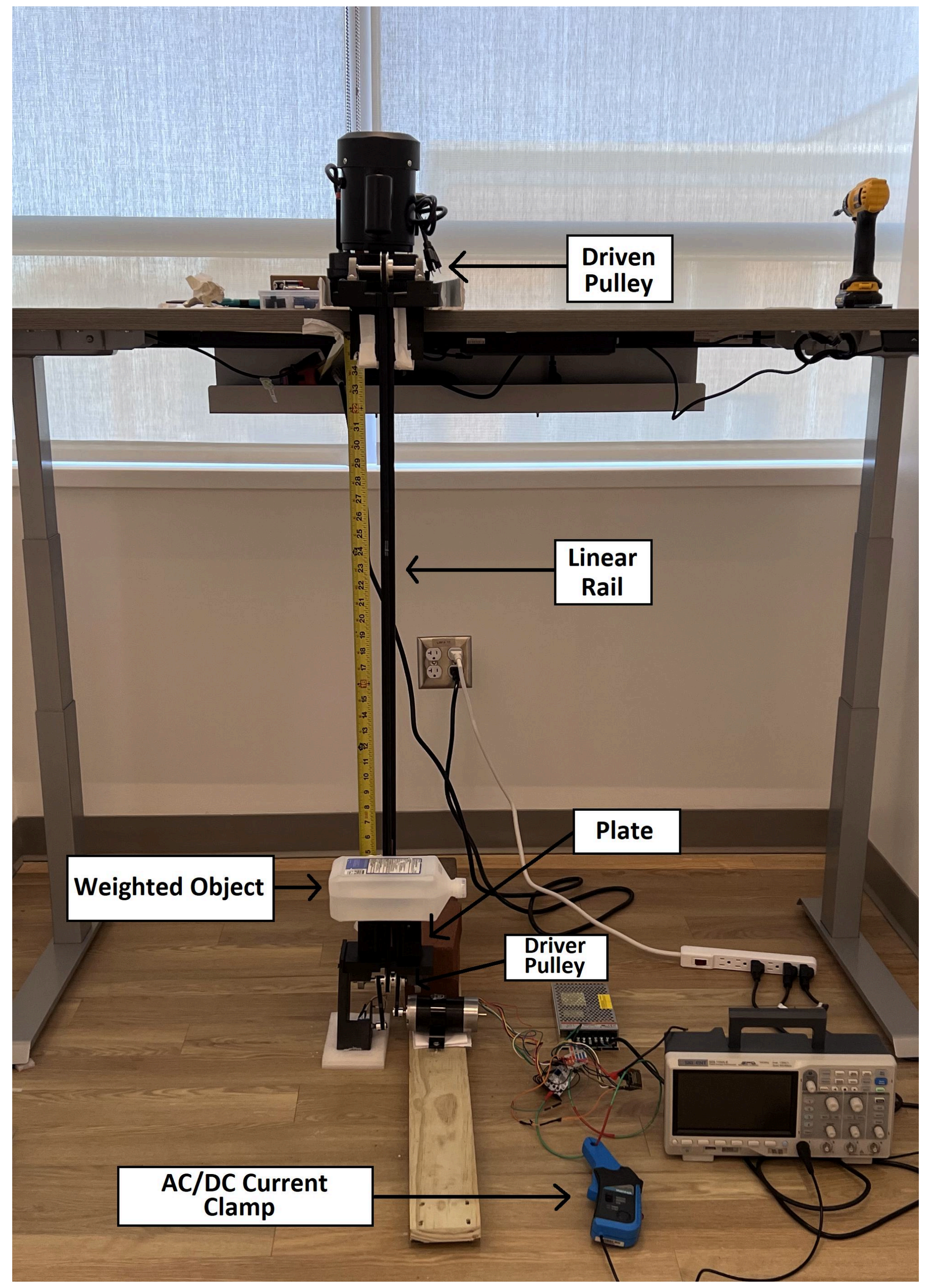

Fig. 4: Experimental Test Setup

RESULTS

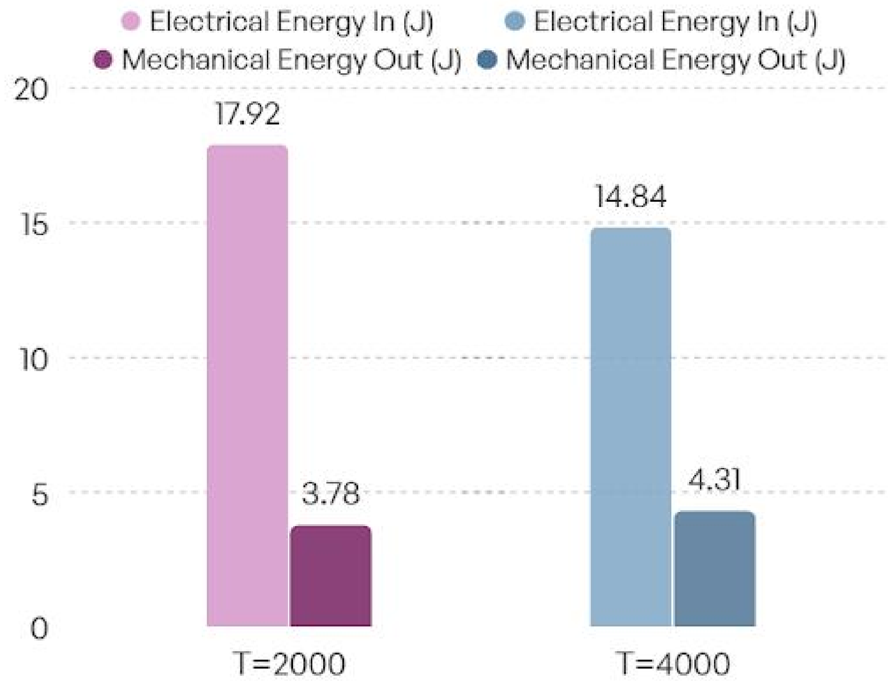

Fig. 5: Calculated Electrical Energy and Mechanical Energy Consumption

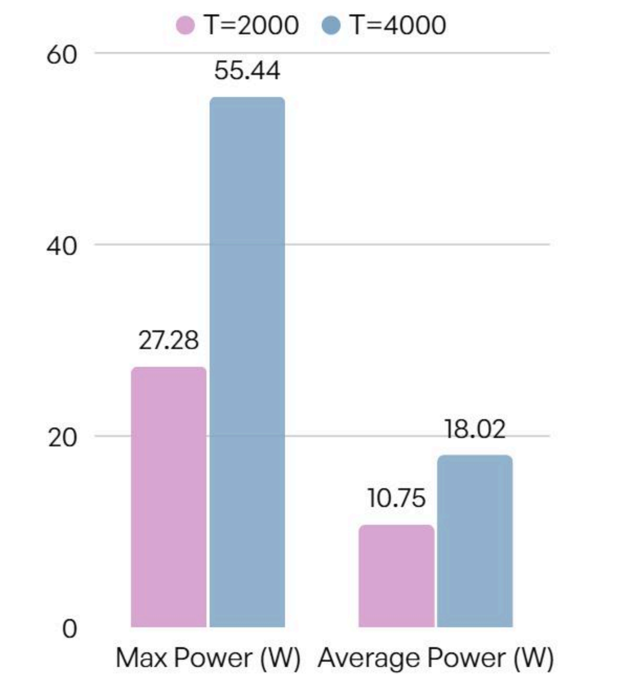

Fig. 6: Max Power and Average Power at Different DMCVs

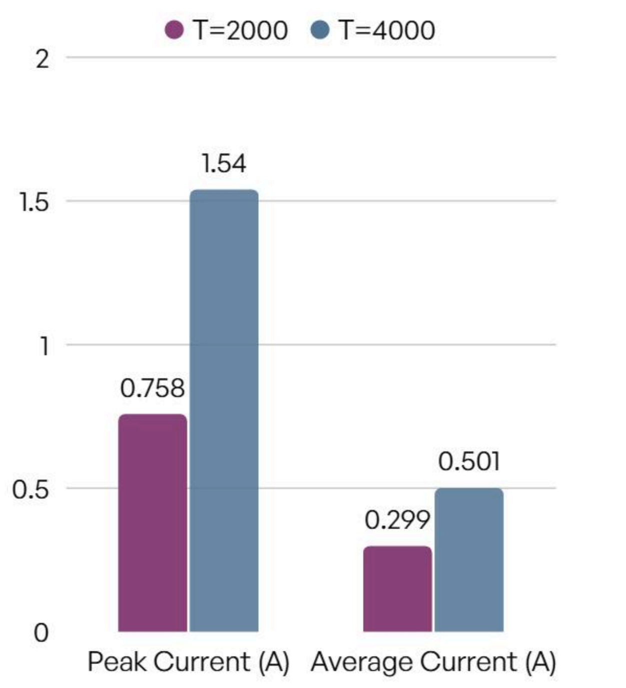

Fig. 7: Peak Current and Average Current at Different DMCVs

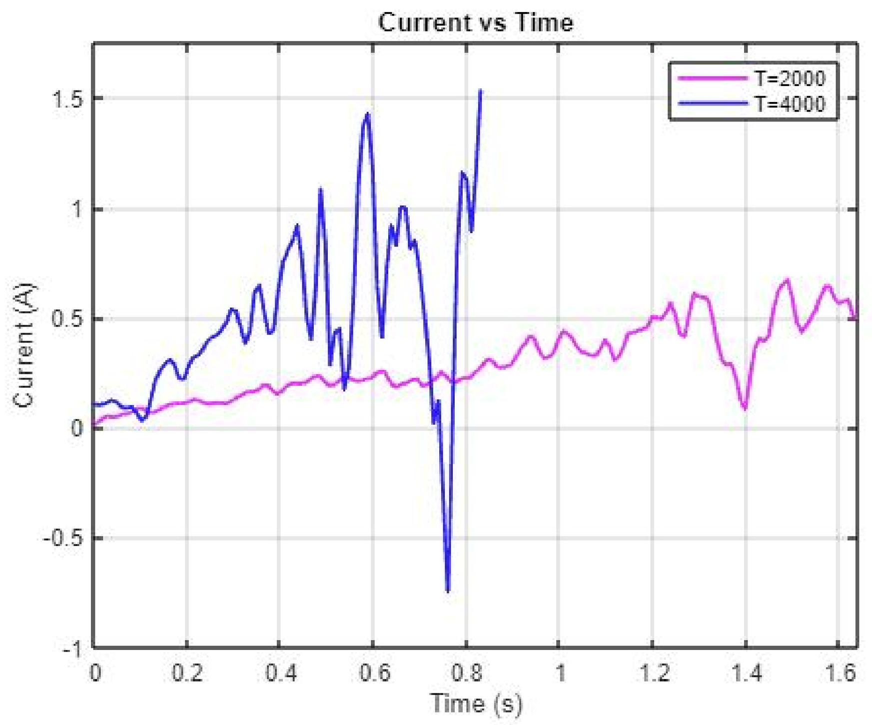

Fig. 8: Measured Current Draw at T=2000 and T=4000

DMCV	Weight (lb)	Avg Acceleration (m/s²)	Force (N)
2000	1	0.5749	4.7105
	2	0.2440	9.1208
	3	0.0695	13.3492
4000	1	2.2833	5.4854
	2	1.7646	10.5003
	3	0.4911	14.0187

Fig. 9: Mapping of Weight and DMCV to Force

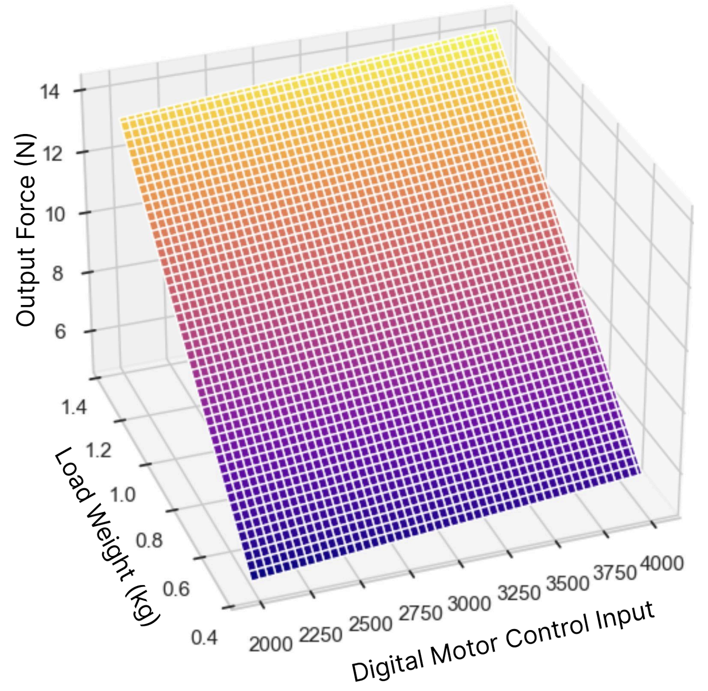

Fig. 10: Modeling Output Force: Effects of Load and Motor Input Parameters

DISCUSSION

Limitations

- Variability in belt tension may have influenced results.
- Calculations assumed a frictionless, drag-free system.
- Limited position and timing data may have introduced calculation inaccuracies.

Conclusion

- Higher DMCVs improve efficiency, increase instantaneous power and current, and reduce total energy consumption.
- Mathematical function maps digital input and load weigh to force output, allowing control of assistive force delivery.

Future Work

- Integrate gear ratio to accommodate heavier weight
- Develop high-fidelity prototype for human-subject testing
- Create predictive mathematical model to optimize the system's performance and calculate assistive forces
- Conduct pilot study on bed-bound patients to evaluate clinical efficacy.

Sources