Design of an Actuated Upper Limb Orthosis for Patients with Neurodegenerative Diseases

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***Abstract*—Amyotrophic lateral sclerosis (ALS), Parkinson's Disease, and spinal muscular atrophy (SMA) are some of the prominent neuromuscular illness. Patients with these diseases experience problems with daily activities such as swallowing food or slow movement, and limb stiffness. This extends to the upper limbs, with them being incapable of sustaining the weight of their arm to perform activities of daily life (ADL). We propose a novel orthosis design that provides external actuation to a patient's arm, to enhance functional reach capability. The proposed orthosis design is a 5 DoF fully actuated upper limb exoskeleton attached to the patient's wheelchair, which helps actuate the shoulder and elbow joints. Electromyography (EMG) data from the upper arm and forearm is utilized in the control loop of the system to govern the appropriate motor torques. The orthosis was designed and simulated for a reach action for bringing a glass to the patient's mouth. The report concludes with a feasibility analysis of this device, with initial analysis showing that it may be a viable and beneficial solution for patients of neurodegenerative illnesses, as well as those suffering from the aftereffects of a stroke.**

I. Introduction

A. Motivation

Around 5,000 new patients in the United States are diagnosed with [1] Amyotrophic lateral sclerosis (ALS) each year, and Parkinson's disease affects approximately 10 million people worldwide [2]. Patients with these conditions have difficulties performing functional motor activities like speaking, eating, walking, and exercising their limbs [4]. Tremor is another common symptom of Parkinson's Disease. Drugs have minimal efficacy in decreasing tremors, and most patients are either ineligible for surgical therapy or are resistant to medication. In such a case, a bio-mechanical solution with the ability to suppress tremors is a better technique as it is a meaningful option for the patients. The target group of patients is those that aim at suppressing tremors using active or passive dampening procedures [5]. Limb stiffness and recurring upper limb tremors are particular characteristics of Parkinson's disease that affects patients ability to perform activities of daily life (ADL). Bradykinesia is the third most common symptom, characterized by sluggish movement and problems in planning, initiating, and carrying out a movement. This inability to control and actuate limbs have a significant impact on the quality of life (QoL) of these patients. As these diseases advance, symptoms such as tremors, stiffness, and muscle weakness intensify impacting a patient's ability to perform basic tasks such as eating, drinking, and grooming. Manual dexterity of the phalanges also becomes more difficult as these diseases progress.

Additionally, there are today an estimated 12 million new cases of stroke are recorded each year across the globe. Existing research highlights that upper limb orthoses and exoskeletons tend to improve rehabilitation and physiotherapy training in stroke patients [6]. The preliminary literature review shows that the field research enhances improvements in ADL performance for people with neurodegenerative illnesses. Upper-limb orthoses can also lower the impact of muscular imbalance and limb stiffness.

B. Problem Statement

The presented work designs an actuated upper limb orthosis for Parkinson's and ALS patients with spinal muscular atrophy (SMA) and tremors. The primary objective of the design is to facilitate and smoothen upper limb movement, focusing particularly on activities of daily living (ADL) such as reaching. An earlier paper utilized a gravity balancing mechanism to assist in functional reaching for ALS patients. This new design removes the gravity balancing apparatus and substitutes the existing shoulder joint with a spherical joint. Additional emphasis is placed on designing a device that is comfortable and ergonomic for daily at-home use. By mounting it on a wheelchair, there is less stress and load on the patient while using the device, and allows for it to be operated in different situations.

II. Solution Approach and System Design

A design of an upper limb orthosis that allows full control and freedom of motion by the user needs to take several factors into consideration. The forward kinematics of the orthosis should closely resemble natural motion of the upper limbs: a spherical ball and socket joint at the shoulder, a rotational hinge about the elbow, and rotation of the forearm/hand about the elbow. Form-fitting to the user is another important factor to consider in the design. The system should be scalable and should comfortably fit to the user to encourage daily usage.

Our proposed design aims to take these considerations and develop a fully actuated, 5 DoF robotic upper-limb orthosis to be mounted to a user's wheelchair or other assistive chair device. By mounting to the user's chair, the added weight of the motors, electronics, power source and structure are entirely offset from the user's body. A rechargeable battery pack is included in the system to power the motors, sensors, and CPU. Figure 1 shows the proposed design attached to a human model and simple wheelchair.

An active spherical joint at the shoulder simulates and actuates the natural motion of the glenohumoral joint. The glenohumoral (ball-and-socket synovial) joint uses two rotations

Fig. 1: Orthosis System Design

for pointing the humerus and one rotation to roll about its longitudinal axis [7]. The first two rotational stepper motors control pointing of the humerus about the scapula. The third stepper motor, mounted above the shoulder, controls the longitudinal roll. Soft straps are used to secure the user's arm to the exoskeleton. The upper arm is strapped in two locations: just below the shoulder to the second stepper motor plane of motion, and close to the elbow, where it is fixed to the third stepper motor motion plane. This ensures the torque produced by the longitudinal roll is properly transferred to the user's limb.

The elbow hinge joint is controlled by the fourth stepper motor, located laterally to the elbow joint. The lower arm is fixed to the fourth motor motion plane through a soft strap just below the elbow. The last stepper motor is located adjacent to the user's wrist, and controls forearm rotation about the elbow. The fifth motor achieves longitudinal torque about the elbow using a special strap that acts as a timing belt wrapped around the user's palm. By combining the motion of the five stepper motors, a user can achieve natural arm motion with minimal muscle input.

Most common arm motion can be deconstructed into two simple motion studies: forward extension/retraction, and lateral abduction/adduction. Figure 2 shows the first motion study with the user reaching forward to grab an object. Motor 2 controls the shoulder flexion/extension and motor 4 controls the elbow flexion/extension.

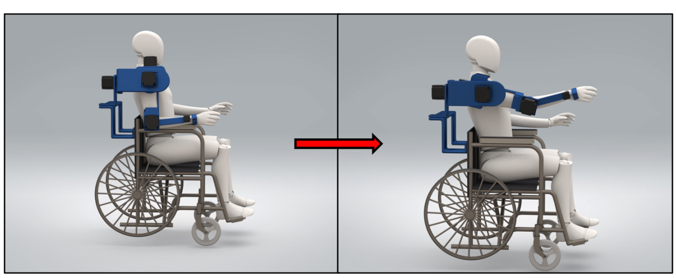


Fig. 2: Motion Study 1: Forward Extension and Retraction

The second motion study (shown in Figure 3) comprises a shoulder abduction and flexion with elbow rotation to bring the hand to the user's mouth.

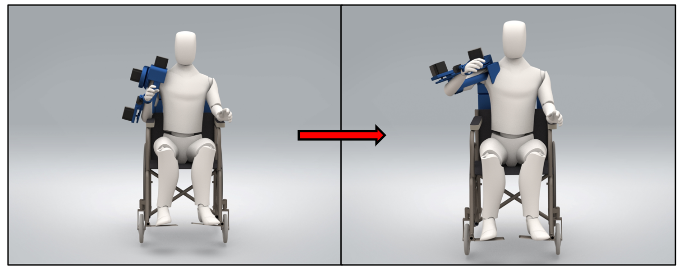


Fig. 3: Motion Study 2: Lateral Abduction and Flexion

In this motion, motors 1 and 2 control pointing of the humerus about the scapula, motor 3 controls the roll about the shoulder, and motor 4 rotates the elbow to guide the hand to the user's mouth. At the peak of this motion, motor 5 keeps the hand steady from any tremors or weakness to enable the user to take a drink from a glass or eat food.

The system design is based on an average adult's arm dimensions obtained from literature [8], but can be tuned per required specifications. These estimated dimensions and parameters are displayed in Table I and are used in the simulation of the orthosis.

TABLE I: Estimated Human Patient Parameters

Parameter	Upper Arm	Forearm
Mass (kg)	4.5	4
Length (m)	0.3	0.4
Width (m)	0.14	0.11
Thickness (m)	0.14	0.11
Height from Ground (m)	1.4	1.2

In order to limit complexity, fixed ranges of motion have been described for each joint, as shown in Table II below.

TABLE II: Orthosis Configuration and Movement Limits

Link	Muscle Action	Range
1	Shoulder Abduction	0 to $\pi/2$
2	Shoulder Extension	0 to $\pi/3$
3	Shoulder Medial Rotation	$-\pi/3$ to $\pi/3$
4	Elbow Extension	0 to $\pi/3$
5	Forearm Rotation	$-\pi/2$ to $\pi/2$

III. Results and Discussion

The simulation and data analysis for this system was conducted with a two-faceted approach.

A. *Mathematical Modeling of the Orthosis*

A Denavit-Hartenberg analysis was conducted on the preliminary arm design, and the coordinate frames were designed, as seen in Figure 4.

Using this set of coordinate frames, we were able to derive the Denavit-Hartenberg parameters [9], and by extension, the

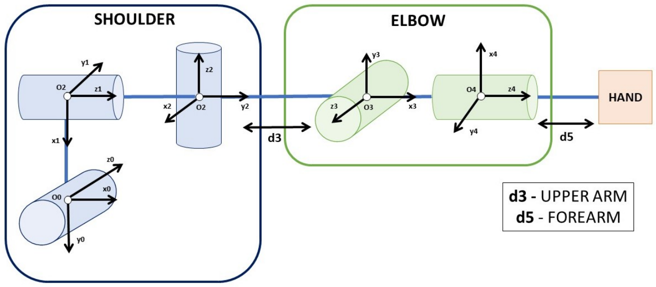


Fig. 4: Denavit-Hartenberg Coordinate System of the Arm Orthosis.

transformation matrices of the entire system. The selected parameters account for the rest position of the orthosis, where the upper arm and forearm are perpendicular to each other.

TABLE III: DH Table for the Arm Orthosis

Link	θ_i	d_i	a_i	α_i
1	$q_1 + \pi/2$	0	0	$\pi/2$
2	$q_2 + 3\pi/2$	0	0	$\pi/2$
3	q_3	d_3	0	$-\pi/2$
4	$q_4 + \pi/2$	0	0	$\pi/2$
5	q_5	d_5	0	0

These homogeneous transformation matrices provide a trigonometric correlation between the base frame (at the shoulder) and the end-effector frame (at the hand). From hereforth, $cos(q_i)$ and $sin(q_i)$ will be written as c_i and s_i respectively.

$$T_0^5 = \begin{bmatrix} t_{11} & t_{12} & t_{13} & t_{14} \\ t_{21} & t_{22} & t_{23} & t_{24} \\ t_{31} & t_{32} & t_{33} & t_{34} \\ 0 & 0 & 0 & 1 \end{bmatrix}$$

where,

$$t_{11} = c_1c_3s_5 - c_1c_5s_3s_4 + s_1s_2s_3s_5 - c_2c_4c_5s_1 + c_3c_5s_1s_2s_4$$
$$t_{12} = c_1c_3c_5 + c_2c_4s_1s_5 + c_5s_1s_2s_3 + c_1s_3s_4s_5 - c_3s_1s_2s_4s_5$$
$$t_{13} = c_1c_4s_3 + c_2s_1s_4 - c_3c_4s_1s_2$$
$$t_{14} = c_2d_3s_1 + c_2d_5s_1s_4 + c_1c_4d_5s_3 - c_3c_4d_5s_1s_2$$
$$t_{21} = c_3s_1s_5 - c_1s_2s_3s_5 - c_5s_1s_3s_4 + c_1c_2c_4c_5 - c_1c_3c_5s_2s_4$$
$$t_{22} = c_3c_5s_1 - c_1c_5s_2s_3 + s_1s_3s_4s_5 - c_1c_2c_4s_5 + c_1c_3s_2s_4s_5$$
$$t_{23} = c_4s_1s_3 - c_1c_2s_4 + c_1c_3c_4s_2$$
$$t_{24} = c_4d_5s_1s_3 - c_1c_2d_3 - c_1c_2d_5s_4 + c_1c_3c_4d_5s_2$$
$$t_{31} = c_4c_5s_2 + c_2s_3s_5 + c_2c_3c_5s_4$$
$$t_{32} = c_2c_5s_3 - c_4s_2s_5 - c_2c_3s_4s_5$$
$$t_{33} = -s_2s_4 - c_2c_3c_4$$
$$t_{34} = -d_3s_2 - d_5s_2s_4 - c_2c_3c_4d_5$$

Furthermore, these transformation matrices allow for the calculation of the Jacobian of the entire system. Through the Jacobian, the Euler-Lagrange equations of the system can be formulated.

$$D(q) = J^T \times M \times J$$

$$D(q)\ddot{q} + C(q,\dot{q})\dot{q} + g(q) = \tau$$

where J is the Jacobian matrix, M is a diagonal matrix of the mass of upper arm and forearm, q refers to the joint variables, $\dot{q}$ refers to the joint velocities, d_3 is the length of the upper arm, d_5 is the length of the forearm, $D(q)$ is a 5×5 inertia matrix, $C(q,\dot{q})$ is a 5×5×5 Coriolis tensor which is simplified into a 5×5 matrix by multiplying by the joint space velocity vector as can be seen above, and $g(q)$ is a 5×1 matrix of gravitational terms. $g(q)$ is calculated by taking partial derivatives of the total gravitational potential energy with respect to the joint variables. τ refers to the motor torques that will applied at each joint to assist the arm in reaching the desired end-effector position in a 3D space.

B. Simulation of the Orthosis

After the arm orthosis was modeled, a desired final end-effector position was chosen for the system. This position was selected based on the functional reach action of bringing the hand toward the patient's mouth. In the final implementation of this system, we aim to utilize EMG data from the arm to define the required end position. Additionally, we assume that the system begins and ends at rest position (all velocities are equal to zero).

TABLE IV: Desired Final Position for Joints

Muscle Action	Position
Shoulder Abduction (q_1)	$\pi/5$
Shoulder Extension (q_2)	$\pi/4$
Shoulder Medial Rotation (q_3)	$\pi/3$
Elbow Extension (q_4)	$\pi/6$
Forearm Rotation (q_5)	$\pi/8$

A computed torque control law was chosen due to its resilience against external disturbances [10], which are expected because of the random arm tremors for Parkinson's patients. This allows us to calculate the desired torques for the end-effector to reach the defined end-location.

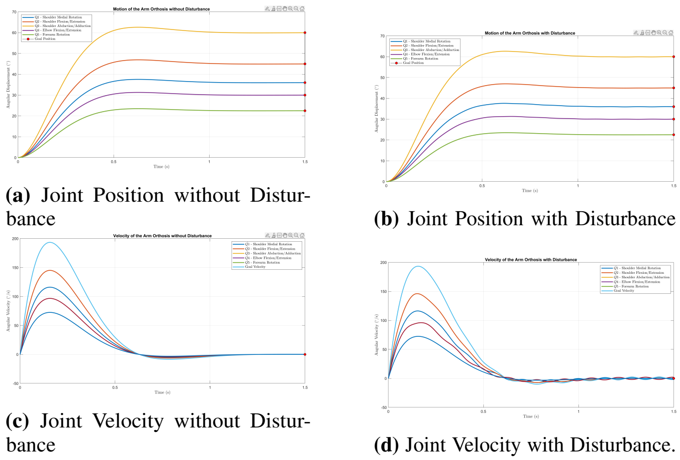
$$\tau_{des} = \tilde{D}(\ddot{q} + k_d\dot{e} + k_pe) + \tilde{C}(q,\dot{q})\dot{q} + \tilde{g}(q)$$

where k_p and k_d are the proportional and derivative control gains, and are taken to be equal to 50 and 10 respectively.

Additionally, literature has shown that the frequency of these tremors can be characterized into distinct categories such as postural and action tremors [11]. We assume for the ease of the simplification of analysis to model the tremor as a constant disturbance with a frequency of 7 Hz, and an amplitude of 2. We apply this disturbance at Link 4, acting on the elbow extension and flexion.

$$\tau_{disturbance} = 2\sin(14\pi t)$$

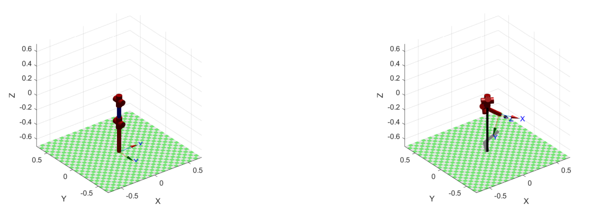
The equations of motion were solved using the ode45 solver on MATLAB for a duration of 1.5 seconds, as this is the expected duration of each action. The simulation was conducted for two scenarios - firstly without the disturbance torque to identify the general performance of the model, and secondly with the constant disturbance.



(a) Joint Position without Disturbance
(b) Joint Position with Disturbance
(c) Joint Velocity without Disturbance
(d) Joint Velocity with Disturbance.

Fig. 5: Simulation of the Orthosis

A preliminary animation of orthosis was also generated [12], showing the motion to the end-point from its initial resting position.



(a) Start Position at $t_i = 0$
(b) End Position at $t_f = 1.5s$

Fig. 6: Animation of the Orthosis

As can be seen from plots in Figure 5, the designed control model is robust and capable enough to reach its goal position with the constant applied disturbance. The red circles mark the desired end position, and are always reached within the time constraint. However, there are some concerns that we have faced with this simulation. Firstly, there is steady-state error noted in joint velocities, which poses a challenge for stability of the arm. This is caused due to an acknowledged simulation constraint - there is no defined frequency at which the tremors will occur in Parkinson's patients, and they are modeled to be consistently occurring. This error can be further reduced by implementing a PID controller [13], instead of the currently implemented PD controller. A second concern can be seen in both joint velocity plots. There is a marked overshoot at the initial command, whereby the joint velocity reaches a value in excess of $200°/s$. In a physical implementation, this will be reflected in a sharp, jerking motion which can be painful to the patient. The model can be further optimized and trained, such as through the implementation of controller gain schedulers [14], which can be used to modify controller gains depending on the operating conditions. Alternatively, we can maintain the PD controller architecture, as the response is faster than required. Thus, we have the ability to systematically reduce the gains to reduce the overshoot.

IV. Conclusions and Future Extensions

The proposed design is mathematically stable, and simulations prove the feasibility and reliability of the solution for patients suffering from neurological disorders. However, a set of iterations to manufacture the orthosis and test the prototype on an actual patient to understand the ease of use, wearability and ergonomics is essential to understand the feasibility of the solution. In the present work, our orthosis model with tremors as disturbance torque and a PD controller is used to control the tremors in Parkinson's patients. A further extension of the present work could be to add a spring-damper system to counteract the impact of tremors on patients. Some possible control strategies could be - using Model-based control with system identification to understand spring-damper system properties and incorporate these design parameters to develop a controller to reduce the amplitude of the tremors. Alternatively, we could use feedback control to measure the tremor amplitude and frequency in real time to vary the stiffness and damping coefficients of the spring-damper system to reduce the impact of tremors. This design iteration aims to stabilize the tremors and help the patient with ADL but does not have assistance for object grasping.

Another possible extension could be to build a grasping mechanism to help with the movement of fingers using sensors to detect muscle activity and a control strategy to help with actuation to complete the grasping. A machine learning-based algorithmic control strategy could be to further detect and understand more about the patient's movements and customise the solution, adapting the design parameters based on individual patient needs. Each of these extensions requires thorough research to understand and improve the efficacy. Initial simulation results are promising, but the present design needs to be re-iterated, thoroughly tested and updated based on user feedback.

References

[1] National Amyotrophic Lateral Sclerosis (ALS) Registry. https://www.cdc.gov/als/WhatisALS.html: :text=On

[2] M. Clinic, "Parkinson's disease," Mayo Clinic, 17 February 2023. [Online]. Available: https://www.mayoclinic.org/diseases-conditions/parkinsons-disease/symptoms-causes/syc-20376055. [Accessed 27 April 2023].

[3] T. PROIETT, E. AMBROSINI, A. PEDROCCH and S. MICERA, "Wearable Robotics for Impaired Upper-Limb Assistance and Rehabilitation: State of the Art and Future Perspectives," IEEE Access, vol. 4, p. 19, 2016.

[4] H. S. Nguyen and T. P. Luu, "Tremor-Suppression Orthoses for the Upper Limb: Current Developments and Future Challenges," Frontiers, vol. 15, no. 1, p. 16, 2021.

[5] N. P. Fromme, M. Camenzind, R. Riener and R. M. Rossi1, "Need for mechanically and ergonomically enhanced tremor-suppression orthoses for the upper limb: a systematic review," Journal of NeuroEngineering and Rehabilitation, vol. 16, no. 93, p. 15, 2019.

[6] "Design of a Gravity-Balanced Orthosis for Upper Limb Functional Reach for Early Stage ALS Patients" Omo, Evan, Sharma, Shruti, Rahman, Danish

[7] J. Lenarcic and M. Stanisic, "A humanoid shoulder complex and the humeral pointing kinematics", IEEE Transactions on Robotics and Automation, vol. 19, no. 3, pp. 499-506, 2003.

[8] G. Singh, A. Singla, and G. S. Virk, “Modeling and Simulation of a Passive Lower-Body Mechanism for Rehabilitation,” in Conference on Mechanical Engineering and Technology (COMET-2016), Department of Mechanical Engineering, IIT (BHU), Varanasi, Jan. 2016, pp. 123–128.

[9] Bertomeu-Motos, Arturo Lledo, Luis Diez, Jorge Catalán, José M. Ezquerro, Santiago Badesa, Francisco Garcia, Nicolas. (2015). Estimation of Human Arm Joints Using Two Wireless Sensors in Robotic Rehabilitation Tasks. Sensors. 15. 30571-30583. 10.3390/s151229818.

[10] H. J. Nagarsheth, P. V. Savsani and M. A. Patel, ”Modeling and dynamics of human arm,” 2008 IEEE International Conference on Automation Science and Engineering, Arlington, VA, USA, 2008, pp. 924-928, doi: 10.1109/COASE.2008.4626407.

[11] Charles PD, Esper GJ, Davis TL, Maciunas RJ, Robertson D. Classification of tremor and update on treatment. Am Fam Physician. 1999 Mar 15;59(6):1565-72. PMID: 10193597.

[12] P.I. Corke, “Robotics, Vision Control”, Springer 2017, ISBN 978-3-319-54413-7.

[13] Briat, Corentin (2015). Linear Parameter-Varying and Time-Delay Systems - Analysis, Observation, Filtering Control. Springer Verlag Heidelberg. ISBN 978-3-662-44049-0.

[14] Leith, D. J., Leithead, W. (2010). Survey of gain-scheduling analysis and design. International Journal of Control, 73, 2000(11), 1001–1025. https://doi.org/10.1080/002071700411304

[15] Durante, F.; Raparelli, T.; Zobel, P.B. Development of a 4-DoF Active Upper Limb Orthosis. Robotics 2022, 11, 122. https://doi.org/10.3390/robotics11060122

[16] Stabilizing the hands of Parkinson’s patients: A collaborative journey between STIL and ACE 26 October 2022 ACE-Press

[17] Ruonala V, Meigal A, Rissanen SM, Airaksinen O, Kankaanpää M, Karjalainen PA. EMG signal morphology and kinematic parameters in essential tremor and Parkinson’s disease patients. J Electromyogr Kinesiol. 2014;24(2):300–6.

[18] Yongkun Z, Haijun W, Mingquan Z, Juzheng M, Masahiro T. Design methodology of portable upper limb exoskeletons for people with strokes. Frontiers in Neuroscience 2023, 17. https://doi.org/10.3389/fnins.2023.1128332