

A Neuromotor Device for Reducing Phantom Limb Pain in Individuals with Spinal Cord Injury

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Abstract. Phantom Limb Pain is a disorder that can be experienced by individuals after amputation or spinal cord injury. In spinal cord injury the paralysis or paresis is often bilateral, thus limiting the application of apparent movement as a therapeutic model for phantom limb pain. This project aimed to develop a robotic rehabilitation device that replicated apparent movement to apply the same therapeutic principles with individuals with lower limb phantom pain that have bilateral paralysis or paresis. The proposed device achieved lower limb planar motion of the knee by a six-bar linkage of a single degree of freedom (DOF). It is driven by a linear actuator while the ankle motion is achieved by a gear motor, reaching an effective 70° range of motion for both joints. The system features closed loop control using feedback from surface electromyography sensors, limit switches and position sensors with an Arduino microcontroller as the control unit. This device will be used to further our understanding of the disorder and create opportunities for robot aided treatment for individuals with phantom limb pain as a result of spinal cord injury.

1 Introduction

Spinal cord injury (SCI) is damage to the spinal cord, which could lead to loss of mobility and sensory function. A high proportion of the people with SCI experience neuropathic pain from an area of their body that is paralysed due to central nervous system damage caused by the SCI, rating it as one of the most unmanageable problems [1].

Neuropathic pain occurs in the insensate region below a complete SCI is usually predominated by central nervous system mechanisms in a way similar to phantom limb pain (PLP) following amputations [2]. The person perceives pain from parts of their body that are paralysed in the absence of actual tissue injury, which is not well managed with medications. It has been hypothesized that the illusion of limb movement in individuals with SCI (Moseley et al.) as they think of moving their limb may reduce the symptoms of neuropathic pain, as the mirror technique alleviated the symptoms of PLP in amputees [3]. It is likely that any successful therapeutic device relies on the coordination of the imaginary movement (specific motor cortex activation) and the concordant visual feedback of the movement. Such coordination is critical as it has been shown that imaginary movement without visual feedback of the limb moving increases temporary dyesthesia or replicates localised pain memory for PLP [4, 5].

Since SCI often induces paralysis or paresis of both lower limbs, the neuromotor device needs to be able to

move the knee and ankle of the patient in accordance to their thoughts of moving the limb. This effectively creates the illusion that the patient is moving their own leg. This concordant afferent and efferent cortical activation forms the basis of the proven mirror-box rehabilitation methods for individuals with amputations [3].

Lower-limb exoskeletons have been used for various purposes. Ferris et al. [6-9] built a pneumatically powered knee-ankle-foot orthosis, which was used to test this device's mechanical performance during human walking. Chu et al. [10] proposed a hydraulically driven lower-limb exoskeleton driving the hip, knee, and ankle joints to compensate for the strength and endurance of a human under a payload. Luginis et al. [11] presented an active stance-control knee-ankle-foot orthosis. Cui et al. [12] proposed a one active degree-of-freedom closed-loop compliant exoskeleton to enable humans to carry loads during long distance steady locomotion. Dollar and Herr [13] reviewed the state-of-the-art of lower limb exoskeletons in terms of design, actuation, sensory, and control.

However, the aforementioned devices are not able to fulfil this specific task as they either neglect to rotate the ankle joint or do not have the ability to incorporate muscle feedback into the locomotion process. In this paper, we presented a robotic rehabilitation device capable of applying therapeutic principles similar to mirror motion with individuals that have spinal cord

injury and experience phantom limb pain by simulating limb movement.

This paper is organized as follows. Section 2 introduces the mechanical design of the neuromotor device. Section 3 presents kinematic and stress analyses. Section 4 discusses the system design and implementation of the device. Section 5 concludes the paper.

2 Overview of mechanical design

2.1 Design considerations

The design considerations defined by the scope of the project are listed in Table 1.

Table 1. Design considerations.

Customer Requirements	Technical Parameters
Able to accommodate patients of different heights	Ranging from 1.6 m to 1.9 m
Able to withstand the weight of the leg	Force and stress safety under the nominal load of 15 kg
Able to stop at any instant	Emergency stop implementation
Enhanced safety	Limit switches and software safety implementation
Ease of setup and use	Attached to a rehabilitation chair

2.2 Mechanical design

We proposed to use a hybrid parallel-serial linkage mechanism to actuate the knee and ankle joints. The knee joint is actuated by a six-bar linkage with a linear actuator drives the knee joint and a servo-motor driven drives the ankle joint, as in Fig. 1.

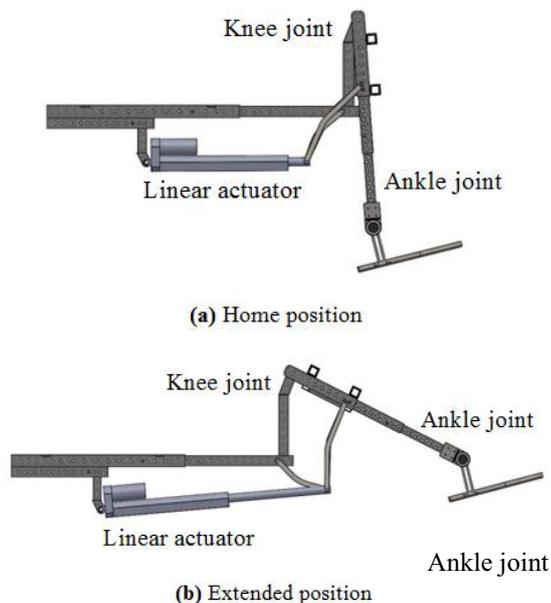


Figure 1. A hybrid parallel-serial knee-ankle orthosis in home and extended positions

The design features standard sized square hollow section tubing to construct the majority of components

supporting the limb. The linear actuator provides an effective range of motion of 70 degrees at the knee joint.

3 Kinematic and stress analyses

3.1 Kinematic Analysis of the knee mechanism

The six-bar linkage consists of a ternary link and five binary links connected by five revolute joints and one prismatic joint, which is a variation of Stephenson type six-bar linkage [14], as in Fig. 2.

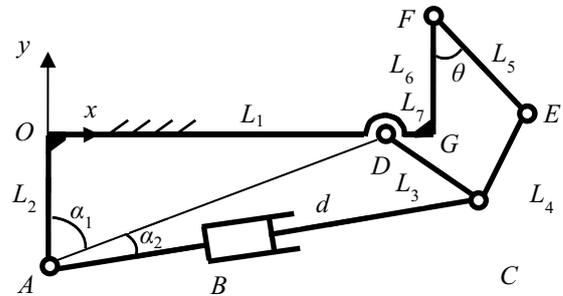


Figure 2. Schematic drawing of the six-bar knee mechanism

The structural parameters are: $OD = L_1$, $OA = L_2$, $CD = L_3$, $CE = L_4$, $EF = L_5$, $FG = L_6$, $GD = L_7$. The input is d , which is the length of AC , and the output is knee angle θ , which is the angle between GF and FE , as in Fig. 2.

The angle α_2 between DA and AC can be obtained as

$$\alpha_2 = \text{acos} \left(\frac{d^2 + L_1^2 + L_2^2 - L_3^2}{2d\sqrt{L_1^2 + L_2^2}} \right) \quad (1)$$

The angle $\alpha_1 = \text{atan}(L_1/L_2)$ is a structural parameter. Hence the coordinates of the joint C are

$$\begin{aligned} C_x &= d \sin(\alpha_1 + \alpha_2) \\ C_y &= -L_2 + d \cos(\alpha_1 + \alpha_2) \end{aligned} \quad (2)$$

The link lengths L_4 and L_5 constraints transform the problem of finding the knee angle θ into circle-circle intersection, where the first circle is centred at the joint C with a radius of L_4 and the second circle is centred at the joint F with a radius of L_5 , as in Fig. 3.

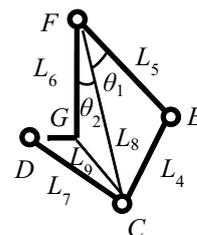


Figure 3. The knee angle θ

It follows that the knee joint θ is

$$\theta = \theta_1 + \theta_2 = \text{acos} \left(\frac{L_8^2 + L_5^2 - L_4^2}{2L_5L_8} \right) + \text{acos} \left(\frac{L_8^2 + L_6^2 - L_9^2}{2L_6L_8} \right) \quad (3)$$

where

$$L_8 = \|CF\| = \sqrt{(C_x - L_1 - L_7)^2 + (C_y - L_6)^2}$$

$$L_9 = \|CG\| = \sqrt{(C_x - L_1 - L_7)^2 + C_y^2}$$

3.2 Stress analysis

We conducted stress analysis in SolidWorks on key parts of the device subject to 15 kg load, considering the average weight of the calf and foot of a 150 kg adult is around 150 kg, as in Fig. 4. The weakest component has a safety factor of 1.85.

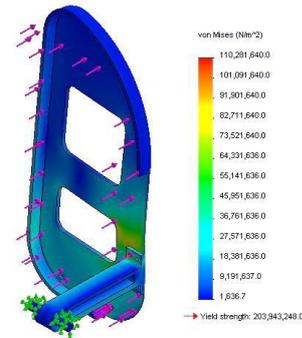
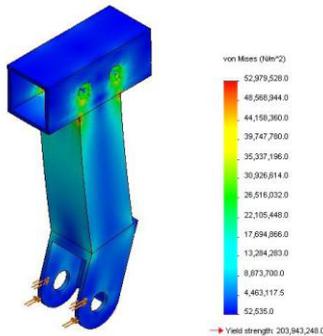


Figure 4. Stress analysis of the base link and the footpad

4 System design and implementation

Safe operation of the system was achieved through the implementation of safety elements such as position sensors, limit switches, emergency hand switch and physical motion stoppers. Also, a graphical user interface was designed to facilitate the operation of the system and to provide supervisory control to the therapist. An overview of the system architecture block diagram is shown in Fig. 5.

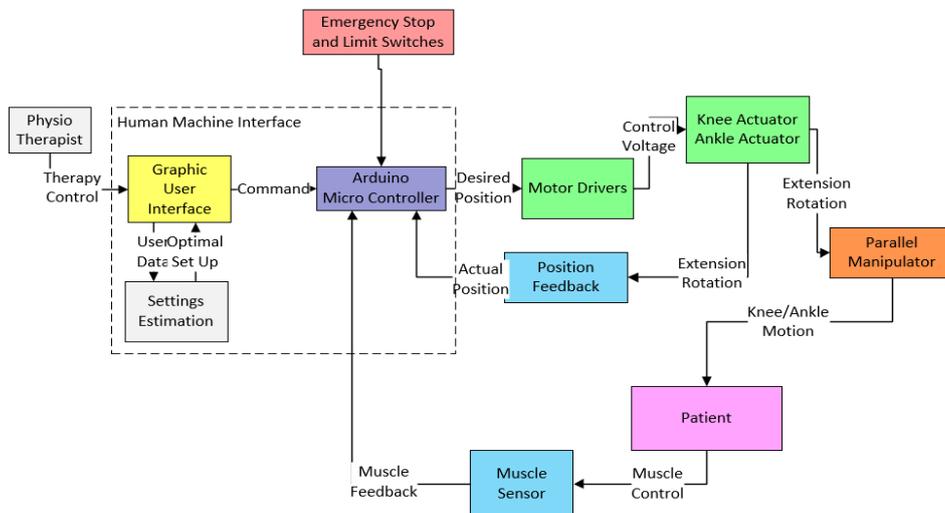


Figure 5. System architecture block diagram

A prototype was built to validate the proposed design, as in Fig. 6.



Figure 6. A prototype of the proposed design

We proposed to use diagonal activations where the controller is the EMG signal of the biceps on the upper limb. The individual was instructed to imagine a rope between the hand and ankle and that they act together resulting in the biceps EMG signal initiating the movement, as in Fig. 7.

We tested the device on one healthy subject. Surface EMG electrodes to trigger the muscle feedback function of the system were placed on biceps and the subject bent his forearm to initiate one cycle of rehabilitation training.

The results showed that the motion and speed performance was satisfactory. The lower limb was elevated at a comfortable pace, which was natural and unforced. The joint angles and sEMG sensor signals are shown in Fig. 8.

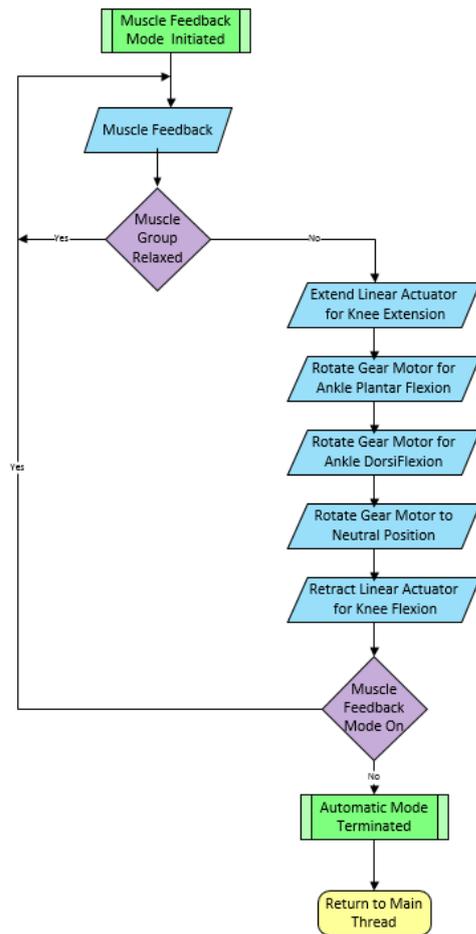


Figure 7. Control flowchart

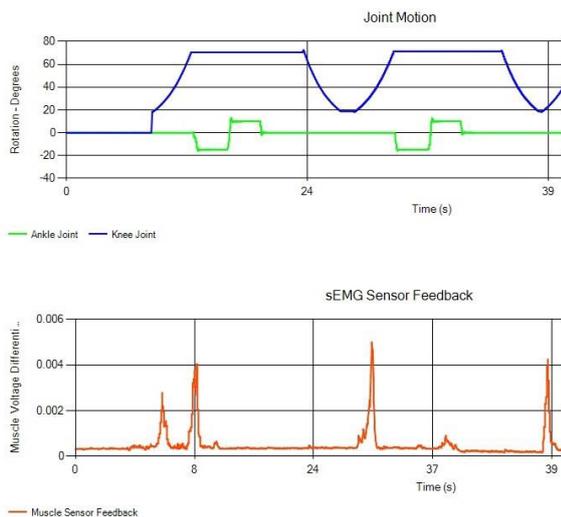


Figure 8. Knee and ankle joint displacements and the corresponding sEMG signals on the biceps

As can be seen on the joint motion plot, the actuation sequence begins with the ankle actuator, upon reaching the desired position; ankle actuation begins and is followed by knee flexion. Muscle feedback above 2 mV threshold trigger actuation sequence, as seen in both cycles, during this time rehabilitation device only records muscle feedback and signals higher than 2 mV are neglected until cycle is terminated.

5 Conclusions

We presented a lower limb rehabilitation system, which introduced planar limb motion to the paralysed limb of patients through a single degree of freedom six-bar linkage for the knee joint and a geared motor for the ankle joint in conjunction with associated control and safety elements. Further development to increase patient comfort in terms of limb support and thorough safety testing would complete the system for patient trials.

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