Development and Control of a Lower Extremity Assistive Device (LEAD) for Gait Rehabilitation

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***Abstract*— This research developed a wearable lower extremity assistive device intended to aid stroke patient during rehabilitation. The device specifically aims to assist the patient in sit-to-stand, stand-to-sit, and level-walking tasks in order to promote active gait rehabilitation exercises. The device adopts an anthropomorphic structure with hip and knee joint actuated in sagittal plane. A finite state machine strategy was proposed to control the device. At different states, appropriate assist torque is added to each joint. EMG signals are used to assess the assist performance. Tests on an able-bodied subject show that the device could successfully detect and transit between states. In sit-to-stand tasks, the integrated EMG (iEMG) of the Vastus Medialis for standing up with 11 Nm of assistance torque were found to be significantly less (P = 0.00187) than the iEMG of without assistance for standing up which indicate reduced muscle effort with the device assistance. Results show the device could potentially assist stroke patient in similar tasks.**

Keywords—Assistive device, lower extremities rehabilitation, finite state machine (FSM), intention detection.

I. INTRODUCTION

Limitations in walking are an important cause of disability and morbidity after stroke, affecting nearly two thirds of stroke survivors [1]. It correlates with inpatient length of stay for both acute treatment and chronic rehabilitation [2], and also increases the burden of care and correlates with the rate of readmission to hospital and long-term institutionalization [3, 4]. High intensity, task-oriented rehabilitation have shown to enhance walking distance and walking speed of patient after stroke, particularly for those with moderate walking deficits [5, 6]. However, conventional gait rehabilitation is a highly repetitive, labor-intensive task. Such training is typically limited to 20-30 minutes each session due to therapist and/or patient fatigue. Repeatability of these training sessions is also poor, as assistance level differs between therapists.

Robots for gait rehabilitation are seen as a solution to this issue, and in the recent years, there has been a significant increase in the research and development especially in the area of wearable lower extremity robots. These devices are designed to be worn by the user and provide force to move the user's leg. They are commonly designed for automated gait training on a treadmill, like the AutoAmbulator, GaitTrainer, Lokomat [7], LOPES [8] and ALEX[9]. However, they are not widely available because they are expensive, large and not portable. It would be advantageous to have a portable rehabilitation system such that it can be taken home to assist with gait training at home. Current portable devices are not designed for rehabilitation, such as the HAL [10], which is designed to aid elderly people, or BLEEX [11], which is designed to amplify human strength.

In the control aspect, most of the devices move the patient through a fixed trajectory. Hence, there is no cycle-to-cycle variation in the kinematics and sensorimotor feedback. This may cause habituation to sensory input, which reduce sensory responses to weight-bearing locomotion and ultimately impair motor learning [12]. Surface electromyography (sEMG) signals are used to drive the HAL. However, the noisy nature of the EMG signals makes this method of power assistance uncomfortable to the user [13]. Force amplification method by the BLEEX could not be used for the rehabilitation since the device appears to be passive to the user, as the device shadows the movement of the user [14].

In this paper, we propose a design of a light-weight and portable lower extremity assistive device, LEAD (Fig. 1), and a finite state machine strategy to control the device with appropriate assist torque being added to each joint at different states to assist the user. This device aims to assist the patient in sit-to-stand, stand-to-sit, and level-walking tasks in order to promote active gait rehabilitation exercises at home.

Figure 1 Lower Extremity Assistive Device (LEAD)

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II. Hardware Design

LEAD was designed to aid in the flexion and extension motion of the user's hip and knee joints. To fit a wide range of users, the device consists of an adjustable anthropomorphic frame based on the anthropometrical data provided in [15], with actuator modules attached at the joints. Orthotic cuffs are used as the attachment interface between the device frame and the user. Each actuator module could deliver up to 35 Nm of torque and is powered by a DC motor attached to a harmonic drive at a 50:1 gear ratio. Optical incremental encoder at 1000 counts/rev is also equipped at the pre-reduction stage of each actuator module to measure the hip angle (θ_h) and knee angle (θ_k). To ensure the safety of the device, the range of motion of each joint is mechanically limited to be slightly smaller than the normal range of motion of a normal human.

The device incorporates several types of sensors to detect the current states of the device and the wearer. Besides the optical incremental encoders at each actuator module, surface-EMG sensors are attached on the wearer's flexor and extensor muscles of the knee to detect their activity level. Two force sensors are attached to the insoles of the wearer's shoe at the first metatarsal and calcaneus position to detect the ground reaction force (GRF) at the front and back of the foot respectively.

A reconfigurable embedded control and acquisition system which consists of a real-time processor, reconfigurable field-programmable gate array (FPGA), and analog and digital I/O is used for the control the device. Each actuator module is controlled by a digital servo drive. All the electronics can be fitted into a small backpack for portability. CAN communication at 1Mbits/s is implemented between the controllers to allow for scalability of the device.

III. Control Method

A. Finite State Machine (FSM) Controller

To effectively assist the user, the control algorithm of the device must ensure the hip and knee joints are driven in accordance to the user's intended motion. For this device, a FSM controller was implemented based on the kinematics and the GRF readings. Each state is divided based on their intended motion so torque added at each state could properly provide assistance.

The tasks required for the device are to aid in sit-to-stand, stand-to-sit, and level-walking. Fig. 2 shows the states transition diagram and Table I lists the state transition condition for the tasks. The thresholds for state transition are determined by an off-line analysis of the data collected from a healthy 28-year-old subject performing these tasks.

After the device is attached, the joint angles are initialized to zero while the user is in standing position. Note that positive hip and knee angles correspond to flexion, while negative values refer to joint in extension. During Initializing state, the controller will decide which initial position the user is in, sit or stand. If the GRF is small and knee angle is large, the device detects the user to be in the sit state. No assistance torque is required for both hip and knee joints in the Sit state. Therefore, both joints are set to a back-drivable mode to behave like free joints. When the user intends to stand up, the

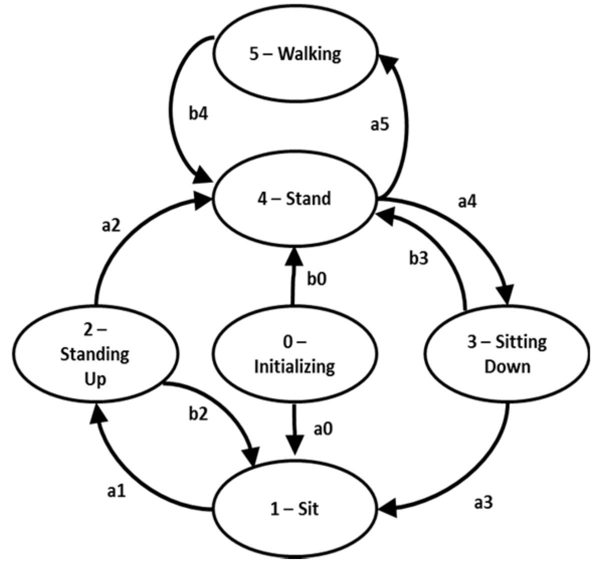

Figure 2 State transition diagram with their corresponding state number

TABLE I State Transition Condition

State Transition	Condition
a0	GRF < Threshold AND θk > Threshold
b0	Not a0
a1	GRF > Threshold
a2	θk < Threshold
b2	GRF < Threshold
a3	GRF < Threshold
b3	θk < Threshold
a4	θk > Threshold AND GRF > Threshold
b4	GRF < Threshold OR Front GRF > Threshold
a5	Back GRF Peak Time OR Time Out

GRF starts to increase as the user begins to loads his legs with his own weight. Assistance is necessary here especially in the knee extension muscles as it is required to lift the body up.

By detecting the GRF, the controller is able to switch to Standing Up state even before the actual movement of the knee joint, hence effective assistance could be achieved. During this state, both the hip and the knee joints are set to provide extension torque. Once the knee is relatively straight, the controller moves to the Stand state. Both joints are set to be free in this state. If knee muscle of the user is weak, a small extension torque could be provided to keep his knees extended. From Stand state, the device can transit into Sitting Down or Walking state. When the user intends to sit down, his knee angle will start flexing while the GRF maintains at load bearing level. This shifts the controller to the Sitting Down state, whereby the knee joint is set to provide extension torque. This is to assist user in his controlled descent to sit down on his chair. The unloading of legs detected by the decreasing GRF indicates the end of the Sitting Down state and the state shifts to the Sit state. More details of the Walking state will be mentioned in the next section. Some

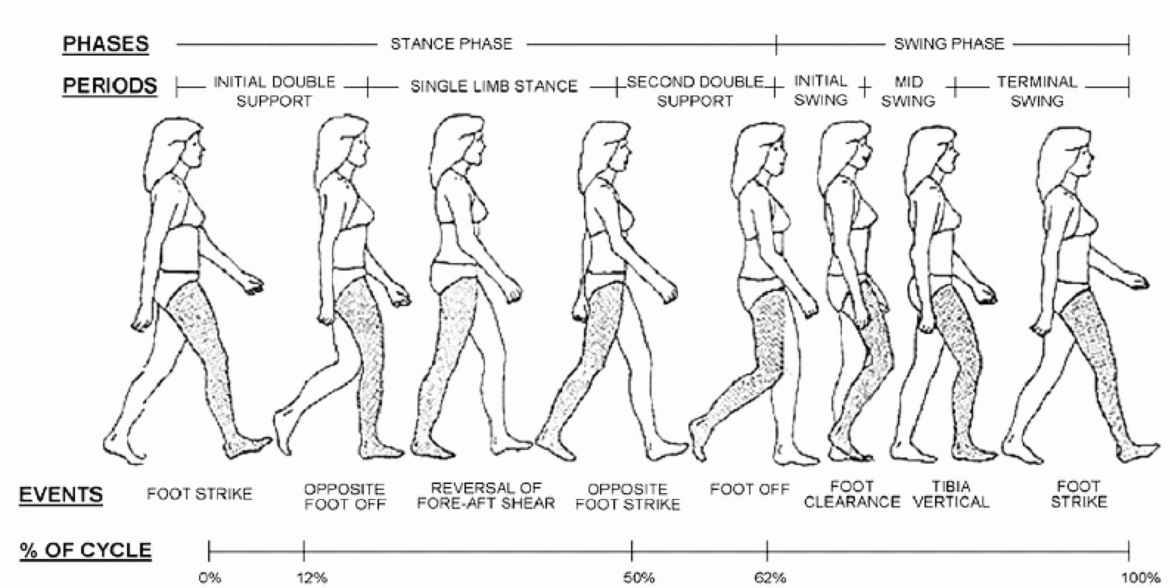

Figure 3 Normal walk cycle illustrating the events of gait [16]

state transitions are included to avoid being trapped indefinitely within a state.

B. Subdivision of Walking State

Walking is a more complicated task as compared to sit-to-stand and stand-to-sit. It involves a rhythmic coordination of body movements that are repeated over and over, step after step. Moreover, gait pattern and trajectory differs between different individuals or even within the same individual under different conditions, for example at different walking speed. Fortunately, much research has been done in the area of human locomotion. It is widely accepted that a basic human gait cycle could be generalized into a periodic sequence of events as shown in Fig.3 [16]. And the human gait cycle could be split into two major phases, the stance phase and the swing phase. The former is when the foot is in contact with the ground and the latter is when the foot is in the air. Moreover, each phase could be subdivided into periods, with each period playing an important function in the gait cycle (Table II).

TABLE II GAIT PERIODS AND FUNCTIONS

Period	Function
Initial Double Support	Loading, weight transfer
Single Limb Support	Support of entire body weight, center of mass moving forward
Second Double Support	Unloading and preparing for swing
Initial Swing	Foot clearance
Mid Swing	Limb advances in front of body
Terminal Swing	Limb deceleration, preparation for weight transfer

In view of the different functional requirements at different stages of the gait cycle, the walking state in our FSM controller is divided into sub-states based on the periods of the gait cycle. If the period of the gait cycle could be determined, the intended function of the joints would be known. Therefore, the device could add torque in the appropriate direction to aid the user in his movement as shown in Table III.

To collect the joint angle and GRF data, the subject was asked to walk while wearing the assistive device. In order to minimize the influence of the device, all joints are set to back-drivable mode. Fig. 4 shows the data of a subject from

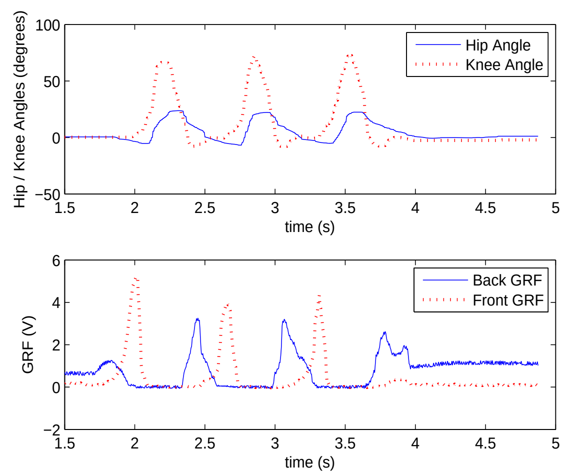

Figure 4 Data collected from left leg of subject during starting walk from standing position and ending walk back to standing position movement

standing position to walking, and from walking back to standing position after taking three strides with his left leg.

Fig. 5 illustrates the transition diagram between the sub-states inside the Walking state, and Table IV shows the state transition condition between the sub-states. An off-line analysis was done with the recorded walking data of the subject to calibrate the threshold for transition between the sub-states. Sub-state transitions are derived based on whether the mechanical goals of each sub-state are achieved, such as foot clearance. Therefore, two additional variables are required, namely, foot forward position (x), and shank angle (θ_s).

TABLE III DIRECTION OF ASSISTIVE TORQUE OF JOINTS FOR EACH SUB-STATE

No.	Sub-state	Direction of Assistive Torque at Joint	
		Hip	***Knee***
1	Initial Swing	Flexion	Flexion
2	Mid Swing	Free	Free
3	Terminal Swing	Extension	Extension
4	Initial Stance	Extension	Extension
5	Mid Stance	Free	Free
6	Terminal Stance	Flexion	Free

Foot forward position, x, is define as the distance at which the foot is in front of the body in the sagittal plane. Throughout this paper, the human body is assumed to be in upright position with respect to gravity during walking. Therefore, to determine foot position, the human leg is modeled as a 2-dof planar robot structure. Using forward kinematics,

$$x = L_1 \sin(\theta_h) + L_2 \sin(\theta_h - \theta_k) \tag{1}$$

where L_1 and L_2 refers to the length of the subject's thigh and shank respectively. And shank angle is defined as the relative angle between the shank of the subject to the line of gravity,

$$\theta_s = \theta_h - \theta_k \tag{2}$$

To detect the start of walking, the subject begins in Stand state. The device transits into Walking state by detecting whether the leg of interest is in stance or swing. Fig. 4 depicts an example of signal when the leg starts off in stance. A characteristic peak in the front GRF can be observed as the subject intends to pushes his body forward. And the FSM controller transit to the Late Stance sub-state. If the leg starts off in swing, the subject will unload his leg, which causes the FSM controller to transit to the Initial Swing sub-state.

To detect the end of walking, a characteristic delay in rise time of the back GRF is observed as the subject ends his step. If the back GRF takes longer then a pre-allocated amount of time to reach a certain value, the FSM controller transit from the Walking state to the Stand state.

IV. Experiments and Results

A. State transition

To test the FSM controller, a healthy 28 year old subject, wearing the device set to back-drive mode, was asked to perform a consecutive series of task. Firstly, the subject stands up from a sitting position on a chair. At standing position, he is asked to walk three strides with his left leg before stopping at standing position. Lastly, the subject is asked to sit back down on the chair. The result of the FSM controller in detecting different tasks performed by subject is shown in Fig. 6. It is observed that the FSM controller can successfully detect and transit between various states while the subject performs the series of task. Furthermore, the controller was found to work robustly in most of the trials performed by the subject.

B. Sit-to-stand Assistance

To ascertain the effectiveness of the device's assistance, muscle effort of the subject with and without assistance was compared. A widely used method to determine muscular effort is by measuring the iEMG signal of the target muscle over the period of the task [17]. Its formula is given as follows

$$iEMG = \int_{t_{start}}^{t_{end}} \|EMG(t)\|\, dt. \tag{3}$$

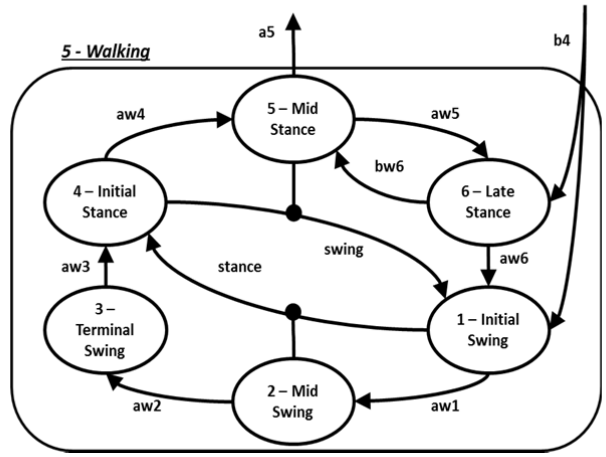

Figure 5 Sub-state transition diagram within Walking State

TABLE IV Sub-State Transition Condition

Sub-State Transition	Condition
aw1	Foot forward position, $x > 0$
aw2	Shank Angle, $\theta s > 0$
aw3	GRF > Threshold
aw4	Back GRF Peaked flag = True OR TimeOut
aw5	Front GRF > Threshold
aw6	GRF < Threshold
bw6	TimeOut
swing	GRF < Threshold
stance	GRF > Threshold

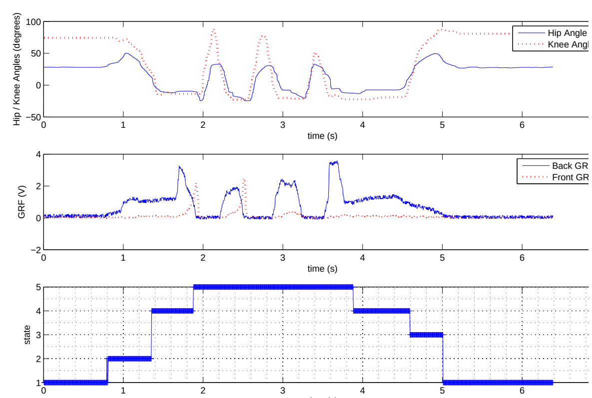

Figure 6 Sensor data of a subject preforming sit-to-stand, level-walking and stand-to-sit tasks in sequence and result of FSM controller. States are indexed as follows, Initializing = 0, Sit = 1, Standing up = 2, Sitting down = 3, Stand = 4, and Walking = 5.

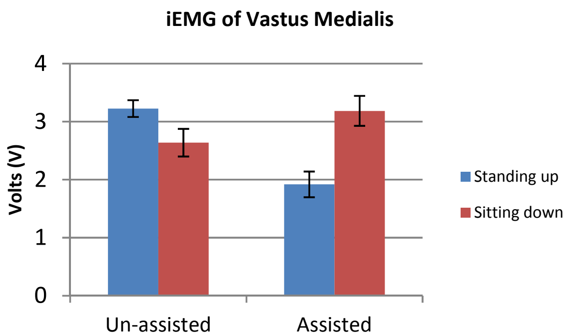

Figure 7 iEMG of Vastus Medialis Muscle during standing up and sitting down task

where EMG(t) is the measured EMG signal, and t_{start} and t_{end} refers to the start and end time of the task, respectively.

The subject was asked to perform 5 cycles of sit and stand at a normal pace with and without the device assistance. The knee extension torque for assisted standing up and sitting down are set to 11 Nm and 4 Nm, respectively. Assistive torque is adjusted based on the comfort level of the subject. Assisted torque for sitting down is consistently chosen to be less than that for standing up for a comfortable sitting down motion. Surface EMG electrodes are attached to the Vastus Medialis muscle which is one of the major muscles involved in knee extension.

Fig. 7 shows the iEMG results for standing up and sitting down task. A one-tailed t-test shows that the iEMG with device assistance is found to be significantly less ($P = 0.00187$) than the iEMG without assistance. The iEMG results for sitting down task shows no statistical difference between the two groups.

C. Sub-state Transition within Walking State

To see the effectiveness of sub-state transition and assistance during the Walking State, the subject performed a series of walking trials. For each walking trial, the subject is required to start walking from a standing position. The trial ends after the subject return to standing position after walking 2.4 m. The assistive torque for both flexion and extension motions of the hip and knee joints are set at a constant value of 7.5 Nm. From Fig. 8, it can be seen that the FSM controller can effectively detect and transit within the sub-states during the level-walking task.

V. Discussion

From the experimental trials, we can see that the device could successfully detect the user's intended motion and transit to the states. In addition, the significant reduction in iEMG signal of the knee extensor muscle during assisted standing up motion indicates that the device can provide effective assistance to the user for such task. Also, the device could detect and cycle through the sub-states within the Walking state. During walking, the subject could feel effective assistance from the device especially in the following cases, namely knee flexion during Initial Swing that helps to clear the foot off the ground, knee extension during late swing that helps to straighten the leg for landing and hip extension or flexion in their respective sub-states which helps to propel the body or leg forward during level forward walking. The risk of tripping during swing phase is minimal during level walking since assistive action takes the mechanical goal of foot clearance into account.

However, this assistive device is not without its limitations. Firstly, the user may not always be in an upright posture during walking. Therefore, in the near future, we intend to include a tilt sensor at the torso to measure the amount of body tilt. This would allow a more accurate

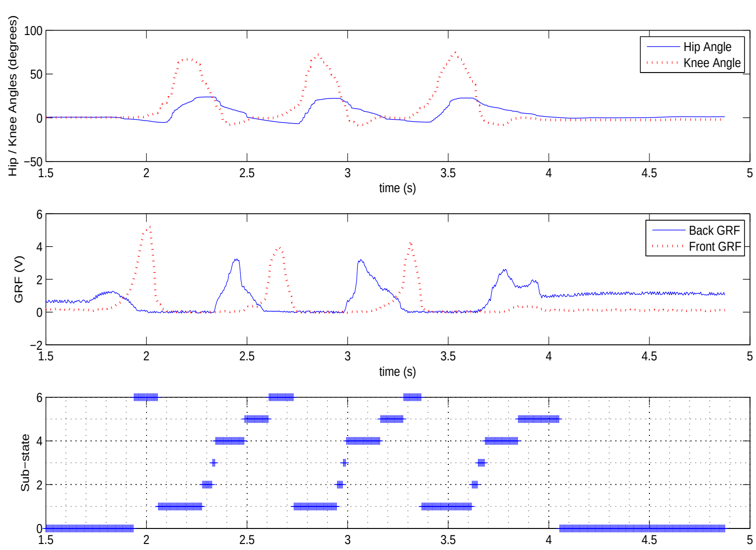

Figure 8 FSM controller result for walking motion starting from stand and ending to stand. Sub-states are indexed as follows, Initial Swing = 1, Mid Swing = 2, Terminal Swing = 3, Initial Stance = 4, Mid Stance = 5 and Terminal Stance = 6. A sub-state index of 0 refers to when the current state is not in the Walking State.

calculation of the forward kinematics of the leg. Secondly, the type of motion the device could detect is limited. Development to recognize and transit in to other motions such as climbing and descending of steps and slopes, and backward motion are currently being investigated.

VI. Conclusion

This research shows the development of a wearable lower extremity assistive device, LEAD, for gait rehabilitation. The development includes the design of the device's mechanical and electronic hardware. A FSM method of control was proposed to detect the user's intended motion and to provide appropriate assistance torque. The FSM controller was implemented on the device and experiment results show the effectiveness of the proposed method. From the results, it could be seen that the device can potentially be used to assist stroke patient in similar tasks. The usefulness of this method to detect and assist in other types of motion will be explored in future works.

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