Gravity-Balancing Leg Orthosis and Its Performance Evaluation

Sai K. Banala, Sunil K. Agrawal, Abbas Fattah, Vijaya Krishnamoorthy, Wei-Li Hsu, John Scholz, and Katherine Rudolph

***Abstract*—In this paper, we propose a device to assist persons with hemiparesis to walk by reducing or eliminating the effects of gravity. The design of the device includes the following features: 1) it is passive, i.e., it does not include motors or actuators, but is only composed of links and springs; 2) it is safe and has a simple patient–machine interface to accommodate variability in geometry and inertia of the subjects. A number of methods have been proposed in the literature to gravity-balance a machine. Here, we use a hybrid method to achieve gravity balancing of a human leg over its range of motion. In the hybrid method, a mechanism is used to first locate the center of mass of the human limb and the orthosis. Springs are then added so that the system is gravity-balanced in every configuration. For a quantitative evaluation of the performance of the device, electromyographic (EMG) data of the key muscles, involved in the motion of the leg, were collected and analyzed. Further experiments involving leg-raising and walking tasks were performed, where data from encoders and force-torque sensors were used to compute joint torques. These experiments were performed on five healthy subjects and a stroke patient. The results showed that the EMG activity from the rectus femoris and hamstring muscles with the device was reduced by 75%, during static hip and knee flexion, respectively. For leg-raising tasks, the average torque for static positioning was reduced by 66.8% at the hip joint and 47.3% at the knee joint; however, if we include the transient portion of the leg-raising task, the average torque at the hip was reduced by 61.3%, and at the knee was increased by 2.7% at the knee joints. In the walking experiment, there was a positive impact on the range of movement at the hip and knee joints, especially for the stroke patient: the range of movement increased by 45% at the hip joint and by 85% at the knee joint. We believe that this orthosis can be potentially used to design rehabilitation protocols for patients with stroke.**

***Index Terms*—Gait rehabilitation, gravity balancing, inverse dynamics, passive orthosis, rehabilitation robotics.**

I. Introduction

A VAST number of people are affected by conditions that result in profound muscle weakness or impaired motor control. People with severe muscle weakness from neurological injury, such as hemiparesis from stroke, often have substantial movement limitations. One of the aims of rehabilitation after stroke is to improve the walking function. However, equipment available to facilitate this is severely limited.

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Several lower-extremity rehabilitation machines have been developed recently to help retrain gait during walking. These machines are still not common in rehabilitation clinics. Lokomat is an actively powered exoskeleton, designed for patients with spinal cord injury. The patients use this machine while walking on a treadmill [1]. The Mechanized Gait Trainer (MGT) is a single degree-of-freedom (DOF) powered machine that drives the leg to move in a prescribed gait pattern. The machine consists of a foot plate connected to a crank and rocker system. The device simulates the phases of gait, supports the subjects according to their abilities, and controls the center of mass (COM) in the vertical and horizontal directions [2]. AutoAmbulator is a rehabilitation machine to assist individuals, with stroke and spinal cord injuries, in leg motion impairments. This machine is designed to replicate the pattern of normal gait [3]. HAL [4] is a powered suit for elderly people and persons with gait disorders, which takes electromyographic (EMG) signals as input and produces appropriate torque to perform the task. A similar power-assisted exoskeleton device is Berkeley's lower-extremety exoskeleton [5], though it is not intended as a rehabilitation device, but more as a human strength multiplier. The Pelvic Assist Manipulator (PAM) is an active device for assisting the human pelvis to allow naturalistic motion [6]. There are a variety of active devices that target a particular disability or weakness, and by repetitive motion, try to regain the lost capability [7]–[11]. A limiting feature of these machines is that they move patients through predetermined movement patterns rather than allowing them to move under their own control. The failure to allow patients to experience and practice appropriate movement patterns prevents necessary changes in the nervous system to promote relearning of typical patterns. Wilmington Robotic Exoskeleton (Training WREX or T-WREX) is a passive gravity-balanced device for retraining upper arm movement [12]. They have shown that gravity-balancing appears to improve motor learning capability. However, to apply a wide variety of forces, Pneu-WREX was developed, which is a pneumatically actuated version of T-WREX [13].

Related work using this gravity-balancing device has focused on the effects of rehabilitation machines on postural stability during walking, and the nature of internal limb forces and moments as different kinds of rehabilitation machines are added to the leg [14]. However, the work described in this paper focuses on the design and performance evaluation of the specific gravity-balancing machine that we have fabricated and tested.

In this paper, the performance of our gravity-balancing orthosis is studied by measuring the EMG signals of the leg muscles involved and also by measuring force-torque data, which are described in detail later in this paper.

Our leg orthosis is designed to assist persons with hemiparesis to walk through elimination of the effects of gravity.

The proposed device is designed to be passive. The main reason why we chose a passive device is because of safety. Usually, patients are more comfortable and feel safer getting into a passive device than an active one. Second, we intend to use this device as a rehabilitation device, helping patients in training their muscles and regaining their former control and strength. This device has the following features: 1) it can fully or partially gravity-balance the human leg over the range of its motion; 2) it is tunable to the geometry and inertia of a specific human subject to achieve the desired level of gravity balancing. This orthosis can be potentially used as a rehabilitation device for individuals with severe muscle weakness. The target population will consist of persons who have motion impairments from stroke. The assistance to movement will be lowered as the patient progresses. The improved ability to walk will restore patient independence, lessen the need for assisted living, may allow people to integrate back into society, and return to work.

Gravity balancing has been used in industries for a variety of robotic applications. In this paper, our design uses a hybrid method to balance the weight of the leg in all configurations. It puts the leg in a state of neutral equilibrium everywhere during motion [15]. To check the practical feasibility of the device, a wooden prototype was first fabricated. Then, an engineering prototype using aluminum was made. For a quantitative evaluation of the performance of the device, EMG data of the key muscles involved in the motion of the leg was collected and analyzed. This study was performed on five healthy subjects. Also, an inverse-dynamics study was performed with the force-torque data obtained from the sensors used at the interface of the device and the subject leg. In conjunction with a walking frame and an overhead safety harness, we believe that this device can be used as an effective orthosis for rehabilitation.

II. Theory

The principle involved in gravity-balancing the leg with the hybrid method is as follows [16]: 1) the COM of the leg is geometrically located using a parallelogram mechanism; 2) the springs are placed at suitable positions so that they completely balance the effect of gravity over the range of motion. In this section, the two concepts are described in more detail.

A. Locating the COM

In the current design, we consider only two segments of the leg and approximate the foot as a point mass. These two segments are the thigh and the shank.

A sketch of the mechanism along with springs is shown in Fig. 1. A schematic diagram of this mechanism is shown in Fig. 2. The segments OA and AB are the primary links of the orthosis, whereas DC and CE are the auxiliary links. The mass at the joints are approximated as point masses m_{p1}, m_{p2}, and m_{p3}. The point mass m_{p3} also includes the weight of the foot. Let us define the following terms:

l_i length of the ith link;

l_i^* distance of the COM of the ith primary link from the joint on the previous link;

l_{ai}^* distance of the COM of the ith auxiliary link from the joint on the previous link;

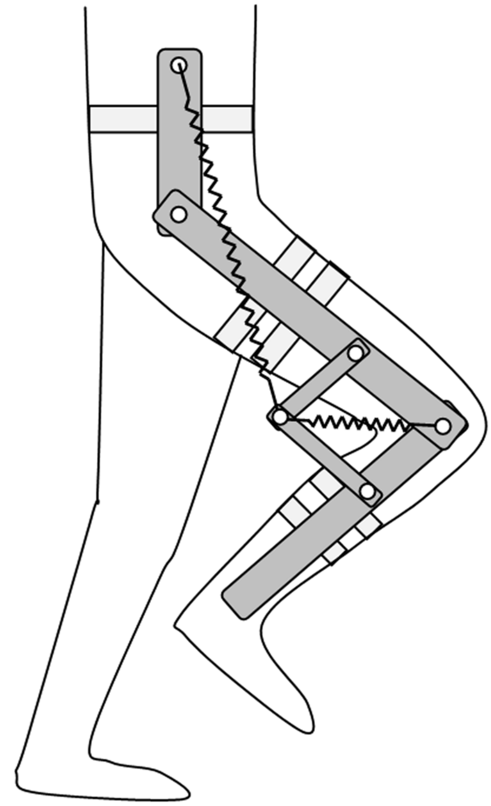
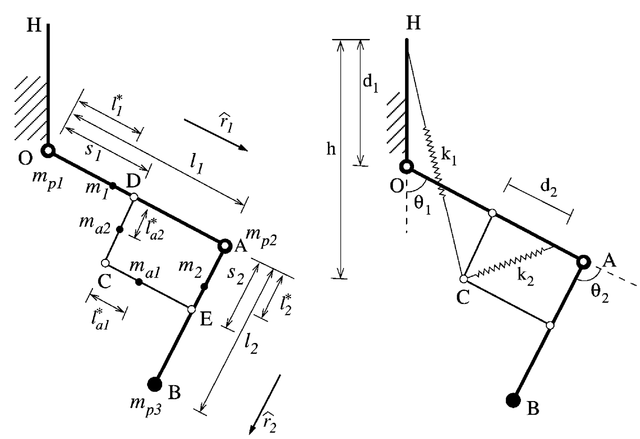


Fig. 1. Basic components of gravity-balancing mechanism.



(a) (b)

Fig. 2. Various terms and parameters of gravity-balancing mechanism.

m_i mass of the ith primary link (mass of the leg segment included);

m_{ai} mass of the ith auxiliary link;

m_{pi} mass of the ith point mass;

$\hat{\mathbf{r}}_i$ unit vector along the ith primary link;

$\mathbf{r}_i$ position vector from the point O to the COM of the ith primary link;

$\mathbf{r}_{ai}$ position vector from the point O to the COM of the ith auxiliary link;

$\mathbf{r}_{pi}$ position vector from the point O to the COM of the ith point mass;

s_1 distance OD in Fig. 2;

s_2 distance AE in Fig. 2.

Among all these quantities, only m_{ai}, s_i, and l_i^* are unknown variables. Also, if we assume that the auxiliary links are made of telescopic members, their mass remains constant, independent of their length. l_{ai}^* would become a linear function of the length of the ith auxiliary link. Hence, the only remaining unknown quantities are s_i. Let

$$\begin{aligned} l_1^* &= \alpha_1 l_1 \\ l_2^* &= \alpha_2 l_2 \\ l_{a1}^* &= \beta_1(l_1 - s_1) \\ l_{a2}^* &= \beta_2 s_2 \end{aligned} \tag{1}$$

where α_i and β_i are the ratios between 0 and 1. The COM of the whole mechanism is given by

$$\mathbf{r}_{OC} = \frac{\sum m_i \mathbf{r}_i}{\sum m_i} \tag{2}$$

where

$$\begin{aligned} \sum m_i \mathbf{r}_i = {} & m_1\mathbf{r}_1 + m_2\mathbf{r}_2 + m_{a1}\mathbf{r}_{a1} + m_{a2}\mathbf{r}_{a2} \\ & + m_{p1}\mathbf{r}_{p1} + m_{p2}\mathbf{r}_{p2} + m_{p3}\mathbf{r}_{p3} \end{aligned} \tag{3}$$

$$\sum m_i = m_1 + m_2 + m_{a1} + m_{a2} + m_{p1} \tag{4}$$

$$+ m_{p2} + m_{p3}. \tag{5}$$

Let us write the vectors $\mathbf{r}_i$, $\mathbf{r}_{ai}$, and $\mathbf{r}_{pi}$ in terms of unit vectors along the primary links $\hat{\mathbf{r}}_i$ in the following:

$$\begin{aligned} \mathbf{r}_1 &= l_1^*\hat{\mathbf{r}}_1 \\ \mathbf{r}_2 &= l_1\hat{\mathbf{r}}_1 + l_2^*\hat{\mathbf{r}}_2 \\ \mathbf{r}_{a1} &= s_1\hat{\mathbf{r}}_1 + s_2\hat{\mathbf{r}}_2 + l_{a1}^*\hat{\mathbf{r}}_1 \\ \mathbf{r}_{a2} &= s_1\hat{\mathbf{r}}_1 + l_{a2}^*\hat{\mathbf{r}}_2 \\ \mathbf{r}_{p1} &= \mathbf{0} \\ \mathbf{r}_{p2} &= l_1\hat{\mathbf{r}}_1 \\ \mathbf{r}_{p3} &= l_1\hat{\mathbf{r}}_1 + l_2\hat{\mathbf{r}}_2. \end{aligned} \tag{6}$$

Since C is the COM of the entire mechanism, $\mathbf{r}_{OC}$ can be written as $\mathbf{r}_{OC} = s_1\hat{\mathbf{r}}_1 + s_2\hat{\mathbf{r}}_2$.

On substituting for $\mathbf{r}_i$, $\mathbf{r}_{ai}$, $\mathbf{r}_{pi}$, l_i^*, l_{ai}^*, and $\mathbf{r}_{OC}$ into (2) and solving for s_1 and s_2, we get

$$\begin{aligned} s_1 &= \frac{l_1(m_1\alpha_1 + m_2 + m_{p3} + m_{a1}\beta_1 + m_{p2})}{m_1 + m_2 + m_{p1} + m_{p2} + m_{p3} + m_{a1}\beta_1} \\ s_2 &= \frac{l_2(m_2\alpha_2 + m_{p3})}{m_1 + m_2 + m_{a2} + m_{p1} + m_{p2} + m_{p3} - m_{a2}\beta_2}. \end{aligned} \tag{7}$$

Hence, with the values of s_1 and s_2 given by (7), the COM of the whole mechanism, including the human leg, can be located and tracked in every configuration. It is important to point out that s_1 and s_2 are proportional to the lengths of primary links l_1 and l_2, respectively, hence the name *scaled lengths* [15]. We expect that the scaled lengths are functions of the distribution of the mass.

B. Gravity Balancing

Gravity balancing is achieved using springs located on the mechanism, as shown in Fig. 2. Our designs use *zero free-length springs*, i.e., the rest lengths of the springs are zero. A zero free-length spring is practically implemented by using a cable and pulley arrangement, shown in Fig. 3.

(a) (b)

Fig. 3. Wooden prototype. (a) Backpack-like design. (b) Mounted on bench.

Let x_1 and x_2 be the extended lengths of the springs, with stiffness k_1 and k_2, respectively. Both springs have one end attached at the COM C. For the gravity to be compensated completely, the total potential energy needs to be constant in all configurations. The expression for the total potential energy is given by $V = (1/2)k_1x_1^2 + (1/2)k_2x_2^2 + Mgh$. Using geometry

$$\begin{aligned} x_1^2 &= \|CH\|^2 \\ &= (d_1 + s_1\cos\theta_1 + s_2\cos(\theta_1 - \theta_2))^2 \\ &\quad + (s_1\sin\theta_1 + s_2\sin(\theta_1 - \theta_2))^2 \\ x_2^2 &= \|CS\|^2 \\ &= d_2^2 + s_2^2 - 2d_2s_2\cos\theta_2 \\ h &= d_1 + s_1\cos\theta_1 + s_2\cos(\theta_1 - \theta_2). \end{aligned} \tag{8}$$

Substituting these values in the expression for V and simplifying

$$V = C_0 + C_1\cos\theta_1 + C_2\cos\theta_2 + C_3\cos(\theta_1 - \theta_2) \tag{9}$$

where

$$\begin{aligned} C_0 &= \frac{1}{2}k_1d_1^2 + \frac{1}{2}k_2d_2^2 + \frac{1}{2}k_1s_1^2 + \frac{1}{2}k_1s_2^2 \\ &\quad + \frac{1}{2}k_2s_2^2 - Mgd_1 \\ C_1 &= k_1s_1d_1 - Mgs_1 \\ C_2 &= k_1s_1s_2 - k_2d_2s_2 \\ C_3 &= k_1s_2d_1 - Mgs_2. \end{aligned} \tag{10}$$

Note that all C_i are constants. If the coefficients of terms containing trigonometric variables vanish, i.e., $C_1 = C_2 = C_3 = 0$, then the total potential energy is given by $V = C_0$, which is a constant. Thus, the total potential energy becomes configuration-invariant, and the gravity balancing is achieved. These conditions yield two independent equations

$$\begin{aligned} k_1 &= \frac{Mg}{d_1} \\ k_2 &= \frac{Mgs_1}{d_1d_2}. \end{aligned} \tag{11}$$

Hence, if two zero free-length springs with stiffness given by (11) are used, the mechanism would become gravity-balanced. Equation (10) shows that the first spring k_1 compensates for the gravity force Mg of the total system, and k_2 helps to make the potential energy invariant with configuration. d_1 and d_2 are arbitrary variables, and can be chosen to vary the level of gravity-balancing.

III. Design

We fabricated two prototypes based on this concept, with wood to test the practical feasibility, and with aluminum to test the effectiveness of the design.

A. Wooden Prototype

The first design was intended to be worn by a person while walking freely [17]. This prototype is shown in Fig. 3(a). In this design, the weight of the leg is taken by the trunk while using this mechanism. Additionally, the weight of the leg and the orthosis can be transmitted to a work bench or a walking frame through the arrangement shown in Fig. 3(b). In Fig. 3(b), an additional vertical translation DOF has been added to the back support system using a parallelogram mechanism, shown in the figure. This interface device could be attached to a walking frame or a treadmill frame so that the forces and torques from the orthosis are transmitted to the frame.

B. Engineering Prototype

An engineering prototype was fabricated in-house, based on preliminary feedback on the wooden prototype from the physical therapy members of the research team. This prototype has the following features: 1) limbs of the machine are made out of lightweight aluminum. Ball bearings are used at the knee and hip joints, and bronze bearings are used at the other joints; 2) limbs are made to be telescopic to accommodate variability in the leg dimensions and inertia across human subjects; 3) the spring locations are adjustable to change the level of gravity during motion, between zero and one gravity; 4) the machine is connected to a walking frame; 5) the backpack attachment to the trunk and the limb attachments to the leg are molded to conform to the contours of a human subject; 6) additional flexibility for the trunk to rotate about a vertical axis is introduced, besides vertical motion.

A picture of the engineering prototype, along with the walking frame and a subject, is shown in Fig. 4. Please note that the white box visible in the figure contains the springs and the pulleys. This initial prototype brought out some useful engineering and clinically relevant observations: 1) the subjects perceived resistance to straightening of the knee (i.e., extension), perhaps due to friction, and too much assistance to bending (i.e., flexion), that pulled the knee into flexion; 2) there is a lateral resistance when straightening the knee from the attachment point of the shank casing with the aluminum bar of the prototype. The lower leg abducts (moves outward) very slightly (about 5°–10°) when the knee is straightened, probably because of the congruency of the surfaces at the knee. Although this is only a slight motion, the lack of lateral "give" of the robot joint caused mild pressure on the lateral fibula; 3) since the device allowed motion in the saggital plane, not allowing any adduction, and there was a limited amount of pelvic translation allowed, the body tended to shift to the left during weight-bearing on the limb wearing the device; 4) the thigh and shank segments in the robot are oriented vertically, while the same segments of the human have some degree of valgus at the knee.

Modifications were made to this prototype based on this feedback. Additional DOFs were introduced at the hip and the pelvis to incorporate pelvic translation and hip abduction/adduction. The anatomical configuration for the thigh segment was incorporated by angling it medially (inward) at the distal end. Bronze bearings were replaced with ball bearings to further reduce the friction. The modified prototype is shown in Fig. 5.

Fig. 4. Engineering prototype mounted on the walking frame and the subject in the gravity-balancing device.

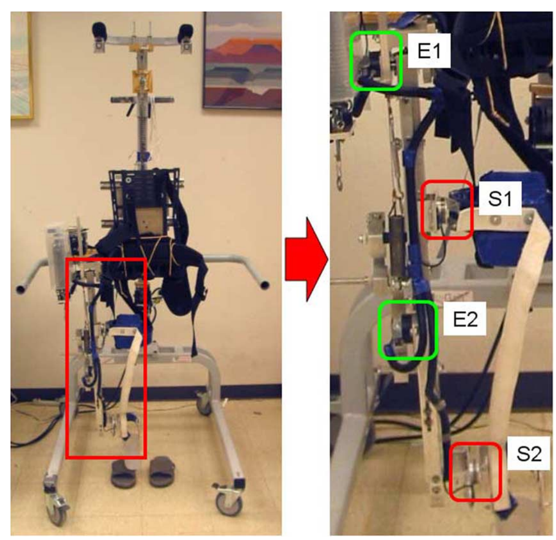


Fig. 5. Position of encoders (E1 and E2) and force-torque sensors (S1 and S2) on the device.

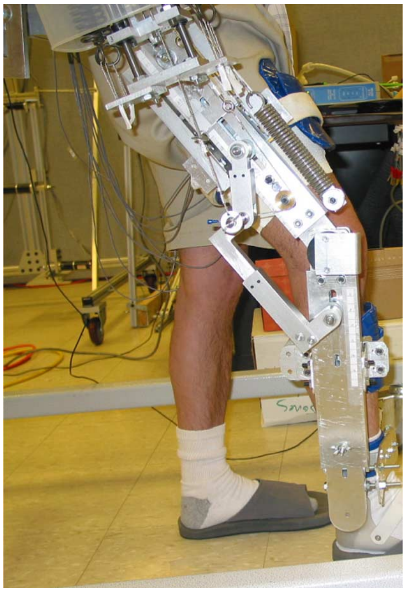


Fig. 6. Modified engineering prototype. Cables for EMG surface electrodes can also be seen.

TABLE I
RANGE OF GEOMETRIC AND INERTIA PARAMETERS OF SUBJECTS

Age (yr)	23 to 48
Gender	M&F
Height (m)	1.60 to 1.78
Body weight (kg)	56 to 82

IV. EXPERIMENT

The modified engineering prototype was used in the experiments, as shown in Fig. 6. In the experiment, EMG data from lower limb muscles of subjects was used to estimate the effectiveness of gravity-balancing at the hip and knee joints. The device is also equipped with encoders and force-torque sensors. The data from these devices were used to determine the motion and torque produced at the hip and knee joints. We hypothesized that lower values of EMG and torque would be seen when the device was set to balance subjects' limbs against gravity than when there was no gravity-balancing of the limb.

Two sets of experiments were performed. In the first experiment, the effectiveness of the device in static limb configurations was tested. Five healthy young adults participated in this experiment. In the second experiment, the device was tested during a leg-raising task and during walking on a treadmill. Tests have been performed to date on five healthy young adults and one individual (66-year-old male) with right hemiparesis following a stroke 2.5 years earlier. Subjects who were not part of the research group gave informed consent according to procedures approved by the institutional review board of the University of Delaware. Table I shows the range of parameters across the subjects involved in the experiments. Following subsections describe the expermental methods used.

A. Experiment I: Tests in Static Configurations

The subjects donned the device that was adjusted such that the hip and knee axes of rotation on the device were aligned with the corresponding axes of the subjects' joints. The spring attachments of the device were adjusted to gravity-balance the limb and the device, so that the subjects could position their limb in various configurations, measured with a goniometer, and maintain these configurations with their muscles "relaxed." All joint angle measurements were made with a hand-held goniometer. Subjects were required to perform two tasks. 1) Hip flexion: The subject stood initially with a mean hip angle of approximately 40° [± 5° standard error (SE)] with his/her foot resting on a one-inch-high block. Upon hearing a computer generated beep, subjects were instructed to lift their limb so that their foot reached an experimenter-specified height. The hip flexion angle at this new position was approximately 60° (± 6° SE across subjects). 2) Knee flexion: The subject stood initially with his/her toes resting on the floor behind the subject, with the knee flexed to approximately 65° (± 6° SE). Again, upon hearing a computer generated beep, they were instructed to lift their toes off the ground to an experimenter-specified height by flexing their knee to approximately 72° (± 7° SE). The knee angle in the hip-flexion task and the hip angle in the knee-flexion task was approximately the same. The limb configuration for a given subject was the same for the two tasks in both conditions; however, different subjects were allowed to find a comfortable position of hip or knee flexion. Subjects performed the static positioning experiments under two conditions: with the "leg and device balanced" and "without device." Five trials were collected for each condition. Trial duration for the conditions with the "leg and device balanced" was 9 s, and "without device" was 6 s, to avoid fatigue. The longer duration for "with the device" is to allow a person to relax the muscles.

B. Experiment II: Tests During a Leg-Raising Task and Treadmill Walking

For this experiment, the device was attached to a treadmill. Subjects were asked to perform two tasks. For the first task, subjects were asked to raise their right limb to a prescribed target position in front of them at a height of about a foot from the treadmill floor. The task was to move their limb to the target and back in synchrony with the beat of a metronome such that each point-to-point motion took approximately 3 s. In this task, the treadmill was not turned on. We will refer to this task as the "leg-raising" task. For the second task, subjects walked on the treadmill. All subjects walked at the same speed as the patient, i.e., 1 mi/h, which was the patient's preferred speed of walking. Three trials of leg-raising and five trials of walking were collected, and the time duration of each trial was about 30 s. The leg-raising and walking tasks were conducted within the device, with either both the leg and device gravity-balanced ("leg and device balanced" condition) or only the device gravity-balanced ("device only balanced" condition). This was done to compare the effects of gravity alone.

For both experiments, EMG data was collected using customized programs written in Labview (National Instruments, Austin, TX). EMG data were collected with a 16-channel EMG system (MA300, Motion Lab Systems, Baton Rouge, LA) at a rate of 1 kHz. For the static-positioning task, surface EMG was recorded from three muscles: rectus femoris, which flexes the hip, and medial (semitendinosus and semimembranosus) and lateral (biceps femoris) hamstring muscles, which flex the knee [18], [19]. The iliopsoas muscle is the primary hip flexor. However, because of this muscle's depth, signals cannot be reliably obtained using surface EMG recordings. Therefore, to limit the usage to surface electrodes, we recorded the rectus femoris muscle, which is also a powerful hip flexor as well as a knee extensor. For the leg-raising and walking tasks, EMG from vastus medialis and vastus lateralis (extensors of the knee) were collected, in addition to the muscles mentioned above.

The EMG recording electrodes with integrated preamplifiers were placed over the bulk of the muscle belly in the direction of the muscle fibers. The centers of the two poles of the electrodes were fixed at 2 cm apart. The EMG signals were preamplified x 20. Offline, the EMG signals were rectified and lowpass filtered with a second-order Butterworth filter at a 15 Hz cutoff frequency. Further, for the static task, the EMG data were integrated over a 1 s time interval, once the limb was in a final resting or elevated position. The device has digital encoders at hip and knee joints. Force-torque sensors are mounted at the interface of human leg and device, one between the thigh segment of the device and the thigh of the subject, the second between the shank segment of the device and the shank of the subject. Data from these were collected by using a D-Space system at 1 kHz sampling rate. The noise from the force-torque sensors was reduced by using a Butterworth second-order filter at a cutoff frequency of 16 Hz. The encoders and sensors on the device are shown in Fig. 5.

V. Measuring Joint Torques

Using data from encoders and force-torque sensors, we can determine the torques applied by the subject at the hip and knee joints by using dynamic force and moment balancing. Following are the terms used in equations:

a_p	linear acceleration at point P;
m_t	mass of human thigh;
m_s	mass of human shank;
W_t	weight of human thigh $= m_t\mathbf{g}$;
W_s	weight of human shank $= m_s\mathbf{g}$;
I_t	moment of inertia of human thigh about its COM;
I_s	moment of inertia of human shank about its COM;
$\mathrm{F}_t, \mathrm{M}_t$	interfacial force and moment at human thigh;
$\mathrm{F}_s, \mathrm{M}_s$	interfacial force and moment at human shank;
$\mathrm{R}_h, \mathrm{T}_h$	reaction force and moment at human hip joint;
$\mathrm{R}_k, \mathrm{T}_k$	reaction force and moment at human knee joint;
C_t	COM of human thigh;
C_s	COM of human shank;
P_{AF}	position vector from point A to the point of application of force F.

Fig. 7 shows the free-body diagrams of forces and moments acting on the human leg segments. Joint angular accelerations were obtained by differentiating joint angles twice.

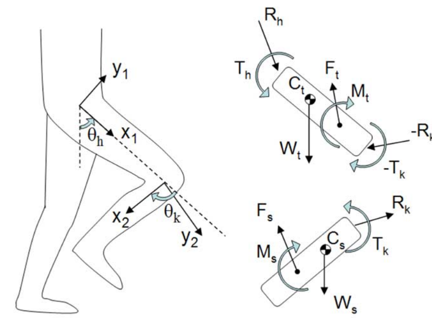


Fig. 7. Coordinates and terms used in force-moment balance.

Force balance for human thigh gives

$$\mathbf{R}_h + \mathbf{F}_t + \mathbf{W}_t + \mathbf{R}_k = m_t\mathbf{a}_{c_t}. \tag{12}$$

Force balance for human shank gives

$$-\mathbf{R}_k + \mathbf{F}_s + \mathbf{W}_s = m_s\mathbf{a}_{\mathbf{c_s}}. \tag{13}$$

Moment balance for human thigh gives

$$\mathbf{P}_{C_tR_h} \times \mathbf{R}_h + \mathbf{P}_{C_tF_t} \times \mathbf{F}_t + \mathbf{P}_{C_tR_k} \times \mathbf{R}_k + \mathbf{T}_h + \mathbf{M}_t + \mathbf{T}_k = \mathbf{I}_t\ddot{\theta}_h. \tag{14}$$

Moment balance for human shank gives

$$-\mathbf{P}_{C_sR_k} \times \mathbf{R}_k + \mathbf{P}_{C_sF_s} \times \mathbf{F}_s - \mathbf{T}_k + \mathbf{M}_s = \mathbf{I}_s\ddot{\theta}_s. \tag{15}$$

Overall, we have four vectorial equations, and four unknown vector quantities R_h, T_h, R_k, and T_k. Hence, these unknown reaction forces and moments can be solved. z components of T_h and T_k give the actual torques applied by the human at hip and knee joints, respectively.

VI. Results

A. Static Positioning Task

Fig. 8 shows the rectified and filtered EMG for a representative subject for the three muscles during the dynamic and static phases of the static positioning task involving either hip flexion or knee flexion, both with and without the device. The left panels show EMGs obtained in the "without device" condition, and the right panels show EMGs in the "leg and device balanced" condition. The top panel shows rectus femoris EMG activity in the hip-flexion task. The middle and bottom panels show the medial and lateral hamstring EMG activity during the knee-flexion task. Note the lower activity of these muscles in the "leg and device balanced" condition (right panels) compared with the "without device" condition. In the final resting position, the limb is expected to be gravity-balanced. This can be seen in the 1-s interval indicated in Fig. 8 between two dotted lines. We integrated EMG (IEMG) over a 1-s interval (corresponding to the dotted lines in Fig. 8), when the limb was held static in the final, flexed position. IEMG for each muscle from the appropriate task

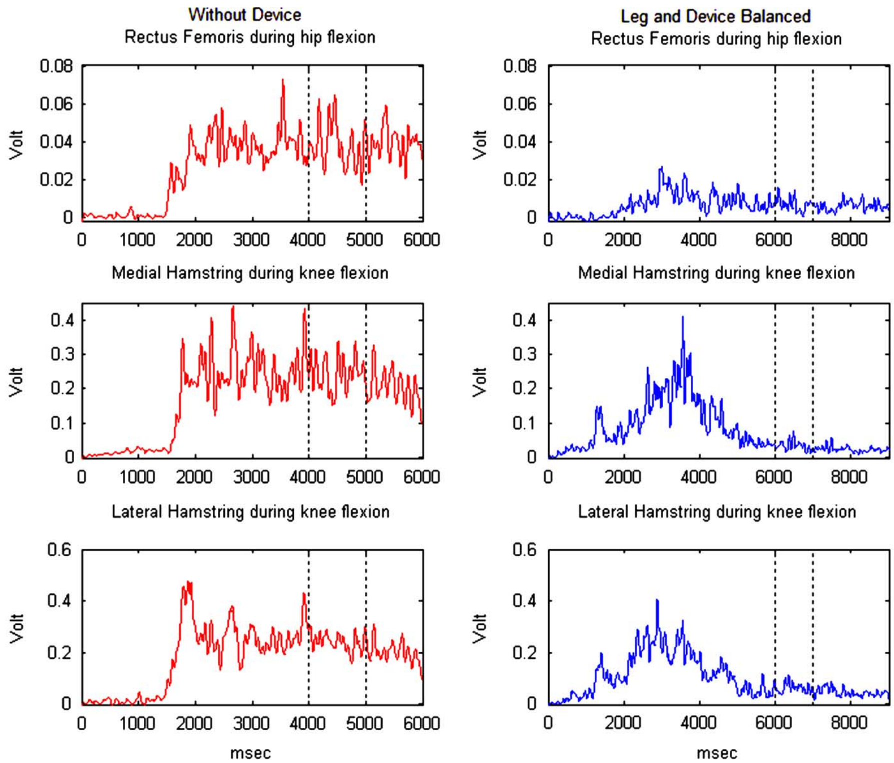


Fig. 8. Rectified and filtered EMG for a representative subject for the three muscles during the dynamic and static phases in two conditions, "without the device" and with "leg and device balancing."

(hip flexion for rectus femoris and knee flexion for the medial and lateral hamstrings) in a subject was normalized to the maximum IEMG obtained from the five trials in the "without device" condition in that subject. These normalized IEMG indices were then expressed as a percentage of the highest IEMG from the "without device" condition. Thus, we expect to see low IEMG percentage values for the "leg and device balanced" condition, and high IEMG percentage values for the "without device" condition. Fig. 9 shows these IEMG percentages averaged across the five subjects with error bars. The blue bars represent the "leg and device balanced" condition, and the gray bars represent the "without device" condition. The left two bars are results for the rectus femoris in the hip-flexion task. The middle and right two sets of bars are the results for the medial and lateral hamstring muscles, respectively, for the knee-flexion task. Note that the IEMG percentages for the "leg and device balanced" condition were always lower than for the "without device" condition. These differences were significant ($p < 0.05$) when tested with a paired t-test for each muscle. Because the average maximum EMG values for the "without device" condition were around 75%–80% (recall that all trials were normalized to the maximum EMG value for the trial showing the highest activity), subjects were not consistent in the amount of EMG activation that they used to flex the hip or knee. This could be due to differences in the contribution of other unmeasured muscles (e.g., iliopsoas) on some trials or greater co-contraction on some trials.

Complete gravity-balancing of the leg and device should have led to zero EMG activity in the recorded muscles in the static position with the device. In fact, although the EMG activity was

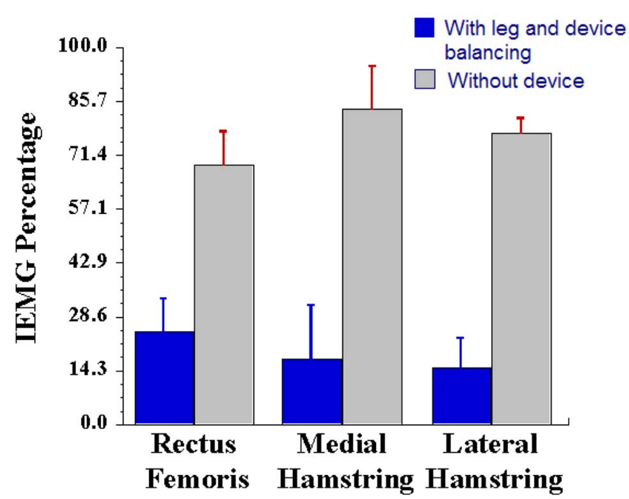


Fig. 9. IEMG percentages averaged across the five subjects for stepping task, with error bars.

substantially and significantly lower in this condition, it was not zero. There may be several explanations for this result. First, while it is possible, in principle, to completely gravity-balance the limb and device, this may be difficult, in practice. For example, factors such as the passive elasticity of the muscles and other soft tissue are not completely taken into account in the current model for balancing the limb. In addition, it may be very difficult for healthy subjects, i.e., without any impairment, to completely relax the muscles of the limb after minimal exposure to the device. This possibility was borne out by the comments of several of the participants, who reported that it was "strange"

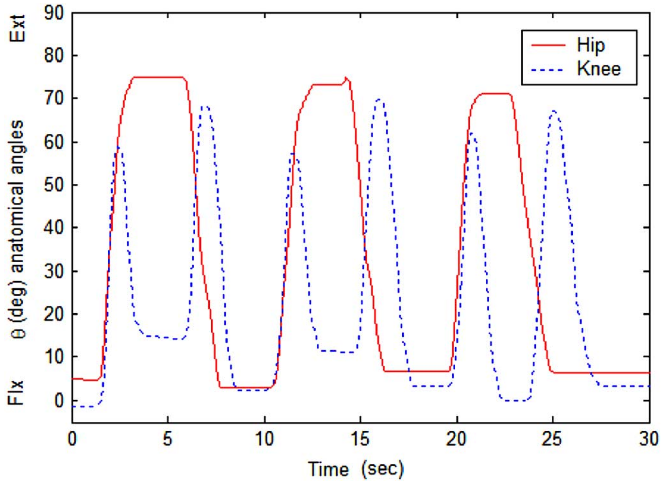


Fig. 10. Joint angles of a representative subject during leg-raising task. Flx: Flexion, Ext: Extension.

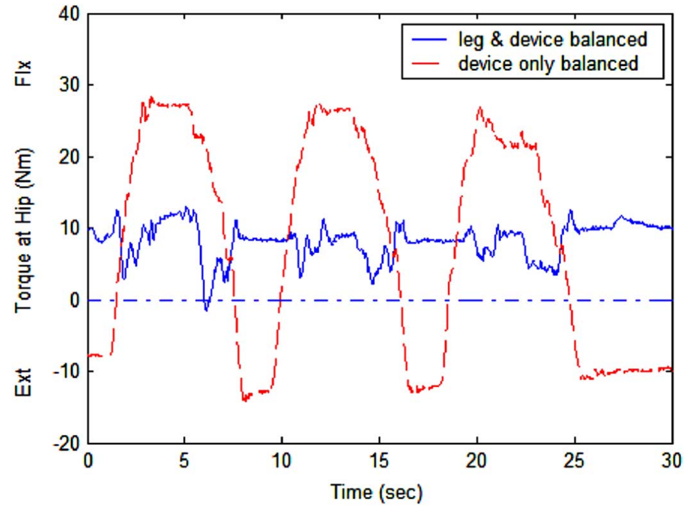


Fig. 11. Torque at human hip joint for leg-raising task at both conditions, "leg and device balanced" and "device only balanced." Flx: Flexion torque, Ext: Extension torque.

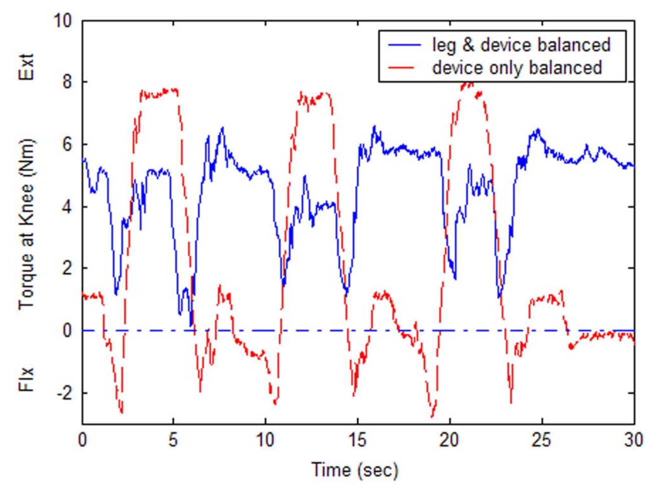


Fig. 12. Torque at human knee joint for leg-raising task at both conditions, "leg and device balanced" and "device only balanced." Flx: Flexion torque, Ext: Extension torque.

to think about lifting the leg up in mid-air and having it stay there while completely relaxed. This suggests that in order for users to completely relax their muscles and allow the device to maximally "help" them, adequate practice is essential. Our device was designed in consultation with an orthotist to ensure that the design of the attachments of our orthosis to the human limb was appropriate. However, all orthotic devices interface with the limb through varying amounts of soft tissue and, therefore, the attachments are not rigid. The soft tissue prevents a precise positioning of the segments, making it difficult to achieve 100% gravity balancing in practice. Nonetheless, the results indicate that the device dramatically diminished the amount of muscle activity required for these static positioning tasks.

B. Leg-Raising Task

In the figures that follow, "leg and device balanced" refers to the condition where both the leg and device are gravity-balanced. The designation "device only balanced" refers to the condition where only the device, but not the weight of the leg, is balanced. Fig. 10 shows the joint angles obtained from optical encoders. It shows three cycles of leg-raising motion. Figs. 11 and 12 show human joint torques in the "device and leg balanced" and "device only balanced" conditions obtained using inverse dynamics with force-torque and encoder data, as explained in the previous section. These plots are for a single trial of one subject's data. Fig. 13 shows the peak value of the joint torques. Gravity-balancing led to a 59.5% and 14.3% decrease in peak hip and knee joint torques, respectively. Fig. 14 shows the mean of human joint torques taken when the leg reaches the target position of the leg-raising motion. This region corresponds to 12.5–13.7 s in joint trajectories, shown in Fig. 10. This is similar to averaging done in EMG plots shown in Fig. 9 for the static positioning task. As in the case of EMG, here too the reduction in joint torque at hip and knee with gravity-balancing of the leg is about 66.8% and 47.3%, respectively. Fig. 15 shows the average of magnitude of torque at hip and knee joints over a period of one complete leg-raising task, including both transient and static phases. This period is from 10.7 s to 16.25 s in Fig. 10. In this case, the hip joint torque is reduced approximately 61.3%, while knee torque increased by approximately 2.7% with gravity-balancing of the leg. One possible reason for this increase in torque at the knee could be the contribution from passive elastic forces in the muscles around the knee joint. Another possible reason could be friction in the joints. This friction is a result of contribution from four joints, the knee joint and three joints of auxiliary links [16]. Second, the magnitude of torque at the knee joint is comparably smaller than the magnitudes of torque at the hip joint; as a result, the contribution to error from undesirable effects were more apparent in the knee joint torque.

C. Treadmill Walking Task

The healthy subjects also walked at several different speeds on a treadmill while wearing the device under both "leg and device balanced" and "device only balanced" conditions. At this point, we also recruited an individual with right hemiparesis to participate in the walking trials. The patient's preferred walking

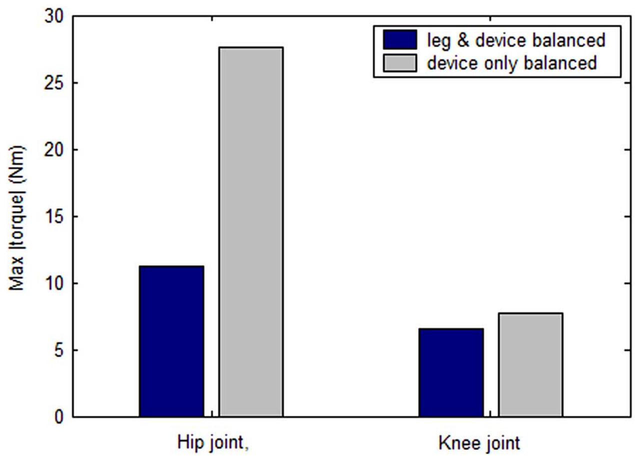


Fig. 13. Maximum magnitude of torque at human hip and knee joints for leg-raising task at both conditions, "leg and device balanced" and "device only balanced."

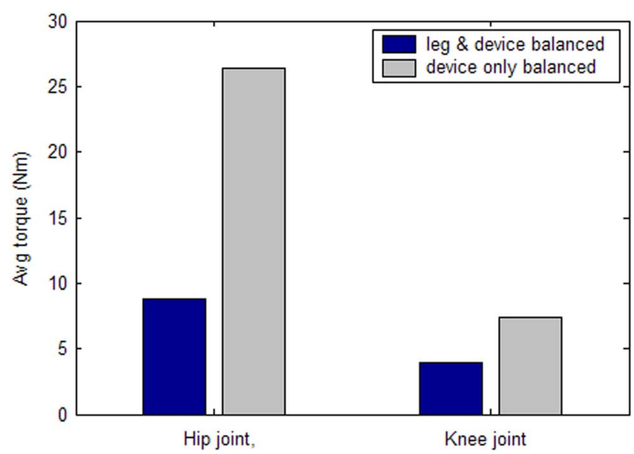


Fig. 14. Average torque at human hip and knee joints at the end of leg-raising task at both conditions, "leg and device balanced" and "device only balanced."

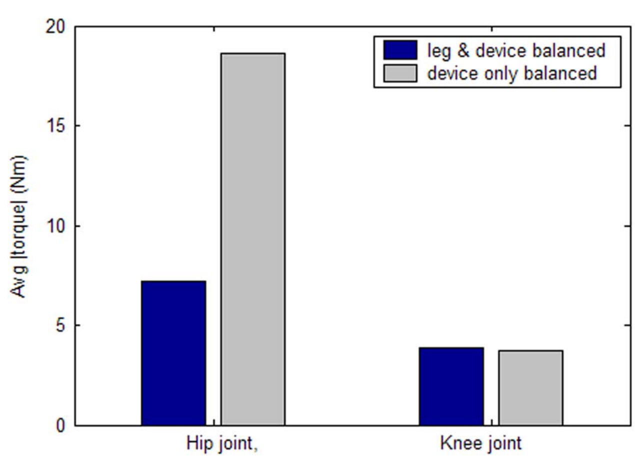


Fig. 15. Average magnitude of torque at human hip and knee joints for leg-raising task at both conditions, "leg and device balanced" and "device only balanced."

speed was 1 mi/h or 0.447 m/s. Therefore, the results for the healthy subjects presented here are for walking at the same approximate speed, which corresponded to 60% of their preferred speed. Figs. 16 and 17 show the contribution of various torques

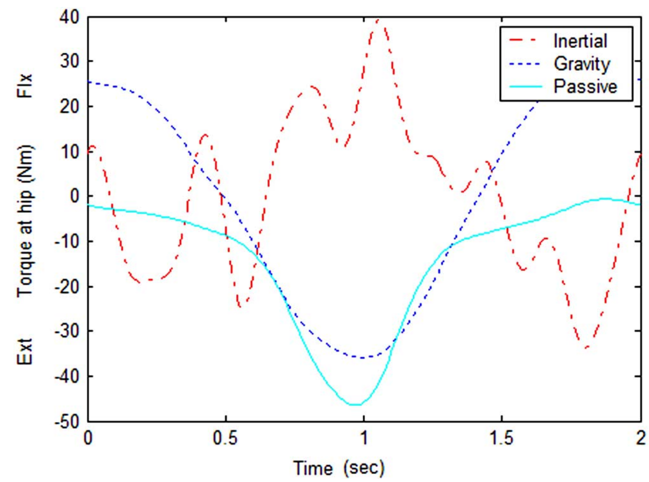


Fig. 16. Various torques at hip joint from simulations based on normal human gait trajectory at a speed of one gait cycle per 2 s. Flx: Flexion torque, Ext: Extension torque.

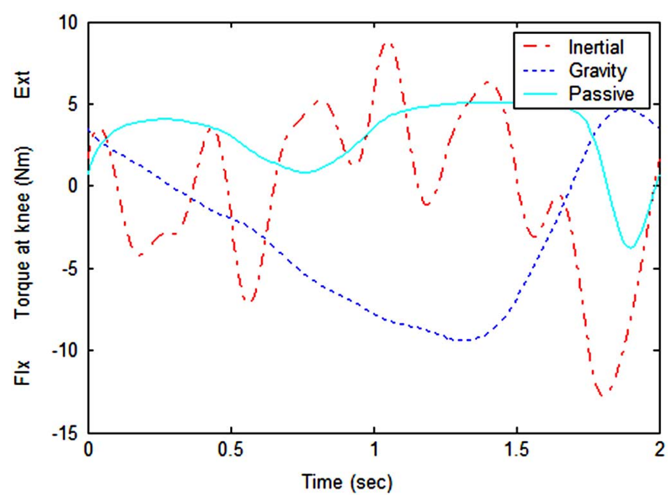


Fig. 17. Various torques at knee joint from simulations based on average human gait trajectory at a speed of one gait cycle per 2 s. Flx: Flexion torque, Ext: Extension torque.

at a speed of one-half gait cycle per second, which is the approximate speed of the subjects in this experiment. These plots were obtained from simulations by using normal gait trajectory [20]. Estimates of passive joint torques were taken from [21]. In the plots, gravity torque is comparable in magnitude with inertial torques. This shows that at this speed, gravity-balancing may not help the subjects to drastically reduce the joint torques. However, very promising results were obtained from these experiments, in terms of increase in the range of motion. Fig. 18 shows the plots of the hip joint angle versus the knee joint angle of a healthy subject performing a walking task. The top panel shows joint angles for the "device only balanced" condition, and the bottom panel shows joint angles for the "leg and device balanced" condition for a representative subject. It is clear from the plots that for the "leg and device balanced" condition, the range of movement at both hip and knee is larger than with device-only balancing. At the hip joint, the increase in range is about 22%, and at the knee joint, the increase in range is about 24%. For the individual with a stroke, this increase in range of joint angles

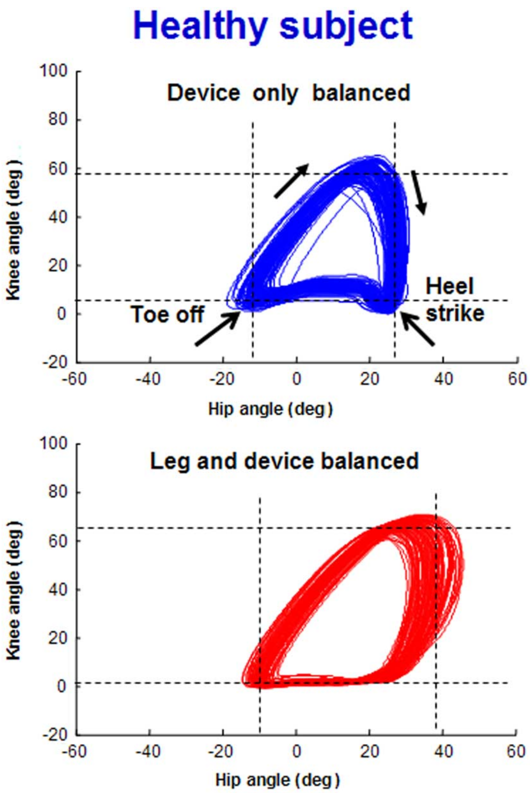


Fig. 18. Hip-joint angle versus knee-joint angle plot of healthy subjects from walking tasks. Data for several gait cycles is shown.

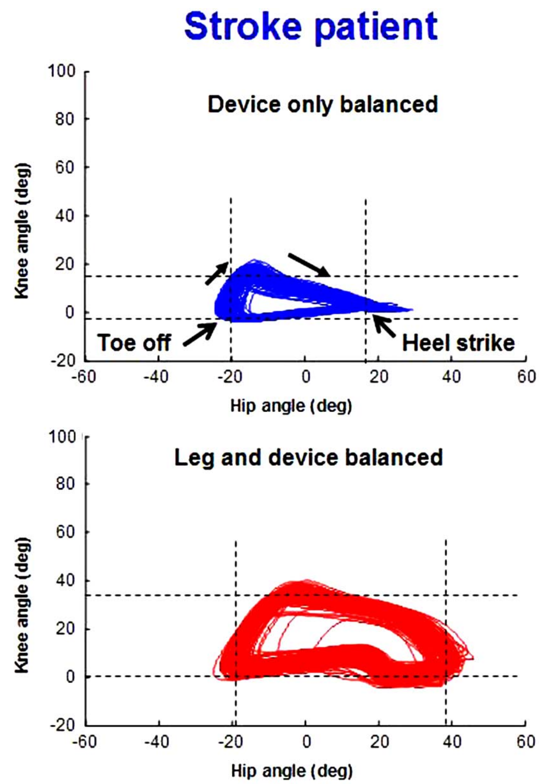


Fig. 19. Hip-joint angle versus knee-joint angle plot of the stroke subject from walking tasks. Data for several gait cycles is shown.

was more prominent than in healthy subjects. Fig. 19 shows the joint angles for the patient. The increase in the range of joint angles is 45% at the hip joint and 85% at the knee joint. Furthermore, the patient voluntarily reported that the device helped him in moving his joints against gravity and allowed him to take longer steps. Actual estimation of the step length showed that the average increase in step length is 5.73%. The increase in amplitude of gait trajectory for the stroke subject is an important positive effect of gravity balancing.

Fig. 20 shows joint torques for the walking task averaged over all trials of one healthy subject in the "leg and device balanced" and "device only balanced" conditions. These plots are normalized time over swing phase. In the stance phase, the subject is not moving the leg against gravity; hence, gravity-balancing does not make much difference (even though gravity of the leg is compensated). In this plot, torque at the hip joint is smaller for the "leg and device balanced" condition, compared with the "device only balanced" condition, for most of the swing phase. However, the knee joint does not show this reduction in torque, and, in fact, shows an increase in torque. Fig. 21 shows the joint torques averaged over all trials of the stroke patient. Here, the joint torques did not show a decrease in the "leg and device balanced" condition. As explained earlier, at the speed at which subjects are walking, inertial torque plays a significant role. Hence, gravity-balancing alone is likely inadequate to reduce the torque magnitudes. In addition, the passive elasticity of the muscles across human joints and friction in the joints of the machine could be the contributing factors, which are not accounted for here. The patterns of muscle activation in individuals with stroke are known to be different from healthy subjects, and may contribute to the lack of an effect of the device on joint torques. In both the leg-raising and walking tasks, the EMGs also did not show differences between the "leg and device balanced" and "device only balanced" conditions. Despite the lack of effects related to EMG and torque for the stroke subject, however, the increase in range of motion of the joints that resulted from gravity-balancing of the leg and device has important implications for improvement in the patient's gait pattern.

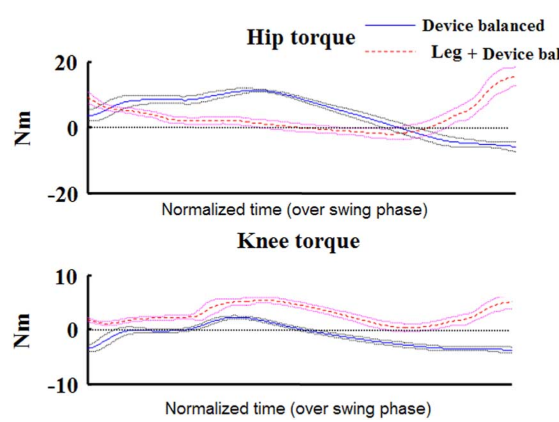


Fig. 20. Joint torques (and its standard deviation) for walking task averaged over all trials of a representative subject.

VII. Discussion

The main purpose of this device is to minimize the effort required to lift the lower limbs against gravity. Patients having

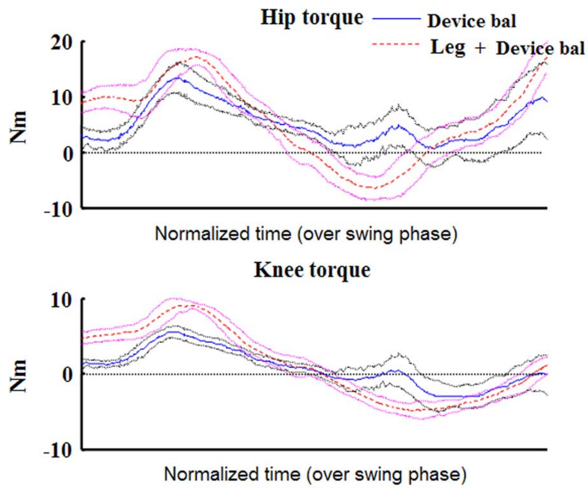


Fig. 21. Joint torques (and its standard deviation) for walking task averaged over all trails of the stroke patient.

weakness of muscles can be expected to benefit from this device, since this device takes away a major source of resistance to their movement, particularly at relatively slow gait speeds. Careful selection of patients who might benefit from this device is required. The ideal patient likely to obtain maximal benefit would be an individual with slow gait and weakness of the leg muscles, such that elimination of gravity would allow them to use their available muscle activation to transport the limb without requiring additional assistance in the form of motors, etc. At present, the device does not compensate for forces such as interaction moments that result from limb dynamics. Thus, for relatively rapid gait speeds, gravity-balancing alone may be inadequate.

In future versions of the device, we plan to compensate for other forces, such as the passive elasticity of the muscles and soft tissue, by modeling these forces and incorporating compensations for them. This study represents an attempt to test the effect of gravity-balancing on muscle activation during static and dynamic tasks. Our tests show that the device generally succeeds in its intended purpose. More detailed studies of its effect on muscle activation and joint torques during gait are currently underway, in parallel with improvements in the device design, to allow movement assistance, where needed, by the addition of motors.

The rationale for the device is that many patients who have suffered a stroke have both primary and secondary muscle weakness, which leads to atypical patterns of muscle activation and resulting movement to compensate for their weakness while still achieving a degree of function [22]. We hypothesize that removing or lessening the weight of the limb will make it easier for the patient to move the limb, and may help patients practice functional movement patterns without the need to develop such compensations, or at least, to limit their extent. This would allow the patient to practice leg movements independently and in the context of locomotion using more normal muscle activation patterns. As the most affected muscles become stronger through such exercise, the amount of gravity compensation can be reduced to increase the load on the muscles and further improve their strength. In this way, we hope to provide a mechanism by which patients can improve function without developing atypical compensations, which are often very difficult to unlearn when more normal muscle strength returns. This device will provide a means for testing this hypothesis.

VIII. Conclusion

In this paper, a leg orthosis was presented that is designed to assist persons with hemiparesis to walk by eliminating the effects of gravity. An engineering metallic prototype was fabricated using aluminum. This aluminum prototype was modified and used in the experiment to check its effectiveness in gravity-balancing. In the experiment, EMG data of the key muscles of the hip and knee were collected and analyzed. This study was performed on five healthy subjects and an individual with right hemiparesis following a stroke. The results showed that the average maximum EMG value for the "leg and device balanced" condition was around 25% of the EMG value for the "without device" conditions for the static experiment. For leg-raising tasks, the average torque at static positioning (at the end of the task) reduced by 66.8% at the hip joint and 47.3% at the knee joint. However, if we include the transient portion of the leg-raising task, the average torque at the hip reduced by 61.3%, and at the knee increased by 2.7%, at knee joints. In the walking experiment, gravity-balancing improved only hip joint torques in the healthy subjects. EMGs were not affected by gravity-balancing. However, there was a positive impact on the range of movement at the hip and knee joints, especially for the stroke patient: the range increased by 45% at the hip joint and by 85% at the knee joint. We believe that this orthosis can be a potential rehabilitation device for individuals with severe muscle weakness.

Acknowledgment

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