A Methodology to Control Walking Speed of Robotic Gait Rehabilitation System using Feasibility-Guaranteed Trajectories

Chan-yul Jung1, Junho Choi2, Shinsuk Park3, and Seung-Jong Kim2

Abstract—This paper presents a novel methodology to control walking speed of an exoskeleton for gait rehabilitation of stroke patients using feasibility-guaranteed trajectories. The controller uses interaction forces to estimate the desired walking speed. Instead of allowing each joint to move around a nominal trajectory, which could lead to infeasible gait patterns, the control algorithm proposed in this paper chooses joint trajectories for desired walking speed, which generates feasible gait patterns. With the interaction forces measured during walking, the walking speed intended by the patient is estimated. Then, based on the estimated walking speed, a reference trajectory stored in a database, which is checked if kinematic constraints required for walking are met, is chosen. Since checking feasibility is performed off-line before the training sessions, it is possible to ensure stability of walking without causing any computational time on-line.

I. INTRODUCTION

Gait rehabilitation using robots has been paid much attention by researchers since robots are suitable for labor-intensive and repetitive tasks such as rehabilitation. During gait rehabilitation using robot systems, it is typical for the patients to remain passive during entire rehabilitation session while the robots move the legs [1]. However, it was found that keeping the patients motivated was critical for effective rehabilitation using robots [2]. In order to keep the patient motivated, controlling the robot system according to the intention of the patients has been widely studied. Banala et al. showed that the patients are allowed to move their legs near the nominal trajectories in [3]. While walking with the exoskeleton, the patients were able to move along the pre-defined trajectories within certain boundaries around the trajectories in Cartesian space. In [4] and [5], the patients were allowed to move their legs so that the trajectories in the joint space were deviated from the pre-defined nominal trajectory. Increased active participation in gait training using robotic orthosis was reported when the patients were allowed to deviate from the pre-defined trajectory in [6]. EMG signals measured from a patient were used to reflect the intention of the patient in [7] and [8].

Changing walking speed according to the intention of the patients is one of ways to keep the patient motivated during training sessions. In [9], the speed of a treadmill was controlled using the position of the patient who was supported by a body weight support system. Since the patient is suspended by a sling from the body weight support system, the patient is able to swing back and forth. For most of gait rehabilitation systems using exoskeletons, however, the positions of the patients are controlled and not changed even if the patients desire to. Therefore, controlling the treadmill speed using changes in position is not appropriate for rehabilitation with exoskeletons. To overcome the problem, interaction force that is applied to an exoskeleton by the patient was measured during swing phase and used to control the treadmill speed [10]. Then, the force was classified into three categories, which indicate three different walking speeds. The threshold values for each category were determined by a physical therapist. In [11], a methodology to change treadmill speed with measured external force acting of the body of the patient was proposed. When walking with an exoskeleton on a treadmill, each joint was controlled to compensate the gravity and friction torque so that the patient was allow to move their joint without any resistance from the robot. Then sheer force on the surface of the treadmill was estimated to be used to control the speed of the treadmill.

Although it is desirable to allow the intention of the patients within the control loop of the robot system in order to motivate the patients, it is not desirable to let the patients entirely control the robot with their intention, since it may produce infeasible gait patterns, which, in turn, results in damage to the robot or the patient. Therefore, it is important to keep the system remain stable while the intention of the patients is allowed to be a part of the control loop of the system.

In this paper, a methodology to reflect the intention of the patients to the speed of walking and to ensure feasibility and stability of the system at the same time is presented. Interaction force between the patient and the exoskeleton was measured. Using the admittance model, desired changes in walking speed are calculated. In order to ensure that the resulted trajectories are feasible and safe, trajectories stored in a database, which are checked for feasibility a priori, are used to change the walking speed. Since only feasible trajectories are used, which are assumed to be available at any desired walking speed, the stability of the system is ensured. Section II describes the COWALK system designed for gait rehabilitation of stroke patients with hemiplegia. Section III explains the control method to change the walking speed.
Section IV shows the experimental result. Conclusion is given in Section V.

II. COWALK

In this section, “COWALK” is introduced. More details on COWALK are found in [12]. COWALK is a gait training system for rehabilitation of stroke patients with hemiplegia. It is composed of an exoskeleton, a treadmill, and a body weight support system, see Fig. 1.

![COWALK system](image)

A. Body Weight Support (BWS)

B. Gravity compensator

C. Exoskeleton

D. Treadmill

Fig. 1. COWALK system is composed of (A) Body Weight Support, (B) Gravity Compensator, (C) Exoskeleton, and (D) Treadmill.

<table>
<thead>
<tr>
<th>Joints</th>
<th>Plane</th>
<th>Motion Planes of COWALK</th>
<th>Actuation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pelvis</td>
<td>Transverse</td>
<td>For-/Backward trans.</td>
<td>Active</td>
</tr>
<tr>
<td></td>
<td>Medial/Lateral trans.</td>
<td></td>
<td>Active</td>
</tr>
<tr>
<td></td>
<td>Rotation</td>
<td></td>
<td>Active</td>
</tr>
<tr>
<td></td>
<td>Upward/Downward trans.</td>
<td></td>
<td>Passive</td>
</tr>
<tr>
<td>Hip</td>
<td>Sagittal</td>
<td>Flexion/Extension</td>
<td>Active</td>
</tr>
<tr>
<td>Corona</td>
<td>Adduction/Abduction</td>
<td>Passive</td>
<td></td>
</tr>
<tr>
<td>Knee</td>
<td>Sagittal</td>
<td>Flexion/Extension</td>
<td>Active</td>
</tr>
<tr>
<td>Corona</td>
<td>Inversion/Eversion</td>
<td>Passive</td>
<td></td>
</tr>
<tr>
<td>Ankle</td>
<td>Sagittal</td>
<td>Dorsiflexion/Plantar Flexion</td>
<td>Active</td>
</tr>
<tr>
<td>Corona</td>
<td>Inversion/Eversion</td>
<td>Passive</td>
<td></td>
</tr>
</tbody>
</table>

The exoskeleton has 14 degrees of freedom. Four of them are for pelvic motion. Nine degrees of freedom are actuated using 10 actuators and 5 degrees of freedom remain passive. Each leg has 5 degrees of freedom, which are extension/flexion and adduction/abduction of the hip, extension/flexion of the knee, and dorsiflexion/plantar flexion and inversion/eversion of the ankle. Actuators are installed at the hip, knee, and ankle for hip extension/flexion, knee extension/flexion, and ankle dorsiflexion/plantar flexion whereas springs are used at the joints for hip adduction/abduction and ankle inversion/eversion, see Table I. Due to the passive joint at the hip and ankle, it is possible for the pelvis to have motion in lateral direction. For pelvic motion, four linear actuators are used to generate translational motions in transverse plane and rotation about the center of the body. Fig. 2 shows the degree of freedom of the COWALK. The joints with dashed circles are passive joints whereas other joints are actuated with linear actuators.

![COWALK system](image)

Fig. 2. Schematics of the COWALK system. It has 14 degrees of freedom. Nine degrees of freedom are actuated and five degrees of freedom, which are indicated by the dashed circles, are passive.

The exoskeleton is attached to the patients at the thighs, calves, and feet using braces. A load cell is attached to each brace at the thighs and calves in order to measure the interaction force between the patient and the exoskeleton, see Fig. 3. The actuators at the joints are equipped with incremental encoders for position measurement of the joint. There is a custom made tachometer to measure the angular velocity of the treadmill.

The gravity compensator supports the weight of the exoskeleton so that the patient on the robot do not have to support the additional load due to the exoskeleton. The gravity compensator is composed of linear springs, cable, and pulleys. The stiffness of the springs as well as other kinematic parameters such as the length of the springs and the location of the pulleys were chosen so that the gravitational torque due to the mass of the exoskeleton is supported regardless of the configuration of the robot.

![COWALK system](image)

Fig. 3. Locations of the load-cell sensors. The load-cell sensor are attached between the braces and the links to measure the interaction force.
III. CONTROL METHOD

In this section, the control methodology is explained. Throughout the paper, the intention of the patient is assumed to be either acceleration or deceleration of walking speed. Other possible intentions such as increasing the step length or flexing the knee are ignored.

The interaction forces are measured by the load-cell sensors attached to the braces to estimate the intention of the user. The estimated intention of the patient is interpreted as acceleration or deceleration of the walking speed. Interaction torques at the joints are calculated using the interaction forces measured with the load-cell sensors attached between the robot and the user. Using the interaction torques at the joints, the intention of the user to change the walking speed is estimated, which results in the desired walking speed. Instead of independently changing a trajectory for each joint according to the measured interaction torque at each joint, a set of trajectories for all joints is selected from the database of trajectory sets to produce the desired walking speed, see Fig. 4.

![Fig. 4. Block diagram of controller of the COWALK](image)

It is possible to alleviate the restriction imposed on the interpretation of the intention if other set of trajectories are included in the database. For example, if the interpreted intention is increasing step length while maintaining walking speed and the trajectories that produce increased step length without changing walking speed are available, the appropriate trajectories are used to control the exoskeleton.

A. Gait Hypotheses

The gait hypotheses (GH) made in this paper are:

GH1) Each leg undergoes alternating swing phase and stance phase, where the swing phase is when the leg is moving forward and the stance phase is when the leg is supporting the body;

GH2) Gait consists of alternating single support phase and double support phase, where the double support phase is when both of the legs are in the stance phase and the single support phase is when one of the leg is in stance phase and the other leg is in the swing phase;

GH3) During the swing phase, the swing foot is always above the ground;

GH4) At any time of walking, all the joints of the patient so no injuries to the patient is caused;

GH5) Feasible joint trajectories are available at any walking speed.

Under the hypotheses, the phases of walking are shown in Fig. 5. The time at the beginning of the $k$-th swing phase is denoted by $t_{SW}^k$ and $t_{DS}^k$ is the time of the beginning of the $k$-th double support phase.

![Fig. 5. Phases of walking](image)

Using the trajectories stored in a database, customized trajectories for a patient are synthesized. The trajectories used to control the exoskeleton are synthesized using trajectories gathered from 113 healthy subjects [13]. The synthesized trajectories are used only if the hypotheses are met. Since the trajectories satisfy the gait hypotheses, the selected joint trajectory is safe to be used. Note that since trajectory synthesis is conducted before training session, it is not necessary to evaluate the feasibility of the trajectory during the session, which is computationally expensive.

B. Desired walking speed

During walking, the interaction forces between the patient and the exoskeleton are measured with the load-cell sensors implemented at the braces of the femur and the tibia. In this research, the interaction forces during swing phases are used to estimate the intention of the users. It is assumed that if the users are willing to walk faster than the current walking speed, the users apply forces to push forward the exoskeleton with the swing leg and if the users are trying to slow down the walking speed, the interaction forces are applied in the backward direction with the swing leg. Then, using the measured interaction force, the walking speed of the exoskeleton is controlled to reflect the intention of the patient.

Let $F_1^i$ be the interaction force at the femur. Let $F_2^i$ be the interaction force at the tibia. The length of the femur is $L_1$ and the length of the tibia is $L_2$. The location of the sensors at the femur and tibia are denoted by $l_1$ and $l_2$, respectively. Fig. 6 shows the schematics of each leg with parameter conventions. Then, joint torques due to the interaction force are given as follows.

$$\begin{bmatrix} \tau_1 \\ \tau_2 \end{bmatrix} = \begin{bmatrix} l_1 F_1^i + (l_2 + L_1 \cos q_2) F_2^i \\ l_2 F_2^i \end{bmatrix}, \quad (1)$$
where \( \tau_1 \) and \( \tau_2 \) are the joint torques at the hip and knee joints, respectively. Let \( J \) be the Jacobian of the robot. Then, the equivalent external force applied at the foot is estimated as
\[
\begin{bmatrix}
F_x^e \\
F_y^e
\end{bmatrix} = (J^T)^{-1} \begin{bmatrix}
\tau_1 \\
\tau_2
\end{bmatrix},
\]
where the Jacobian is given as
\[
J = \begin{bmatrix}
-L_1 \sin(q_1) - L_2 \sin(q_1 + q_2) & -L_2 \sin(q_1 + q_2) \\
L_1 \cos(q_1) + L_2 \cos(q_1 + q_2) & L_2 \cos(q_1 + q_2)
\end{bmatrix}.
\]
Since the determinant of the Jacobian is
\[
|J| = L_1 L_2 \sin q_2,
\]
the Jacobian is invertible if \( \sin q_2 \neq 0 \), which is when the swing leg is fully extended. In order to avoid singularity, the range of motion of the swing knee is controlled such that the knee is not fully extended. This does not limit the gait pattern since the knee is flexed during swing phase for ground clearance. In order to calculate the desired walking speed, only \( F_x^e \) in (2), which is a projection to the horizontal plane, is used.

During the double support phase, the robot needs to satisfy an additional constraint, which is both of the legs being on the ground. Therefore, changing trajectory is not allowed during the double support phase since trajectories for different speed results in different step length. The desired walking speed is allowed to be changed at the beginning of each swing phase and kept constant during the swing phase and the following double support phase. Then, the desired velocity for \( k + 1 \)-th swing phase is calculated as follows.
\[
v_{k+1}^d = a_t v_k^d + \frac{1}{b_t(t_k^{DS} - t_k^{SW})} \int_{t_k^{SW}}^{t_k^{DS}} F_x^e \, dt,
\]
where \( a_t \) and \( b_t \) are constants. Note that the equation (5) is a difference equation whose stability is determined by \( a_t \). It converges to 0 if \( |a_t| < 1 \), remains constant if \( |a_t| = 1 \), and diverges if \( |a_t| > 1 \) when the applied interaction force is 0. If \( 0 \leq a_t < 1 \), it is necessary for the patient to keep applying the interaction force to maintain the walking speed since the walking speed converges to 0 without the interaction forces. Then, the patient need to be more engaged in the gait training. If \( a_t = 1 \), since the desired speed remain unchanged as long as the interaction force remain zero, the patient becomes passive once the desired walking speed is reached. Note that by adjusting \( a_t \), it is possible to change the level of engagement of the patient in controlling the walking speed.

Note that the walking speed of the \( k + 1 \)-th step is calculated using the equivalent force at the ankle during the \( k \)-th step. In results in one-step delayed change of walking speed. For stroke patients with hemiplegia, since they have difficulties in applying interaction forces with their paretic leg as intended, the estimation of the intended walking speed using the interaction force on the paretic leg has errors. In order to avoid the errors, the interaction force is measured during the swing phase of healthy leg. Then, the walking speed is changed at the beginning of the swing phase of the paretic leg, which results in changing walking speed every other steps.

IV. EXPERIMENTS

In this section, the results of a preliminary experiment are presented.

A. Method

A healthy male subject in the COWALK system walked at a given speed at first and was instructed to apply force to change the walking speed according to the a sequence of commands. The commands were:
1) Maintain the walking speed;
2) Accelerate;
3) Maintain the walking speed;
4) Decelerate;
5) Maintain the walking speed;
6) Accelerate;
7) Decelerate;
8) Maintain the walking speed.

The subject was instructed to accelerate the walking speed until the speed reached the maximal speed during acceleration and decelerate the speed until the speed became the minimal speed. The subject was commanded to maintain the walking speed otherwise. Each command was given in sequence every 10 seconds.

Interaction force is measured using a load-cell sensor attached between each brace and link of the robot. The braces are located at the femurs and tibiae of the robot, see Fig. 3.

B. Results

Fig. 7 shows the desired walking speed and measured speed of the treadmill along with the commands given. The solid line is actual speed of the treadmill and the dotted line is the desired walking speed. For safety reason, walking speed of the system has limitation at 2.5 km/h, \( a_t \) is set to be 1, which means that the walking speed remains constant unless the subject applies interaction forces.
Fig. 7. The desired walking speed and actual speed of the treadmill.

Fig. 8 shows the desired walking speeds with different $a_t$. The solid line is the actual speed of the treadmill, the dashed line is the desired speed of the system, and dotted line is corresponding values of $a_t$. The desired speed converges to 0 with $a_t$ less than 1. The convergence rate of the desired speed is determined by $a_t$. As $a_t$ becomes smaller, the rate of convergence becomes faster. Therefore, in order for the desired walking speed to be remain constant, interaction force needs to be applied to compensate the deviation from the desired speed.

Fig. 8. Walking speed with different $a_t$. For $a_t > 1$, acceleration occurs whereas the walking speed decreases for $a_t < 1$.

Fig. 9 compares the interaction forces of the swing leg under different $a_t$. The user was asked to maintain the desired walking speed by applying interaction force if necessary. The swing femur was $-20.71 N$ whereas the mean value of the interaction force of the swing femur is $-1.32 N$ when $a_t = 0.99$. These graphs show that the interaction force required to maintain a certain desired value becomes larger when $a_t$ is smaller. Therefore, by adjusting the value of $a_t$, it is possible to change the level of participation of the patient during gait training.

V. CONCLUSIONS

In this paper, a methodology to control walking speed of a gait rehabilitation system using feasibility-guaranteed trajectories was introduced and implemented in the COWALK system. The intention of the patient, which was interpreted into acceleration or deceleration of the walking speed, was estimated with the interaction force measured with the load-cell sensors between the patient and the robot. The interaction force was used to calculate the equivalent force at the swing foot. Using the equivalent force, the desired walking speed for the next step was obtained. Since the desired walking speed was in the form of a difference equation, the desired walking speed for the next step was modeled with a difference equation. It was found that the level of involvement of the patient was able to be varied. Since trajectories suitable for the desired walking speed, which had been checked for feasibility before the training sessions, were used, the stability of walking was ensured. Experimental result showed that the actual walking speed was varied with different interaction forces.

The experiment showed the feasibility of changing walking speed using interaction force. However, the experiment was carried out with only one healthy subject and further investigation is necessary. It is necessary to conduct experiments with stroke patients with hemiplegia.

REFERENCES