

Development of Gait Rehabilitation System Capable of Assisting Pelvic Movement of Normal Walking

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Gait rehabilitation training with robotic exoskeleton is drawing attention as a method for more advanced gait rehabilitation training. However, most of the rehabilitation robots are mainly focused on locomotion training in the sagittal plane. This study introduces a novel gait rehabilitation system with actuated pelvic motion to generate natural gait motion. The rehabilitation robot developed in this study, COWALK, is a lower-body exoskeleton system with 15 degrees of freedom (DoFs). The COWALK can generate multi-DoF pelvic movement along with leg movements. To produce natural gait patterns, the actuation of pelvic movement is essential. In the COWALK, the pelvic movement mechanism is designed to help hemiplegic patients regain gait balance during gait training. To verify the effectiveness of the developed system, the gait patterns with and without pelvic movement were compared to the normal gait on a treadmill. The experimental results show that the active control of pelvic movement combined with the active control of leg movement can make the gait pattern much more natural.

Key words: exoskeleton, gait rehabilitation, balance control, pelvic movement, gravity compensation

With ever-increasing aging population, the number of patients with motor impairment is on the rise worldwide. Stroke is one of the major neurological disorders that cause motor disability, and it has become a public health issue in many countries [1]. The brain area affected by stroke becomes non-functional, which may lead to motor, speech, or visual impairment.

In the acute phase following stroke, immediate medical treatment should be taken to stabilize the vital functions of the patient. After the acute phase, the

patient is provided with rehabilitation training at the hospital for 6 to 12 weeks. Rehabilitation training provides the patient with systematic ways to help get back to normal daily life. To restore the walking ability, which is essential for active and independent life, the lower-limb gait rehabilitation training is carried out. However, the limited resources of experienced therapists and rehabilitation facilities make it difficult to meet the needs of the increasing number of stroke patients. As one way to alleviate the problems, robotic rehabilitations have been gaining interests for the past years.

Ever since the first commercialization gait rehabilitation system, Lokomat (Hocoma AG, Switzerland),

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was introduced to the public a decade ago, exoskeleton-type robotic systems are being used for gait rehabilitation. Most of the rehabilitation robots are mainly focused on locomotion training in the sagittal plane, such as ALEX [2], Lokomat [3,4], LOPES [5], and WalkTrainer [6].

Mizuno *et al.* proposed two requirements for normal human walking: 1) *locomotion (the ability to initiate and maintain rhythmic stepping)* and 2) *equilibrium (the ability to assume the upright posture and maintain balance)* [7]. The locomotion and equilibrium are the major functions for gait rehabilitation systems, as well as for normal gait. In gait rehabilitation, the equilibrium during gait can be controlled by positioning the center of mass (CoM) of the whole body. Many studies reported that the pelvic movement has large effects on the movement of CoM and thus gait balance control [8-12]. It was also reported that the step length of gait is difficult to control without anterior-posterior motion of the pelvis.

While the control of CoM by pelvic movement plays an important role in gait balance control, none of the robotic gait rehabilitation systems developed so far provide the function of active CoM control. In gait rehabilitation system, unnatural restriction imposed on pelvic movement affects the joint movements of the patient's legs, which in turn causes abnormal gait patterns. Despite the importance of pelvic motion for a well-balanced gait, the developed gait rehabilitation systems have no or little consideration for the actuation of pelvic movement. While Lokomat and ALEX are equipped with actuators to generate joint movements in the hip and knee joints, neither system provides actuated pelvic movement. The LOPES system, on the other hand, can generate translational pelvic movement in the horizontal plane, while allowing only restricted rotational pelvic motion.

Other considerations for design of an exoskeleton-type system include the degrees of freedom (DoFs) and weight of the system. In general, natural gait patterns can be generated with a high number of actuated DoFs in the exoskeleton system. Exoskeleton systems with high DoFs, however, require a large number of mechanical parts, which leads to high complexity of the mechanism and control, high cost, and heavy weight of the whole system [13]. The excessive weight of the exoskeleton worn by the patient can also lead to incorrect gait patterns as well as the discomfort of the patient, which

may result in unsuccessful recovery of normal gait [14].

This paper introduces a novel exoskeleton-type robotic system, COWALK, for gait rehabilitation. The COWALK system is capable of driving the pelvic motion to enable the control of CoM of the patient's body. To relieve the extra weight of the exoskeleton worn by the patient and its resulting discomfort, the system is equipped with a gravity compensation unit. These features of the COWALK system enable the generation of natural gait patterns in training to help the post-stroke patient recover normal gait ability.

Materials and Methods

Gait rehabilitation system. The COWALK system was developed to train the post-stroke patient to correct abnormal gait patterns resulting from hemiplegia. The developed system is composed of three main components: (A) exoskeleton leg unit, (B) pelvis motion unit, and (C) gravity compensation unit (Fig. 1). The exoskeleton leg unit drives the joint movements of the lower extremity. The pelvis motion unit generates the pelvic movement in the horizontal plane. The gravity compensation unit counterbalances the total weight of the exoskeleton leg unit and the pelvis motion unit.

Fig. 2 illustrates the moving parts of the COWALK system. The system has a total of 15 degrees of freedom

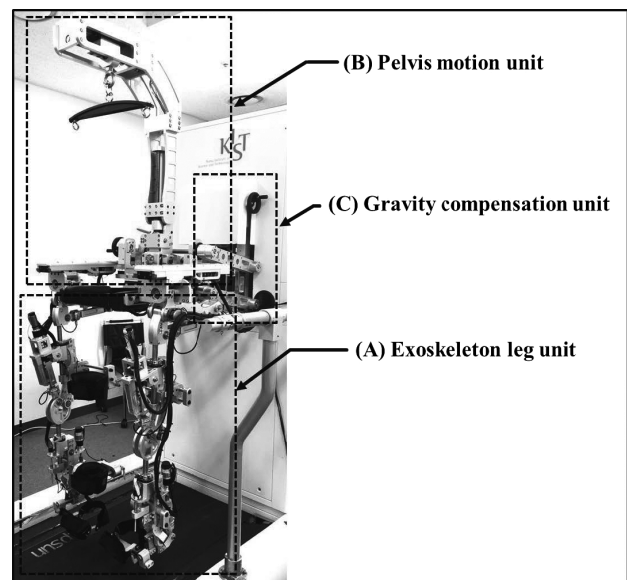


Fig. 1 COWALK system with three main components for gait rehabilitation: (A) Exoskeleton leg unit, (B) Pelvis motion unit, (C) Gravity compensation unit.

(DoFs) with 10 revolute joints and five prismatic joints: three active and two passive DoFs for each of the two exoskeleton legs; 3 active DoFs for pelvic movement; one passive DoF for the vertical movement of the

patient’s whole body; one DoF for anterior-posterior motion of the patient’s upper body. The passive joints are marked with dashed circles in Fig. 2. The dimensions and DoFs of the COWALK system were determined based on anthropometry [15-17]. Table 1 summarizes the joints of the COWALK system and their corresponding DoFs and motions.

1. Exoskeleton leg unit

The exoskeleton leg unit generates the movements of the hip, knee, and ankle joints of both legs using 6 active DoFs and 4 passive DoFs (Fig. 3).

Fig. 2 shows that each of the 2 hip joints has two DoFs: one active DoF (R_{hip_L2} and R_{hip_R2}) for flexion and extension in the sagittal plane, and one passive DoF (R_{hip_L1} and R_{hip_R1}) for adduction and abduction in the coronal plane. Each of 2 knee joints has one active DoF (R_{knee_L} and R_{knee_R}) for flexion and extension in the sagittal plane. Each of 2 ankle joints has two DoFs: one active DoF (R_{ankle_L2} and R_{ankle_R2}) for dorsiflexion and

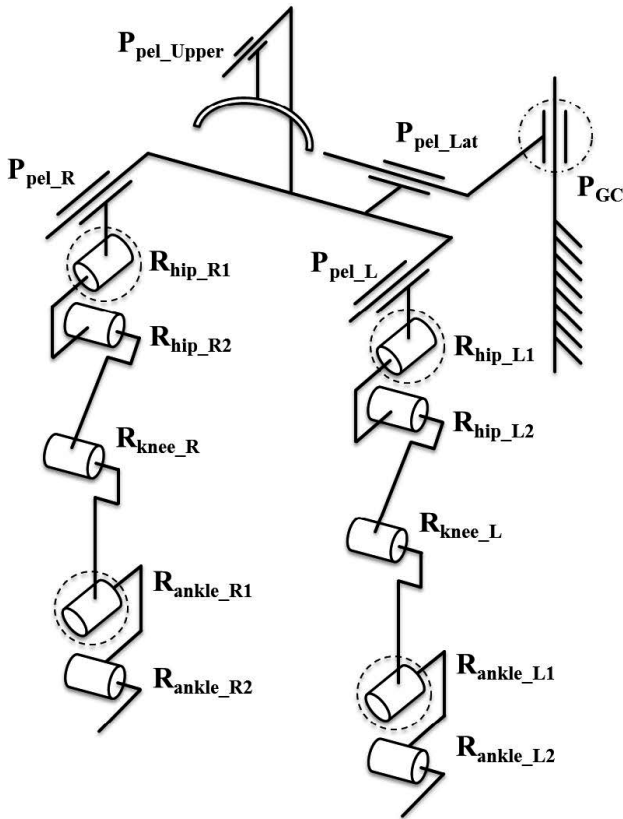


Fig. 2 Schematics of the COWALK system: PGC is DoF of gravity compensation unit in the sagittal plane, P_{pel_lat} , P_{pel_L} , P_{pel_R} and P_{pel_upper} are DoFs of pelvic motion unit in the horizontal and sagittal plane, R_{hip_L1} , R_{hip_L2} , R_{hip_R1} , and R_{hip_R2} are DoFs of the hip joints, R_{knee_L} and R_{knee_R} are DoFs of the knee joints, R_{ankle_L1} , R_{ankle_L2} , R_{ankle_R1} , and R_{ankle_R2} are DoFs of the ankle joints. The hip, knee and ankle joints are DoFs of exoskeleton leg unit. Dotted circles denote passive DoFs equipped with leaf and coil springs.

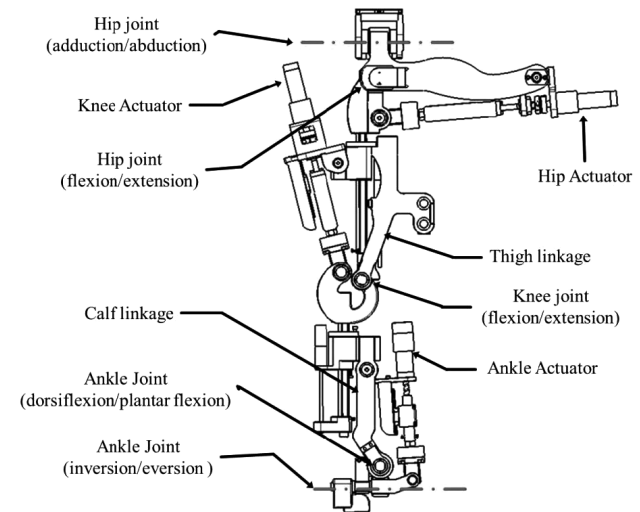


Fig. 3 Side view of the exoskeleton leg unit.

Table 1 Motion planes and DoFs of COWALK

Component	DOF	Motion
Pelvis motion unit	1 × 2 (pelvis)	Anterior-posterior translation
	1 (pelvis)	Lateral-medial translation
	1 (upper body)	Anterior-posterior translation
Exoskeleton leg unit	2 × 2 (hip joint)	Adduction and abduction
	2 × 2 (hip and knee joint)	Flexion and extension
	1 × 2 (ankle joint)	Dorsiflexion and plantar flexion
	1 × 2 (ankle joint)	Inversion and eversion
Gravity compensation unit	1 (whole body)	Vertical translation

plantar flexion in the sagittal plane and one passive DoF (R_{ankle_L1} and R_{ankle_R1}) for inversion and eversion in the coronal plane. The passive joints (R_{hip_L1} , R_{hip_R1} , R_{ankle_L1} , and R_{ankle_R1}) are mechanically loaded with leaf springs.

Fig. 4 shows the equilibrium position of the passive joints compared with Q angle of the knee. The normal value of Q angle is $13.5^\circ \pm 4.5^\circ$ for healthy people [18], and the equilibrium angle for the passive hip joint is set to be around 5° based on the normal Q angle.

For actuation of each of 6 active DoFs, a linear actuator equipped with a ball screw and a Brushless DC electric motor (Maxon EC-4pole 200 W) is implemented. The joint forces and angles are measured by

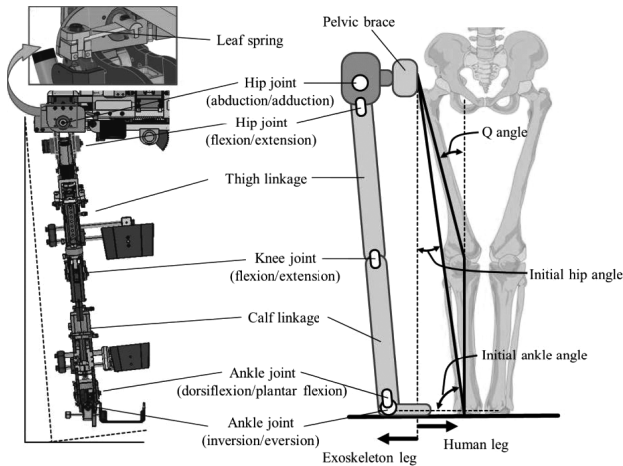


Fig. 4 The equilibrium angles of the exoskeleton leg compared with the Q angle of the human skeleton in the coronal plane.

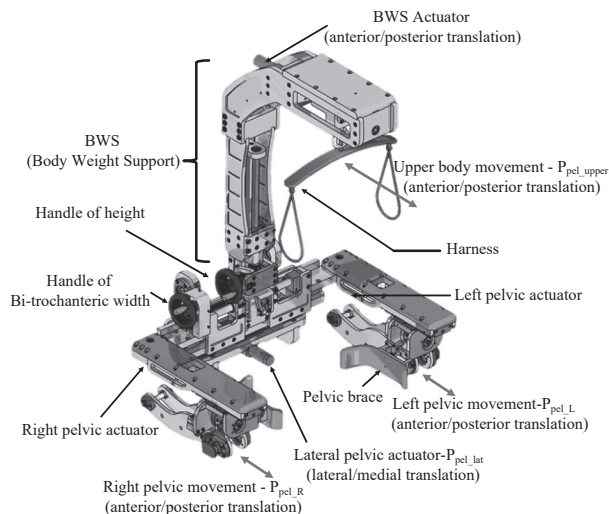


Fig. 5 Pelvis motion unit with 4 active prismatic joints.

load cells and incremental encoders, respectively.

2. Pelvis motion unit

The pelvis motion unit generates the movement of the pelvis in the horizontal plane using 4 active DoFs.

As shown in Fig. 5, the pelvis driving mechanism has 4 active prismatic joints (P_{pel_L} , P_{pel_R} , P_{pel_upper} , and P_{pel_lat}). The linear actuator in P_{pel_upper} generates the anterior-posterior translation for upper body of the patient. The linear actuators in P_{pel_L} and P_{pel_R} generate the anterior-posterior translation of the left and right sides of the pelvis. The rack-and-pinion mechanism in P_{pel_lat} generates the lateral translations of the center of the pelvis, respectively. These joints (P_{pel_L} , P_{pel_R} , and P_{pel_lat}) allow three DoF motion of the pelvis in the horizontal plane: the rotation, the anterior-posterior translation, and the lateral translation in the horizontal plane, as illustrated in Fig. 2 and Fig. 6. This feature of the pelvis motion unit enables the shift of the center of mass of the patient.

To support the weight of the patient, the patient is suspended in a harness by the Body Weight Support (BWS) system that is implemented in the pelvis motion unit. The patient is attached to the COWALK system by braces at the pelvis, thighs, and calves. The braces have rigid supports with straps and pads to hold the braces tightly to the skin surface [19]. They limit the relative motion between the patient and the system. The locations of the braces in the COWALK system can be adjusted by the patient. The braces at the thighs and calves are equipped with load cells to monitor the interaction force between the patient and the system in real time.

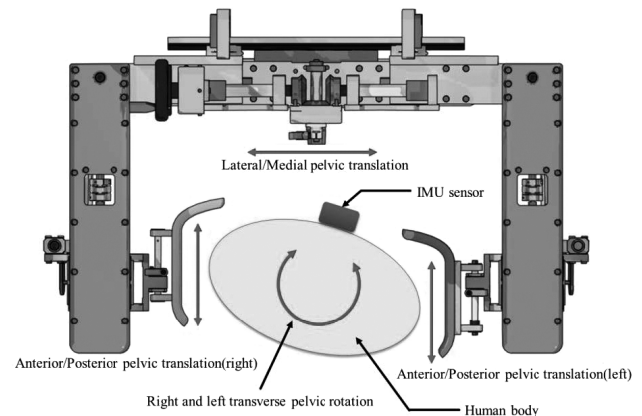


Fig. 6 Top view of the pelvis motion unit and corresponding pelvic movement of the patient.

3. Gravity compensation unit

The total weight of the exoskeleton leg unit and the pelvis motion unit described above is around 122.4 kg. The heavy-weight exoskeleton worn by the patient can cause discomfort or even pain, as well as unnatural motion pattern.

The gravity compensation unit counterbalances the weight of the exoskeleton leg unit and the pelvis motion unit to relieve the discomfort of the patient and to prevent from producing unnatural gait motion.

The gravity compensation unit is composed of a parallelogram linkage mechanism, a pulley-wire mechanism, and counterbalancing weight and spring. Fig. 7 (a) shows the schematic view of the gravity compensation unit developed in our previous work [20]. The mechanism of the unit is designed to compensate for the

weight regardless of the vertical position of the weight. The vertical motion of the 4-bar linkage in Fig. 7 (a) corresponds to the passive prismatic joint P_{GC} in Fig. 2. It was reported that the vertical displacement of the center of gravity is typically around 5 cm for normal gait [20]. The values for the counterbalancing weight and spring were determined so that the peak-to-peak amplitude of the vertical motion is maintained at around 5 cm.

The deflection S of counterbalancing spring k is described as:

$$S = \sqrt{h^2 + b^2 + 2bh\cos\theta} - S_0 \quad (1)$$

Here, h is the distance between the hinge joints $P1$ and $P3$, and b is the distance between the hinge joint $P3$ and wire attach point $P2$ on the link, as shown in Fig. 7 (a). θ is the angular displacement of the parallelogram linkage. S_0 is the equilibrium length of the counterbalancing spring when $\theta = 180^\circ$ ($S_0 = h - b$).

The potential energy stored in the gravity compensation mechanism is calculated as:

$$V = \frac{1}{2}KS^2 - mgl\cos\theta + m_e gS \quad (2)$$

Here, m is the total mass of the exoskeleton leg unit and the pelvis motion unit, and m_e is the counterbalancing mass. l is the distance between the hinge joints $P3$ and $P4$ shown in Fig. 7 (a). The first term represents the elastic potential energy ($V_k = kS^2/2$) of spring k , and the second and third terms represent the gravitational potential energy of masses m and m_e , respectively ($V_m = mgl\cos\theta + m_e gS$).

By plugging Eq. (1) into Eq. (2), the total potential energy becomes:

$$V = \frac{1}{2}k(h^2 + b^2 + S_0^2) - m_e gS_0 + (kbh - mgl)\cos\theta - (kS_0 - m_e g)\sqrt{h^2 + b^2 + 2bh\cos\theta} \quad (3)$$

To make the total potential energy invariant for all values of θ , we set the counterbalancing spring k and counterbalancing weight m_e as follows:

$$k = \frac{mgl}{bh} \quad (4)$$

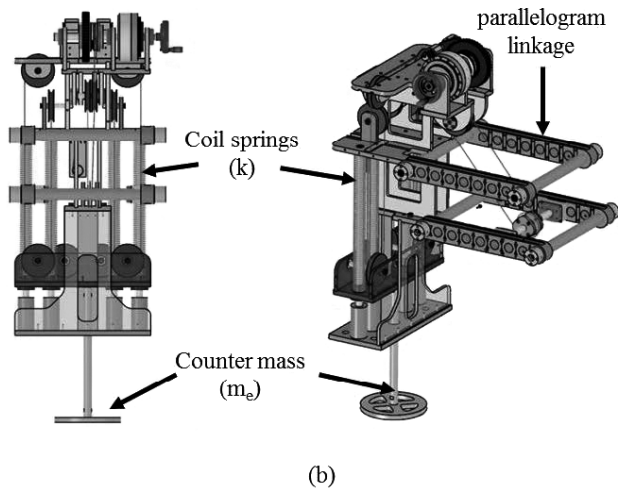
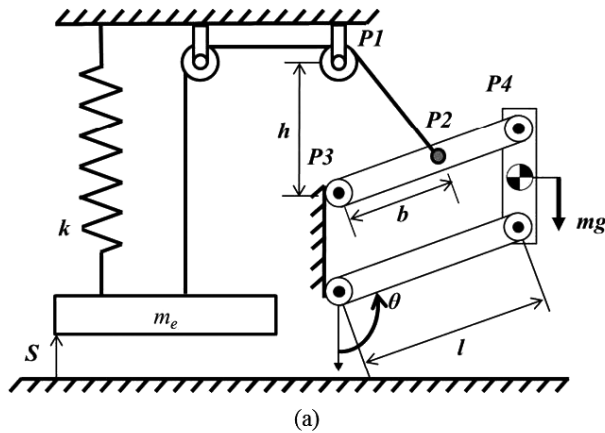


Fig. 7 Gravity compensation unit: (a) Schematics of parallelogram link mechanism, (b) CAD model of gravity compensation unit.

$$m_e = \frac{kS_0}{g} \quad (5)$$

The total potential energy becomes invariant as follows:

$$V = \frac{1}{2}k(h^2 + b^2) \quad (6)$$

4. Control system

Fig. 8 shows the schematic diagram of the control system of the COWALK system. The control system is composed of the trajectory generator and a joint controller. The trajectory generator produces the reference joint trajectories for normal gait. The reference trajectories were generated based on the 3D motion capture data acquired from 113 human subjects during normal gait [21]. The joint trajectory data of the human subjects was collected while freely walking on a treadmill without the exoskeleton system. Using the joint trajectory as a reference input, the joint controller with PD feedback control commands the joint movement of the COWALK system. The motor command to the COWALK system is regulated by using the mathematical models of gravity and friction compensation as shown in Fig. 8.

The control system was implemented on a PC running in real time using xPC Target and MATLAB/Simulink software. The PC is connected with servo controllers (ELMO G-DC-WHI5/100EE) and sensors (load cells and encoders) through AD/DA converters and analog and digital I/O driver blocks (PCI 6602 and 6703, NI Corp., USA). A detailed explanation of the control system can be found in our previous study [22].

Gait experiment. To evaluate the performance of the COWALK system, we carried out experiments with

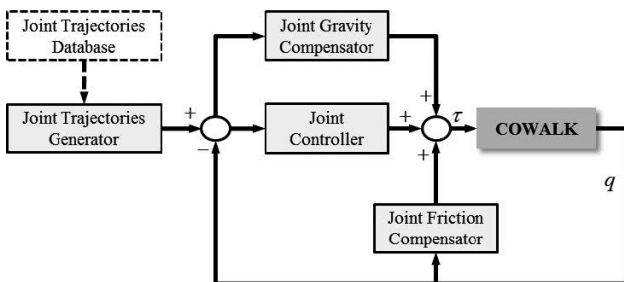


Fig. 8 Block diagram of the controller of COWALK.

a healthy subject to evaluate the performance of the developed system and with 2 post-stroke hemiplegic subjects as a preliminary clinical study. All the subjects were males with the age ranging from 36 to 65 years old. Both hemiplegic subjects were at Brunnstrom stage 5 for their left lower limbs, and they had FAC scores of 5, which denotes that they were capable of independent ambulation on level surfaces. The detailed information of the 3 subjects is summarized in Table 2. In the experiments, the subjects, both healthy and post-stroke, were suspended over a treadmill by the BWS system and attached to the exoskeleton leg unit and the pelvis motion unit by the braces. The gait speed was controlled by the treadmill.

Gait experiments were performed with 3 different settings on a treadmill:

- (1) *Free Walking (FW)*: FW for healthy subject (H_FW) and FW for hemiplegic subject (He_FW)
- (2) *Actuated Legs and Actuated Pelvis (ALAP)*: ALAP for healthy subject (H_ALAP) and ALAP for hemiplegic subject (He_ALAP)
- (3) *Actuated Legs and Locked Pelvis (ALLP)*: ALLP for healthy subject (H_ALLP) and ALLP for hemiplegic subject (He_ALLP)

In the FW mode, the human subjects walk freely on the treadmill without using the COWALK system. In the ALAP mode, the human subjects walk in the exoskeleton both with robot-assisted leg motion and with robot-assisted pelvis motion. In the ALLP mode, the human subjects walk in the exoskeleton with robot-assisted leg motion, but the pelvic motion of the human subjects are locked by the pelvis motion unit.

The speed of the treadmill was set at 2 km/h. Each experimental session was conducted 3 times for duration of 2 min. During the gait experiment session, the rotational and translational motion of the pelvis and the plantar pressure on the foot soles were measured by an IMU (inertia measurement unit) and insoles (capacitive sensors), respectively.

The Shimmer3 Bridge Amplifier (Shimmer, Dublin, Ireland) was attached to the back of the subject, at approximately the L4 vertebra level as illustrated in Fig. 6. The Pedar-X insole system (Novel, St. Paul, MN, USA) is composed of an array of 99 capacitive sensors in the shape of sole. This sensor system records the plantar pressure distribution during the gait and estimates the trajectory of the center of pressure (CoP) based on the recorded pressure distribution.

While each experimental session was performed for the duration of 2 min, a minute's worth of data (from 30 sec to 90 sec from the beginning of the session) was used for gait analysis to avoid the effect of transient gait

Table 2 Information of subjects

	Subject 01 (Healthy)	Subject 02 (Hemiplegia)	Subject 03 (Hemiplegia)
Age	36	65	63
Sex	M	M	M
Height (cm)	170	172	160
Weight (kg)	53	77	63
Side of hemiparesis	Non	Left	Left
Duration from onset (Months)	-	19	11
FAC scoresa	-	5	5
Brunnstrom stageb	-	5	5

^aFAC: Functional Ambulation Category

1. Nonfunctional Ambulator: Patient cannot ambulate, ambulates in parallel bars only, or requires supervision or physical assistance from more than one person to ambulate safely outside of parallel bars.
2. Ambulator—Dependent for Physical Assistance—Level II: Patient requires manual contact of no more than one person during ambulation on level surfaces to prevent falling. Manual contact is continuous and necessary to support body weight, as well as to maintain balance or assist coordination.
3. Ambulator—Dependent for Physical Assistance—Level I: Patient requires manual contact of no more than one person during ambulation on level surfaces to prevent falling. Manual contact consists of continuous or intermittent light touch to assist balance or coordination.
4. Ambulator—Dependent for Supervision: Patient can ambulate on level surfaces without manual contact of another person but, for safety, requires stand-by guarding of no more than one person because of poor judgment, questionable cardiac status, or the need for verbal cuing to complete the task.
5. Ambulator—Independent, Level Surfaces Only: Patient can ambulate independently on level surfaces, but requires supervision or physical assistance to negotiate any of the following: stairs, inclines, or nonlevel surfaces.
6. Ambulator—Independent: Patient can ambulate independently on nonlevel and level surfaces, stairs, and inclines.

^bBrunnstrom stage

- Stage 1: No activation of the limb
- Stage 2: Spasticity appears, and weak basic flexor and extensor synergies are present
- Stage 3: Spasticity is prominent;the patient voluntarily moves the limb, but muscle activation is all within the synergy pattern
- Stage 4: The patient begins to activate muscles selectively outside the flexor and extensor synergies
- Stage 5: Isolated movements are performed in a smooth, phasic, well-coordinated manner

patterns. The gait cycle was determined by dividing the raw data from the IMU and the capacitive sensors using a peak detection algorithm (Fig. 9 (a)). The representa-

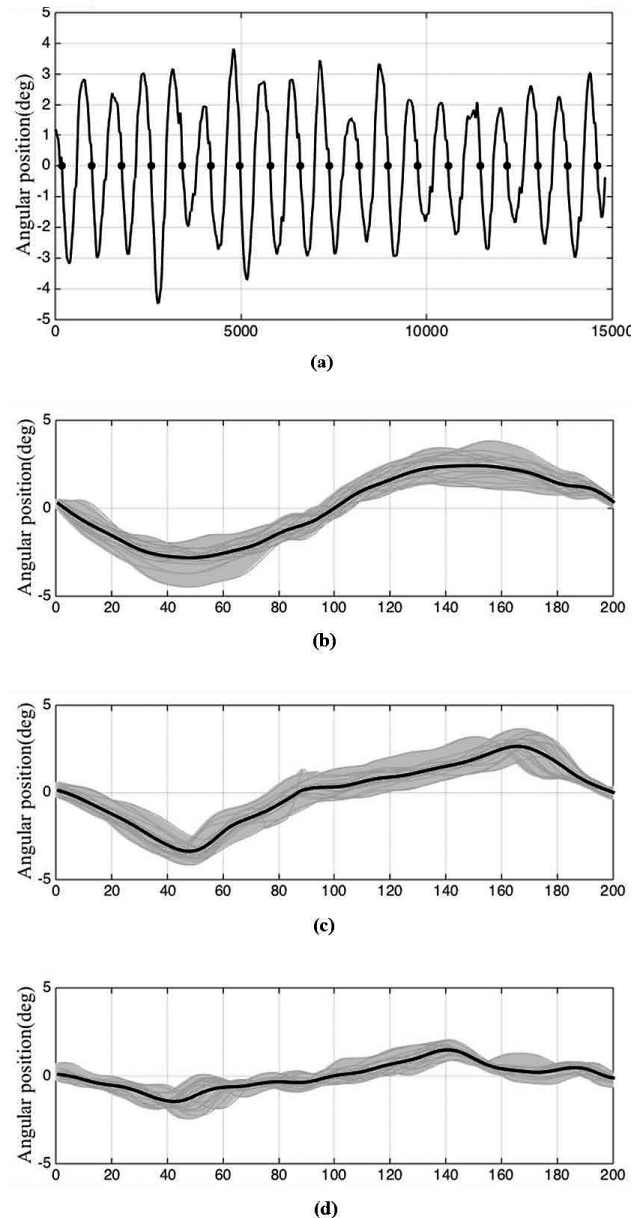


Fig. 9 Ensemble average of raw data (healthy human subject): (a) the phase detection example of the IMU raw data of natural walking (black-thin line), and phase detection point (black dot point), (b) the ensemble average result of pelvic rotation in the transverse plane (H_FW), (c) the ensemble average result of pelvic rotation in the transverse plane (H_ALAP), (d) the ensemble average result of pelvic rotation in the transverse plane (H_ALLP). In the (b), (c), and (d): ensemble average value (block-thin line), one gait phase data set (gray-thin line).

tive gait cycle was then obtained by ensemble-averaging of over 30 ± 5 gait cycles (one subject) as shown in Fig. 9 (b), (c), and (d).

The gait experiments with the COWALK system were approved by the Institutional Review Board (IRB 2016-004 and IRB 2017-036) at the Korea Institute Science and Technology (KIST).

Results

Healthy subject. Fig. 10 compares the yaw angles of the pelvis (the pelvic rotations in the horizontal plane) of the healthy subject measured from the IMU sensor under the three experimental conditions (H_FW, H_ALAP, and H_ALLP) with the reference of data from natural, overground walking [23]. The figure shows that the yaw angle from H_FW (free walking on the treadmill) ranging from -3° to $+3^\circ$ has a similar pattern to that of natural, overground walking. As can be seen in the figure, H_ALAP ranges from -3° to $+3^\circ$ as H_FW, while the range of H_ALLP is only $\pm 1^\circ$.

Cross-correlation function (CCF) analysis allows us to quantify the resemblance of two plots in the time domain. To further check the similarities of the plots of Fig. 10, we calculated CCFs between H_ALAP and H_FW and, between H_ALLP and H_FW.

Fig. 11 compares the two CCFs with the reference of the auto-correlation function of the H_FW plot of Fig. 11.

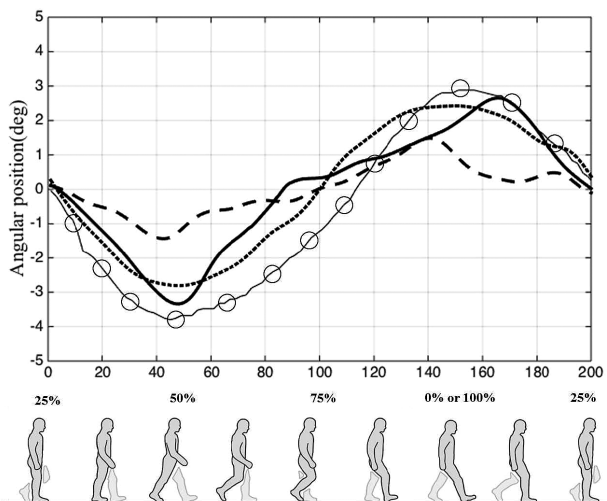


Fig. 10 Pelvic rotation of healthy human subject in the horizontal plane: reference data during overground walking (thin solid line with \circ mark, figure adapted from *Levens et. al* [23]), H_FW (dotted line), H_ALAP (solid line), H_ALLP (dashed line).

The CCFs are normalized with the auto-correlation function of H_FW. As can be seen in the figure, the peak values of the CCF of H_ALAP and H_ALLP are 0.8713 and 0.184, respectively. This indicates that the similarity between the yaw angles of H_ALAP and H_FW is much higher than that between H_ALLP and H_FW.

Fig. 12 (a) compares the three-dimensional trajectories of pelvis position under the three experimental conditions (H_FW, H_ALAP, and H_ALLP), and Fig. 12 (b) shows the projections of the trajectories on the horizontal plane. The trajectories were obtained by time-integrating the acceleration data from the IMU sensor and then ensemble-averaging the time-integrated data. The obtained data demonstrates the typical butterfly-shaped trajectory, as *Zhao et al.* reported in their study on the pelvis motion during gait [24].

As can be seen in Fig. 12 (a) and (b), the pelvis trajectory of H_ALAP is very much similar to that of H_FW, while the range of the pelvic movement of H_ALLP is restricted. The average difference between H_ALAP and H_FW is 8 ± 16.7 mm, while that between H_ALLP and H_FW is 25.2 ± 80.5 mm.

Fig. 13 compares the trajectories of CoP on the sole during stance phase under the three different experimental conditions (H_FW, H_ALAP, and H_ALLP). The figure shows that while the CoP trajectories of both H_ALAP and H_ALLP are medial to that of H_FW, the trajectory of H_ALAP is closer to that of H_FW in comparison with H_ALLP. The average differences between the lateral CoP trajectories of H_ALAP and

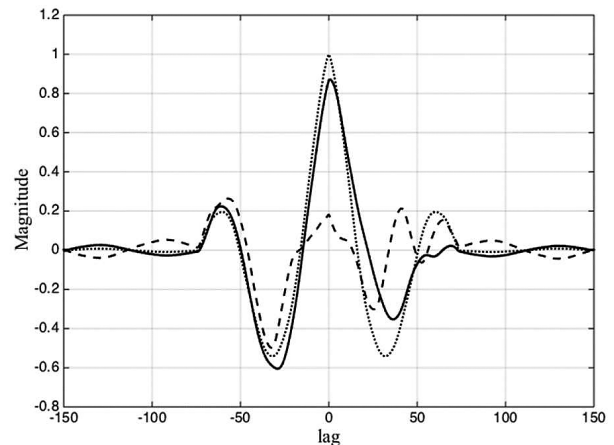
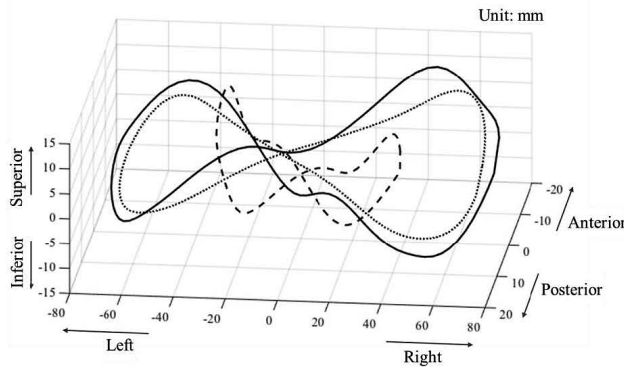


Fig. 11 Cross-correlation functions between H_FW and H_ALAP and between H_ALLP and H_FW: H_ALAP and H_FW (solid line), H_ALLP and H_FW (dashed line), compared with auto-correlation of H_FW (dotted line).

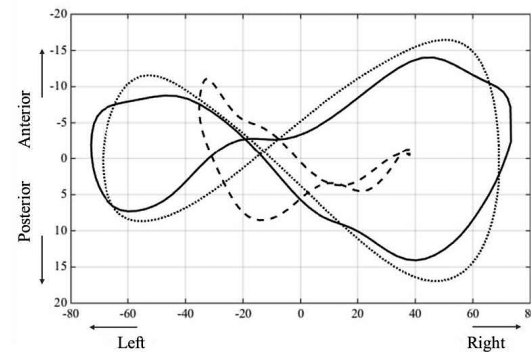
H_FW and between those of H_ALLP and H_FW are summarized in Table 3.

Hemiplegic subjects. Fig. 14 compares the angles of pelvic rotation in the horizontal plane measured from the IMU sensor under the three experimental conditions (He_FW, He_ALAP, and He_ALLP) of the hemiplegic subjects with the reference of data from natural, overground walking [23]. In the figure, the data from H_FW in Fig. 10 is included for comparison. The figure shows that the rotation angle for He_FW ranges from -14° to $+14^\circ$. The hemiplegic subjects appear to walk with excessive pelvic rotation with an abnormal pattern. The average difference between the rotation angles for

He_FW and H_FW is 6.55° , and that between He_FW and reference of data from natural, overground walking [23] is 6.17° . The detailed data of pelvic rotation is summarized in Table 4. The CCFs are normalized with the auto-correlation function of H_FW of the healthy subject. The peak values of the CCFs for He_ALAP and He_ALLP are 0.9973 and 0.9971, respectively.



(a)



(b)

Fig. 12 Trajectories of pelvis position (a) three-dimensional trajectory, (b) projection of trajectories on the horizontal plane: H_FW (dotted line), H_ALAP (solid line), and H_ALLP (dashed line).

Table 3 Average lateral differences between CoP Trajectories of the healthy subject

	Left Foot	Right Foot
H_ALAP and H_FW	6.9 ± 5.4 mm	2.4 ± 2.3 mm
H_ALLP and H_FW	10.6 ± 6.3 mm	5.4 ± 4.2 mm

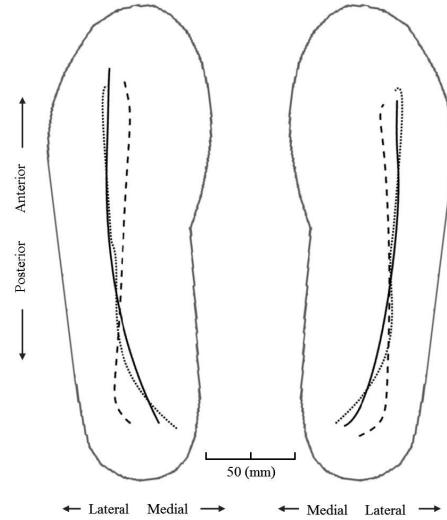


Fig. 13 CoP trajectories (left and right soles): H_FW (dotted line), H_ALAP (solid line), and H_ALLP (dashed line).

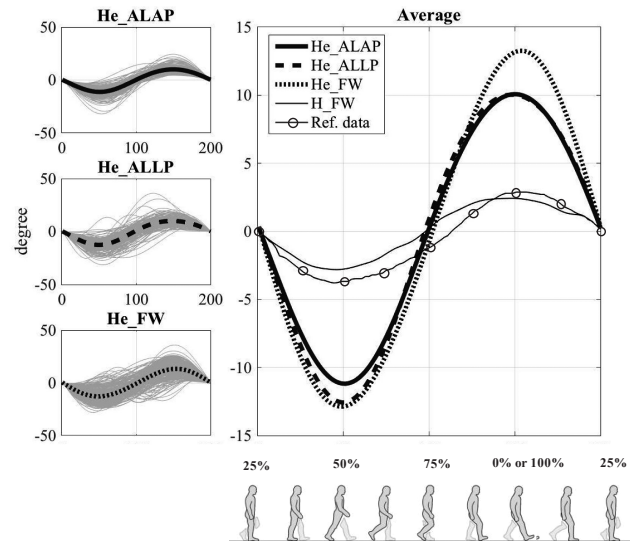


Fig. 14 Ensemble average of pelvic rotation (hemiplegic subjects) in the horizontal plane: reference data during overground walking (thin solid line with \circ mark, figure adapted from *Levens et. al* [23]), H_FW (thin solid line), He_FW (dotted line), He_ALAP (solid line), He_ALLP (dashed line).

Fig.15 shows the lateral movements of the pelvis for He_FW, He_ALAP, and He_ALLP with that of H_FW. As can be seen in the figure, the lateral movement for He_ALAP is closer to that of H_FW than that for He_ALLP, while the range of the pelvic movement of He_ALLP is restricted. The average difference between He_ALAP and H_FW is 8.11 mm, while that between He_ALLP and H_FW is 17.63 mm.

Fig.16 compares the lateral CoP trajectories for He_FW, He_ALAP, and He_ALLP with that for H_FW during stance phase. The average differences between the lateral CoP trajectories of He_ALAP and H_FW and between those of He_ALLP and H_FW are summarized in Table 4. The results from this analysis for the hemiplegic subjects were somewhat mixed. While the average difference between He_ALAP and H_FW is smaller than that between He_ALLP and H_FW as shown in Table 5, the patterns of the CoP trajectories appear to be quite different from that of the natural free walking of the healthy subject.

The experimental results with the healthy and hemi-

plegic subjects indicate that robot-assisted pelvic motion combined with robot-assisted leg motion can generate a gait pattern close to a natural walking on a treadmill or a natural, overground walking.

In conclusion, we developed a gait rehabilitation system to generate natural pelvic movements combined with exoskeleton leg motion. By active control of the pelvic motion, the COWALK system enables the control of the patient's CoM, which is essential for balance control during gait.

The gravity compensation unit is equipped in the system to relieve the patient from the heavy weight of the exoskeleton system. To test the effectiveness of the actuated pelvic movement in generating natural gait patterns, we carried out experiments with a human subject (healthy subject and hemiplegic patients). The experimental results show that the active control of pelvic movement combined with the active control of leg movement can create a natural gait pattern that is very close to a natural gait on a treadmill or in the overground condition. This also shows that the COWALK

Table 4 Average differences between the pelvic rotation angles of the hemiplegic subjects

	H_FW	Reference data
He_FW	6.55°	6.17°
He_ALAP	5.01°	4.70°
He_ALLP	5.48°	5.18°

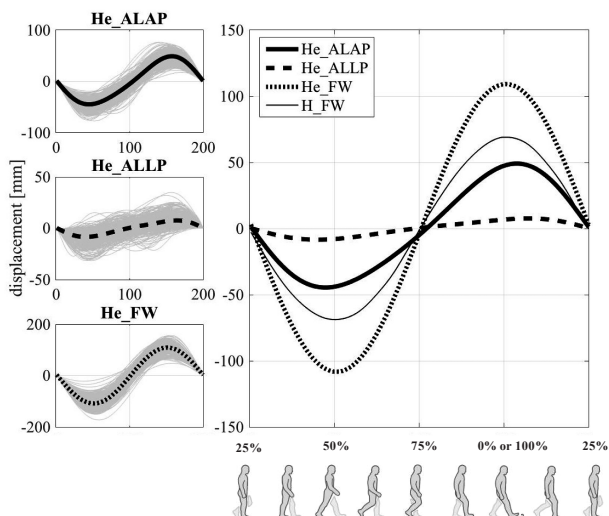


Fig. 15 Ensemble average of lateral pelvic movement (hemiplegic subjects) in the horizontal plane: H_FW (thin solid line), He_FW (dotted line), He_ALAP (solid line), He_ALLP (dashed line).

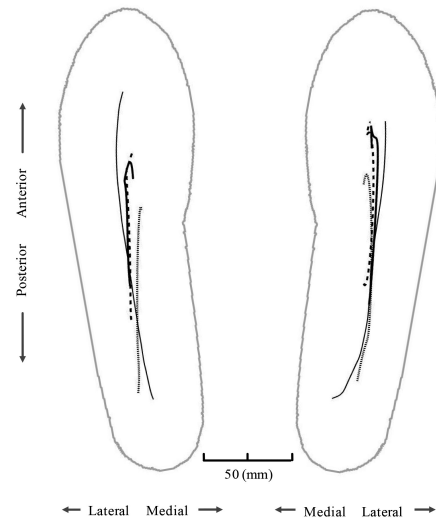


Fig. 16 CoP trajectories (left and right soles): He_FW (dotted line), H_ALAP (solid line), and H_ALLP (dashed line), and H_FW (thin solid line).

Table 5 Average lateral differences between CoP trajectories of the hemiplegic subjects

	Left Foot	Right Foot
He_ALAP and H_FW	4.20 mm (max. 5.80)	8.45 mm (max. 12.33)
He_ALLP and H_FW	4.66 mm (max. 7.57)	10.46 mm (max. 14.63)

system can be effectively used in gait rehabilitation to help the post-stroke patients recover their normal gait ability.

For our future works, we will clinically evaluate the developed rehabilitation system with post-stroke patients with hemiplegia. We are also planning to develop a new version of the COWALK system that can be used in the overground condition.

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