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Motion-Based Control Strategy of Knee Actuated Exoskeletal Gait Orthosis for Hemiplegic Patients: A Feasibility Study

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Abstract: In this study, we developed a unilateral knee actuated exoskeletal gait orthosis (KAEGO) for hemiplegic patients to conduct gait training in real-world environments without spatial limitations. For this purpose, it is crucial that the controller interacts with the patient's gait intentions. This study newly proposes a simple gait control strategy that detects the gait state and recognizes the patient's gait intentions using only the motion information of the lower limbs obtained from an embedded inertial measurement units (IMU) sensor and a knee angle sensor without employing ground reaction force (GRF) sensors. In addition, a torque generation method based on negative damping was newly applied as a method to determine the appropriate amount of assistive torque to support flexion or extension movements of the knee joint. To validate the performance of the developed KAEGO and the effectiveness of our proposed gait control strategy, we conducted walking tests with a hemiplegic patient. These tests included verifying the accuracy of gait recognition and comparing the metabolic cost of transport (COT). The experimental results confirmed that our gait control approach effectively recognizes the patient's gait intentions without GRF sensors and reduces the metabolic cost by approximately 8% compared to not wearing the device.

Keywords: exoskeletal gait orthosis; GRG sensor; stroke; hemiplegia; gait intention detection

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1. Introduction

Stroke or other acquired brain injuries often lead to significant physical impairments, including hemiplegia, motor disorders, cognitive decline, and language impairments [1]. Among these, gait impairments caused by spinal or brain injuries particularly impact a patient's independence and quality of life but also predispose the patient to secondary medical complications, such as pressure sores, cardiovascular disorders, and musculoskeletal diseases [2,3]. These complications are often exacerbated by a decline in the function of various organs due to reduced motor activity. In this context, regular rehabilitation training becomes a critical component of the care plan, aimed at minimizing these risks [3–5].

Traditional rehabilitation therapies, primarily focused on restoring walking functions, involve manual gait training by physical therapists or the use of exercise equipment to enhance lower limb muscle strength, endurance, and joint range of motion (ROM). However, these methods often heavily rely on the experience of therapists, which can lead to variability in treatment effectiveness [6]. To overcome these limitations, various lower limb robot-assisted gait training systems and exoskeleton robots have been developed over the past several decades, and their effectiveness has been verified in numerous clinical studies [7–11]. This progress has extended the role of physical therapists beyond providing physical assistance to also performing a supervisory role, thereby contributing to the efficiency of wearable robot-assisted gait training [7,10].

In the previous study, there was an increasing emphasis on overground gait training in real-world environments. Such training not only facilitates active participation and interaction with the environment but also enhances trunk balance, control, and visuo-spatial awareness. However, it is not advisable to use full lower-limb exoskeletons for overground gait training in patients who retain some functional abilities, such as those with hemiplegia [1]. In line with this, recent research has explored the use of lightweight lower limb exoskeletons, which assist specific joints and reflect the wearer's gait intention [1,12–15]. These exoskeletons are not only cost-effective and easy to wear but also enable training in real-world environments without spatial constraints.

Hemiplegic gait after stroke is characterized by unilateral muscle weakness, spasticity, and insufficient forward propulsion, resulting in an asymmetric walking pattern. Key issues include limitations in hip, knee, and ankle movements during the swing phase, causing undesirable compensatory movements, like hip hiking and circumduction. In the stance phase, notably, knee hyperextension is observed in about 40–68% of stroke patients, significantly affecting gait speed and symmetry, and increasing the metabolic cost of walking [1,16,17]. Addressing these abnormal gaits requires innovative solutions like the knee-actuated exoskeletal gait orthosis (KAEGO), which assists knee joint flexion and extension based on the patient's gait intentions [16,18–21].

To develop such control strategies, accurate detection of the user's gait phase and recognition of gait intentions are essential. This has led to the use of various sensors, including electroencephalographic (EEG) [22], electromyographic (EMG), pressure, torque, angle sensors, and inertial measurement units (IMUs) [23]. Among these, ground reaction force (GRF) sensors have been widely used for gait phase detection [19,24–27]. In a previous study of this research topic, a GRF sensor using load cells was developed and successfully implemented in the bilateral KAEGO for a patient with an incomplete spinal cord injury (SCI) [24]. However, the placement of the sensors in the foot makes the sensors susceptible to damage due to continuous impacts [28].

This study developed a unilateral KAEGO for hemiplegic patient gait training and proposes a simple gait control strategy that employs an IMU sensor and knee angle sensor to detect gait phases and recognize the patient's gait intentions, eliminating the need for GRF sensors. Additionally, we applied a simple torque generation method based on knee angular velocity to assist the knee joint movements [13]. This paper aims to evaluate the effectiveness of this approach by assessing the accuracy of gait recognition and comfortable gait assistance through two types of walk tests.

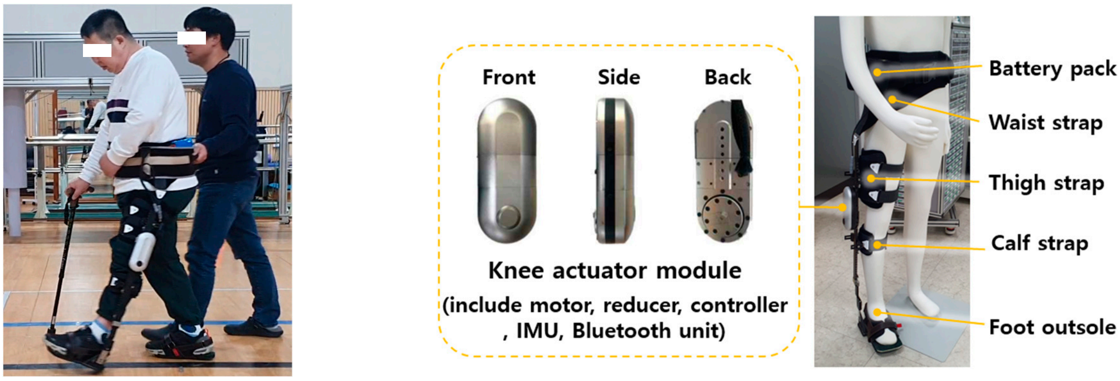
The first experiment involved a hemiplegic patient and a healthy individual, each performing 10 m walk tests [29], during which data were collected to analyze the accuracy of gait recognition and how quickly the gait intentions were identified. In the second experiment, we hypothesized that if the developed KAEGO provided appropriate assistive torque, it would lead to a decrease in metabolic cost during walking, despite the potential discomfort from the weight (3.4 kg) increase and wearability of the device. To test this hypothesis, we adopted a method of measuring the metabolic cost of transport (COT) during a six-minute walk test [24,30,31].

The 10 m walking test (WT) results showed that the proposed control strategy accurately recognized the patient's gait intentions before the swing phase transition and generated customized assistance torque accordingly. In the 6 min WTs, it was observed that even though gait speed and travel distance increased, the metabolic cost (VO_2, heart rate) remained similar or decreased. This result is consistent with the user survey result indicating that the participant felt more comfort and stability when walking with the developed KAEGO compared to walking without it.

2. Materials

2.1. KAEGO System Structure

Figure 1 shows the configuration of the KAEGO system developed for gait training in a hemiplegic patient. Apart from the knee actuation module, it has the same configuration as a conventional knee ankle foot orthosis (KAFO) and includes an adjustable length feature to fit various patients. The knee actuation module is designed as an embedded system that incorporates an integrated control module with a motor, reducer, sensors, and communication components. It assists with the flexion and extension movements of the knee joint in the swing phase and also supports weight bearing in the stance phase.



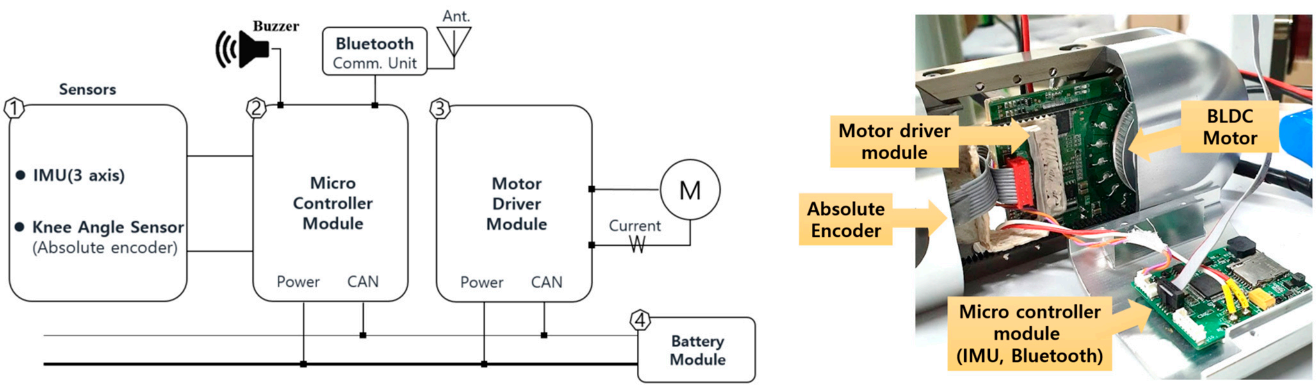
(**a**) (**b**)

Figure 1. The system configuration of the developed KAEGO: (**a**) Gait training for a hemiplegic patient wearing the developed KAEGO; (**b**) The system structure of the KAEGO. It is similar to the KAFO structure except for the knee actuator module.

The waist strap plays a crucial role in ensuring that the orthosis fits closely to the body of a hemiplegic patient when worn, preventing it from shaking during walking. Additionally, a battery (29.5 v/3.5 Ah, 400 g) module is attached to the waist strap for weight distribution. The weight of the knee actuator module is approximately 1.13 kg, and the total weight including adjustable exoskeletal brace and battery pack is approximately 3.8 kg. The rated output of the knee actuator was selected to be approximately 20 Nm based on the final stage of the reducer (51:1), as verified based on usability in a previous study [24]. Unlike the conventional serial structure, the driving part is arranged in parallel with the motor and reducer to fit as closely to the body as possible, minimizing the weight felt by the patient.

2.2. Development of Embedded Control System

The system configuration diagram and the actual product of the control system developed in this study are shown in Figure 2. As shown in Figure 2a, the developed control system utilizes only a knee angle sensor and a 3-axis IMU sensor to recognize the user's gait intentions. These components are integrated within the drive module, eliminating the need for external wiring, which reduces the risk of failure and simplifies repairs.



(**a**) (**b**)

Figure 2. The system configuration of the developed embedded control system: (**a**) The control system configuration diagram; (**b**) The manufactured knee actuator module and the developed embedded controller system including the sensor and communication module.

The control system primarily consists of a microcontroller (STM32F405), sensor module (EBIMU and absolute encoder), Bluetooth module (115,200 bps, 8 bit), and a motor driver module (maxon EC) interconnected via a CAN bus to facilitate potential future expansions. The microcontroller, IMU, and Bluetooth module are designed to operate at 100 Hz.

The Bluetooth module is included to monitor the system's operational status and enable remote control. Additionally, the operational state of the gait orthosis can be audibly indicated by a buzzer sound. This feature allows patients to be aware of the orthosis's operation status, complementing the visual feedback.

3. Methods

In a previous study on this research topic, the walking phase was classified into four states: swing flexion (SWF), swing extension (SWE), stance flexion (STF), and stance extension (STE). The user's gait intention was recognized by utilizing the GRF sensor values from both sides and motion information acquired from the IMU [24].

The flexion torque of the hip joint during walking naturally induces rotation in the knee joint [23]. However, in hemiplegic patients with brain damage, spasticity in the knee muscles makes it hard for the joint to move naturally [1]. Accordingly, this study defined an additional pre-swing state to apply appropriate torque in advance, enabling natural flexion of the knee joint during the swing phase. Particularly, the time difference from the pre-swing phase to the beginning of knee flexion in the SWF state was used as a performance indicator to evaluate how quickly the proposed gait control strategy recognizes the user's gait intention.

Figure 3 shows the finite state machine developed in this study for determining the control state. The initial state is stance extension (STE), a stance phase where the orthosis's knee is fully extended. In this state, the knee joint maintains a high impedance state to prevent the patient from falling due to sudden flexion [24]. When gait intention is detected, the system transitions to the pre-swing state and generates a predefined minimum flexion torque according to the individual patient's pathological condition, supporting natural bending of the knee joint during the swing phase due to hip movement.

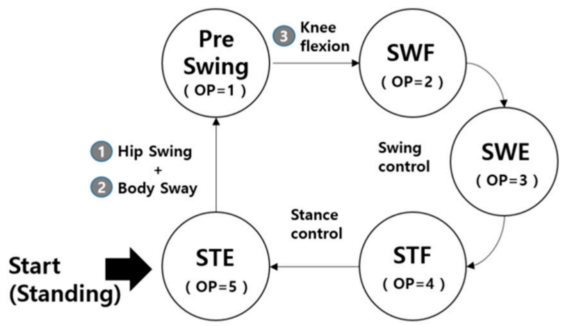


Figure 3. The finite state machine for the KAEGO gait control.

3.1. Gait Intention Detection Strategy

In this study, the progression of a single gait cycle is categorized into three phases: hip extension, weight shift, and knee flexion.

Especially, as shown in Figure 4a, the actuator module is aligned with the thigh, allowing the IMU to capture not only the thigh's rotational movement in the sagittal plane but also the rotational movement caused by weight shifts in the coronal plane.

During the hip extension phase, when the normal leg steps forward, the thigh on the affected side naturally rotates backward. This displacement corresponds to the roll displacement of the IMU connected to the thigh. In other words, the rotation of the IMU's roll axis includes the user's intention to move forward.

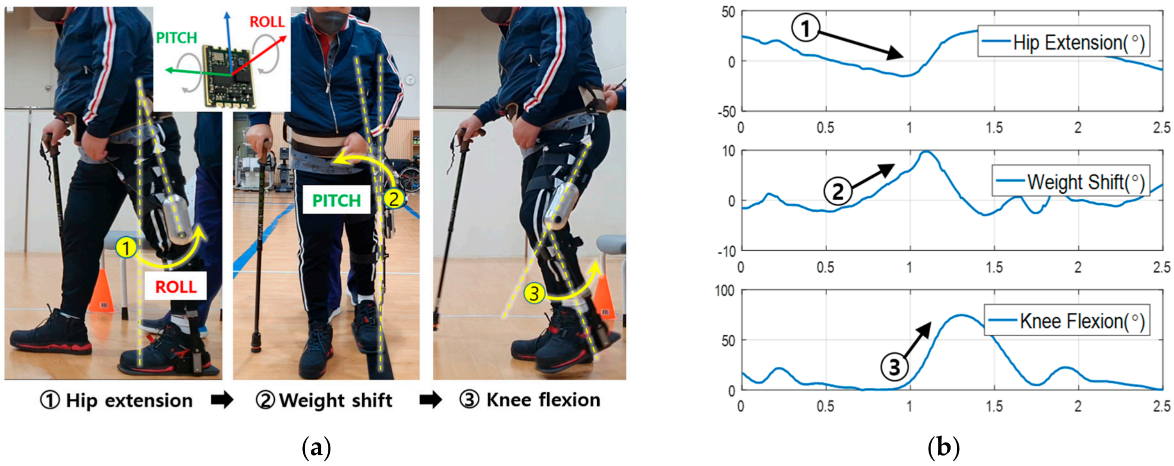


Figure 4. Motion categorization into three phase during a single gait cycle: (**a**) Stepwise motion changes in a gait cycle; (**b**) The angular displacement of the IMU roll, pitch, and knee angle linked to the progression of a single gait cycle, such as hip extension, weight shift, and knee flexion, respectively.

In the weight shift phase, the center of gravity moves either to the left or right side. This movement is accompanied by a rotation in the pitch direction of the IMU fixed to the thigh, meaning that the rotation of the IMU's pitch axis contains information about the patient's weight shift.

The final stage, knee flexion, is when the bending of the knee joint naturally begins due to the flexion torque of the hip joint. This stage signifies the transition from the stance phase to the swing phase. In this study, these three continuous rotational displacements are defined as the user's gait intention, and the information from the built-in IMU and knee angle sensor reflects these gait intentions.

As shown in Figure 4b, the angular displacement of the IMU roll and pitch is linked to hip extension and weight shift motions, respectively. However, there are limitations to processing based solely on angular displacement, as each displacement includes noise and a random offset. Additionally, during walking, various events, such as changes in speed or direction, can occur, and often it is necessary to stop the walking process in response to these events. Therefore, in this study, to classify only motions related to the actual gait intention, angular velocities that periodically meet a minimum threshold are extracted, and only the cumulative result of these is used as a meaningful displacement value. This approach enables the controller to detect unusual movements and respond to emergency situations. The formula below represents the relationship for calculating the cumulative angle from the angular velocity of IMU roll or pitch, where θ represents the roll or pitch angle.

$$\theta_{cum} = \theta_0 + \sum_{i=1}^{n} [\omega_i \times \Delta t \times I(\omega_i > \omega_{th})] \quad (1)$$

Here, θ_0 is the initial angle, θ_{cum} is cumulative angle, ω_i is the angular velocity at the *i*-th time interval, ω_{th} is the threshold angular velocity, Δt is the time interval (in this case, 10 ms), and the $I(\omega_i > \omega_{th})$ is indicator function, which takes the value 1 when $\omega_i > \omega_{th}$ and 0 otherwise. This function ensures that only the angular velocities exceeding the threshold contribute to the cumulative angle.

Figure 5 below shows a control block diagram that represents the transition to the pre-swing assist state, which assists the initial knee flexion after recognizing the gait intention, and finally switches to assist mode upon confirming the knee flexion. As illustrated in the control block diagram, abnormal movements can be detected in the process of calculating the cumulative angle, and during the pre-swing state, such movements are distinguished from changes in angular velocity. This approach minimizes the impact of malfunctions or misrecognitions.

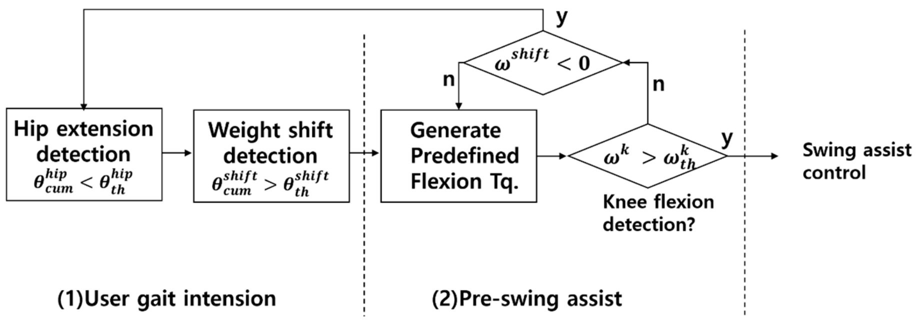


Figure 5. A control block diagram representing the process of user gait intention recognition and assist mode transitions.

3.2. Negative Damping Based Swing Assist Control Strategy

Lee developed a hip exoskeleton to assist human hip movements by simplifying the human leg into an inverted pendulum model and reducing the impact of the human musculoskeletal system on hip movements to energy losses due to viscous friction and the effects of gravity. To compensate for this, Lee proposed a negative damping control strategy [13].

Therefore, the control input for an exoskeleton robot designed for gait assistance, as stated in Equation (2), consists of a negative damping torque τ_{ND} to compensate for human viscous friction, a gravity compensation torque τ_{gc}^{H} for the equivalent mass of the human leg, a torque to counteract the nonlinear friction τ_{fc} of the exoskeleton robot system, and a gravity compensation torque τ_{gc} to offset the weight of the exoskeleton robot. This is mathematically expressed as follows [13].

$$\tau_{out} = \tau_{\mathrm{ND}} + \tau_{gc}^{H} + \tau_{gc} + \tau_{fc} \tag{2}$$

In this study, the control strategy of the developed KAEGO system aims to facilitate the natural flexion of the knee joint in response to the movement of the hip joint. Additionally, since the knee joint flexes naturally due to the rotational force of the hip joint and the inertial force caused by the weight of the lower limb, applying a torque τ_{gc}^{H} to compensate for the weight of the lower limb could lead to excessive flexion of the knee joint. Therefore, in this study, the effect of the lower limb's weight is not compensated for. Consequently, the final control input for gait assistance in the KAEGO system is as follows.

$$\tau_{out} = \tau_{\mathrm{ND}} + \tau_{gc} + \tau_{fc} \tag{3}$$

$$\tau_{\mathrm{ND}} = b\omega_{k} \tag{4}$$

Here, τ_{ND} is calculated as the product of the knee joint's negative damping coefficient b and the rotational angular velocity ω_k of the knee joint. The coefficient b represents the intrinsic viscous friction coefficient of the human lower limb, but it is impossible to estimate accurately. Moreover, the gait of hemiplegia patients not only varies greatly from individual to individual but also changes according to their condition, making it appropriate to adjust experimentally according to the state of each patient.

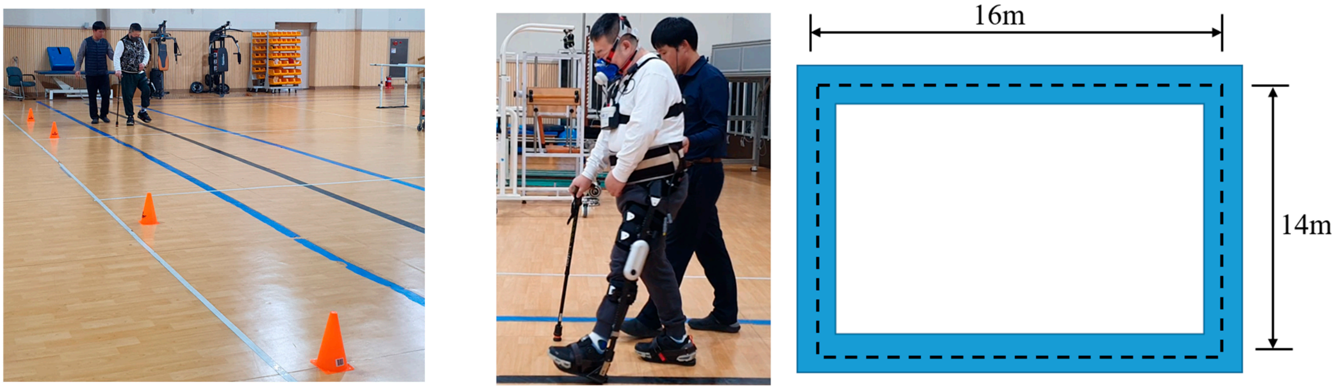
The application of negative damping torque and gravity compensation torque can vary depending on the walking state. For instance, during the swing flexion state, where the movement is opposite to gravity, gravity compensation torque is needed. Conversely, in the swing extension state, where the movement is in the same direction as gravity, gravity compensation is not applied.

The magnitude of the negative damping torque is determined based on the feedback of the knee flexion angular velocity ω^{k}, offering the advantage of responding to real-time changes in speed during walking. However, if set too high, it could lead to an abnormally large knee flexion, so its effect is limited to within a set angular range.

Details on gravity and friction compensation methods are the same as those noted in reference [13] and thus are not described in detail in this paper.

3.3. Experimental Protocol

To verify the effectiveness of the developed KAEGO system and its motion-based gait recognition and control method, a 10 m walk test (WT) [29] and a 6 min WT [31] were conducted as shown in Figure 6.



(a) (b)

Figure 6. Experimental setup for 10 m and 6 min walking tests: (**a**) 10 m walking test environment; (**b**) 6 min WT and the 60 m track.

The 10 m walking test was performed four times each on a hemiplegia patient and one able-bodied person to validate the accuracy and reproducibility of the proposed motion-based gait intention recognition method. During the 10 m walk, the number of swings and the moving speed were measured, and the reliability was verified by comparing the variances between each data set. Additionally, the time difference between pre-swing and swing flexion in each experiment was measured to determine how quickly the proposed motion-based gait intention recognition method identified the gait intention. Comparisons with data from an able-bodied individual were made to confirm the applicability of the proposed gait intention recognition method.

The 6-min walking test was conducted on one person with hemiplegia, comparing oxygen consumption (VO_2) and distance traveled when walking with and without the developed gait-assist device to calculate the metabolic cost of transport (COT) [30]. This test verified the extent to which the developed gait-assist device contributed to the patient's comfortable and stable walking.

All tests were conducted after ensuring that a patient with hemiplegia had sufficient rest and acclimatization. These tests were performed in accordance with the safety guidelines and approved by the Rehabilitation Engineering Research Institute's Institutional Review Board (RERI-IRB-221102). Additionally, prior to wearing the developed KAEGO, it was recommended that the participant use a bicycle-type lower limb exerciser to reduce muscle tension and enhance flexibility in the knee joint. From a functional perspective, the KAEGO system and existing rehabilitation robots or simple devices are complementary.

Table 1 shows the characteristics of the patient's disability. The patient suffering from left-sided hemiplegia due to a traumatic brain injury has a Berg Balance Scale (BBS) of 38. He can walk independently with a cane but with caution due to the risk of falling. To adapt to the developed assistive device, the patient engaged in familiarization training for 60 min per session, three times a week, for about four weeks.

The equipment used to calculate the COT was K4b2 (CosMed, Rome, Italy). Data were gathered while walking at a comfortable speed on a rectangular track for 6 min under the guidance of a professional therapist, as shown in Figure 6b.

Table 1. Characteristics of the hemiplegic patient.

Gen.	Age	Hei.	Wei.	Injury	BBS	Onset
M	54	170 cm	94 kg	Left Hemi.	38	2019

4. Experimental Results

4.1. 10 m Walking Test

Figures 7 and 8 show the real-time gait results and control states from the first experiment of the 10-m walking test for a hemiplegic patient and an able-bodied individual, respectively. In each chart, the vertical lines indicate when the proposed gait controller recognized the user's gait intention and switched from the control state to the pre-swing state, marking the start of the swing phase. According to the results of Figures 7 and 8, 12 swings occurred in the 10 m walk of the person with hemiplegia, and 7 swings were noted for the able-bodied individual with the controller generating assistive force (negative damping torque, ND) at the correct moments to aid knee joint actuation.

The negative damping coefficient b applied to the gait assistance of the person with hemiplegia was set at 0.11 and 0.28 mNms/deg for the SWF and SWE states, respectively. As shown in Figure 8, for the able-bodied individual who did not require assistive force, the output of the negative damping torque was set to zero, but only 650 mA pre-flexion torque was applied.

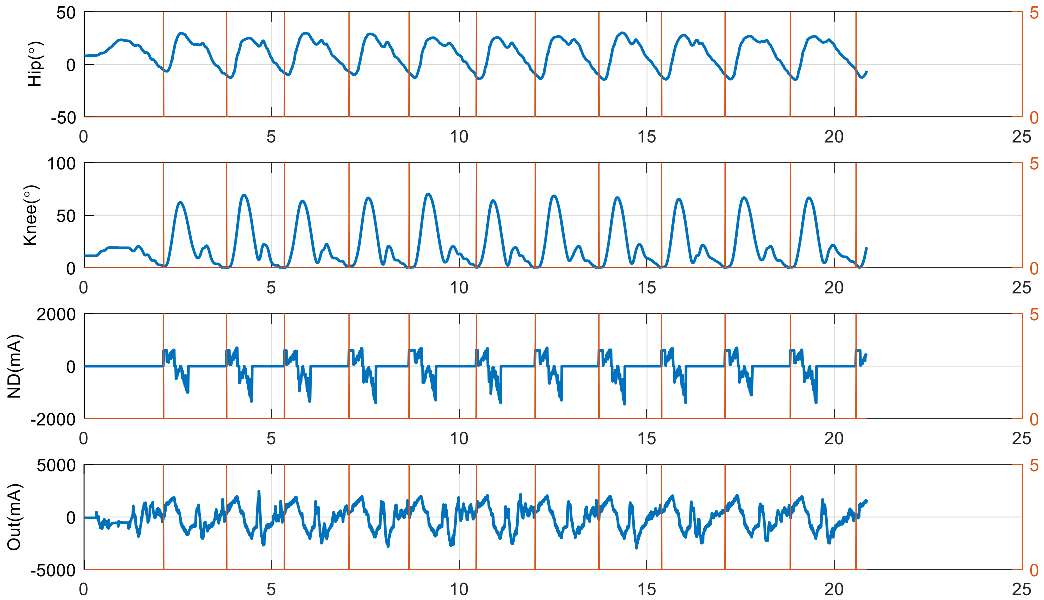


Figure 7. This illustrates the controller status during a hemiplegic patient's 10 m WT. The graphs sequentially illustrate the hip angle, knee angle, negative damping torque, and total output torque (mA). The negative damping coefficients are set at 0.11 and 0.28 mNms/deg during the SWF and SWE states, respectively. Total output torque includes friction compensation and gravity compensation torque. The vertical lines in the graphs mark the transition to the 'pre-swing' state from the stance phase as the proposed controller recognizes the user's gait intention.

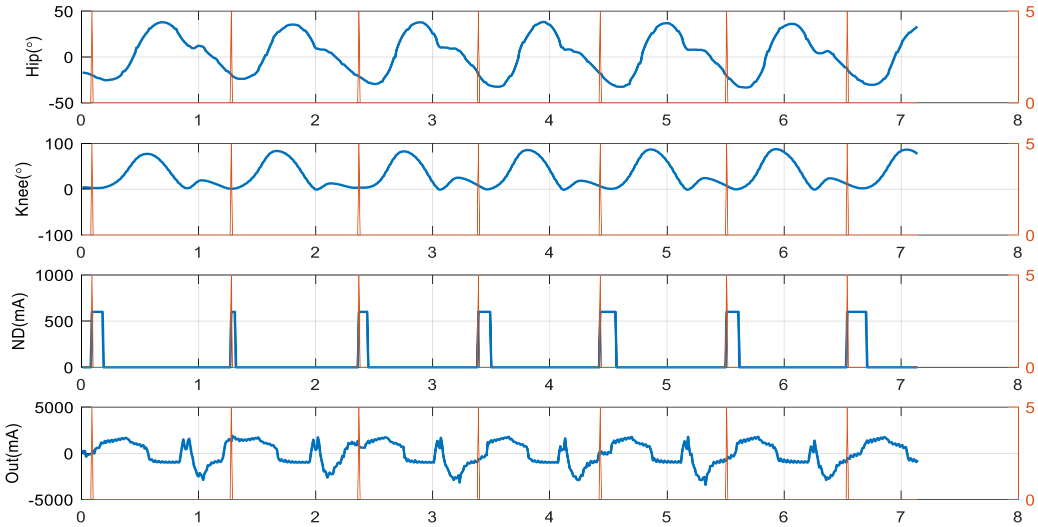


Figure 8. This illustrates the controller status during a normal person's 10 m WT. The graphs sequentially illustrate the hip angle, knee angle, negative damping torque, and total output torque (mA). The negative damping coefficients are set to 0 as no power assistance is needed for a normal person. The graph only displays the pre-swing torque. Total output torque includes friction compensation and gravity compensation torque. The vertical lines in the graphs mark the transition to the 'pre-swing' state from the stance phase as the proposed controller recognizes the user's gait intention.

The 10 m walking results for the person with hemiplegia and the able-bodied individual are presented in Tables 2 and 3, respectively. Each experiment was conducted four times. Upon careful review of all experimental data, no instances of failure to recognize or misinterpretation of gait intentions were found. The results indicate that the proposed gait intention recognition algorithm quickly recognized the user's gait intention on average 100 ms before the start of knee flexion. This not only applies to patients with hemiplegia, as the system also accurately recognized the gait intentions of able-bodied individuals with faster walking speeds. These results suggest that the developed gait recognition and control method could be universally applied to patients with hemiplegia exhibiting various walking patterns.

Table 2. Results from a hemiplegic patient's 10 m WT.

Trial		**1**	**2**	**3**	**4**	**Mean ± SD**
Swing Cnt.		12	10	10	10	10.5 ± 1.0
Velocity (m/s)		0.27	0.27	0.30	0.29	0.28 ± 0.01
Pre swing delay (ms)	Min	50	20	20	50	
	Max	160	190	230	140	
	Mean	105	96	114	94	102.3 ± 9.2

Table 3. Results from a normal person's 10 m WT.

Trial		**1**	**2**	**3**	**4**	**Mean ± SD**
Swing Cnt.		7	6	7	7	6.75 ± 0.5
Velocity (m/s)		0.72	0.79	0.72	0.69	0.73 ± 0.04
Pre swing delay (ms)	Min	40	60	20	40	
	Max	170	140	210	170	
	Mean	107.1	110	125.7	101.4	111.1 ± 10.4

4.2. 6 MWT and COT

To quantitatively assess how much the developed KAEGO system aids in comfortable and stable walking for hemiplegic patients, a 6 min walking test was conducted, and the COT was calculated to compare the metabolic energy expenditure during walking. The experiment was conducted twice, considering the patient's condition, and the COT was measured before and after wearing the assistive device for each trial.

Tables 4 and 5 present the results of the first and second 6 min walking test, respectively. The results from both the first and second trials showed a reduction in metabolic energy of approximately 7–8% when the assistive device was worn. The higher COT in the first trial is attributed to the patient's decreased condition.

Table 4. The first trial of the 6 min walking test and the results of the metabolic cost of transport (COT).

Trial	COT	Velocity (m/s)	VO_2 (mL/min/kg)	HR (bpm)	Distance Traveled (m)
without KAEGO	11.5	0.39	12.0	129.4	153.2
with KAEGO	10.5	0.44	12.3	132.4	154.1
% of difference	−8.70	12.82	2.50	2.32	0.59

Table 5. The second trial of the 6 min walking test and the results of the metabolic cost of transport (COT).

Trial	COT	Velocity (m/s)	VO_2 (mL/min/kg)	HR (bpm)	Distance Traveled (m)
without KAEGO	8.8	0.42	10.9	128.0	154.5
with KAEGO	8.1	0.46	9.8	127.2	167.5
% of difference	−7.95	9.52	−10.09	−0.62	8.41

When wearing the developed KAEGO, walking speed increased by approximately 10% in both trials. In contrast, VO_2 and heart rate (HR) increased by about 2% in the first trial but decreased in the second, particularly VO_2, which showed a reduction of great than 10%. Therefore, the test results indicate that patients with hemiplegia can travel farther distances with a similar level of metabolic energy expenditure when wearing the KAEGO. In other words, assuming that the same distance is traveled, the findings indicate that movement occurs with lower metabolic energy consumption.

5. Discussion

In this study, we developed a unilateral knee-actuated exoskeletal gait orthosis (KAEGO) for gait training in patients with hemiplegia. The KAEGO system facilitates rehabilitation progress through interaction between the orthosis and the patient in over-ground environments, operating in accordance with the gait intentions of the hemiplegic patient.

To achieve this, this study proposes a simple gait control strategy using an IMU sensor and a knee angle sensor. This approach detects gait phases and recognizes the patient's gait intentions, eliminating the need for GRF sensors. Additionally, we implemented a simple torque generation method based on a negative damping model to assist knee joint movements. Two types of walk tests were conducted to evaluate the effectiveness of the proposed control strategy, focusing on the accuracy of gait recognition and the comfort of gait assistance.

The first experiment involved a hemiplegic patient and a healthy individual performing 10 m walk tests. Data collected during these tests were used to analyze gait recognition accuracy and how quickly the gait intention were identified. In the second experiment, we hypothesized that the KAEGO, by providing appropriate assistive torque, would reduce the metabolic cost despite potential discomfort from the device's weight (3.4 kg) and wearability issues. To test this hypothesis, we measured the metabolic cost of transport (COT) during a six-minute walk test.

Results from the 10 m WT showed that the gait intentions were accurately recognized without errors, and transitions could be rapidly detected approximately 100 ms before the start of the swing phase. In the six-minute WT, results indicated an approximate 8% reduction in the metabolic cost of transport when wearing the developed KAEGO.

However, this study has some limitations. The sample size was small and focused on specific types of gait impairments. Future studies will include a more diverse group of participants to validate the KAEGO's effectiveness across a broader spectrum of hemiplegic patients. Additionally, long-term studies are necessary to assess the device's safety and durability, especially in varied and challenging real-world environments. Moreover, for further clinical effect analysis, we plan to use traditional camera-based motion analysis techniques to quantitatively compare the effects of interventions before and after gait training with the orthosis and to focus on assessing improvements in gait balance.

For the practical implementation of the KAEGO for wide use in everyday life, it is crucial that the device is easy for patients or caregivers to wear. It should offer simple adjustability of assistive force and robust durability for long-term use. Furthermore, reducing and redistributing the device's weight is essential to minimize its impact on patients. Ensuring the device's reliability and safety to prevent malfunctions is also critical.

Author Contributions: Conceptualization, Y.H. and H.-J.C.; methodology, Y.H. and H.-S.C.; software, Y.H.; validation, Y.H. and H.-S.C.; investigation, H.-J.C.; data curation, H.-J.C. and J.-W.L.; writing—original draft preparation, Y.H. and H.-J.C.; writing—review and editing, H.-J.C.; visualization, J.-W.L.; supervision, H.-S.C.; project administration, G.-S.K. All authors have read and agreed to the published version of the manuscript.

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Institutional Review Board Statement: The study was conducted according to the guidelines of the Declaration of Helsinki, and approved by the Institutional Review Board of the Rehabilitation Engineering Research Institute, Korea Workers' Compensation & Welfare Service (RERI-IRB-221102).

Informed Consent Statement: Written informed consent has been obtained from the patient to publish this paper.

Data Availability Statement: The data presented in this study are available on request from the corresponding author.

Conflicts of Interest: The authors declare no conflicts of interest.

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