Comparative Study on Overground Gait of Stroke Survivors With a Conventional Cane and a Haptic Cane

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***Abstract*— The conventional cane (single cane) is widely used to promote gait ability of stroke survivors as it provides postural stability by extending the base of support. However, its use can reduce muscle activity in the user's paretic side and cause upper limb neuropathies due to the intermittent and excessive loading of the upper limb. The provision of low magnitude support and speed regulation may result in collective improvement of gait parameters such as symmetry, balance and muscle activation. In this paper, we developed a robotic Haptic Cane (HC) that is composed of a tilted structure with motorized wheels and sensors to allow continuous haptic contact with the ground while moving at a regulated speed, and carried out gait experiments to compare the HC with an Instrumented conventional Cane (IC). The results show that use of the HC involved more continuous ground support force of a comparatively lesser magnitude than the IC, and resulted in greater improvements in the swing symmetry ratio and significant improvements in the step length symmetry ratio. Percentage of Non-Paretic Activity (%NPA) of paretic muscles (vastus medialis obliquus (VMO), semitendinosus (SMT), tibialis anterior (TBA) and gastrocnemius medialis (GCM)) in swing phase was significantly improved by the use of either device at fast speed. However, the use of HC improved %NPA of paretic VMO and SMT more than the use of IC at both preferred and fast speeds. It also significantly improved %NPA of paretic GCM in stance phase. Furthermore, comfortable speed with the HC was higher than with the IC and exhibited better RMS of anteroposterior (AP) tilt. Thus, the developed device with a simple and intuitive mechanism can provide efficient assistance for overground gait of stroke patients with a high possibility of widespread use.**

***Index Terms*— Rehabilitation robotics, haptics, gait assistance, cane, stroke patients.**

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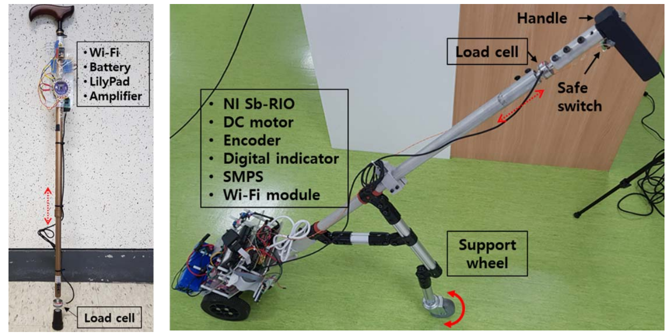
I. Introduction

STROKE typically leads to hemiplegia, and even 3 months after stroke, approximately 70% of the patients walk with reduced velocity and capacity [1]. Furthermore, hemiplegia increases metabolic cost of transport [2]–[4], causes asymmetric gait, and reduces walking speed [5], [6]. Stroke survivors with balance control or sensorimotor deficiencies tend to exhibit problems associated with gait, including increased likelihood of falling and loss of independence [7], [8]. Thus, gait rehabilitation is vital for improving post-stroke quality of life and performance of activities of daily living [7], [9]. Since the causes and symptoms of gait disorders experienced by stroke survivors are diverse, easily measurable gait parameters such as gait speed and symmetry, which are closely related to balance and gait ability, are considered as appropriate evaluation indices [5], [10], [11]. Studies have shown that improvement of step length symmetry coincides with improvements in gait speed [6] and cost of transport [5], [12], [13].

The conventional cane (single cane) is widely used as a therapeutic or assistive device to promote patient's gait ability. It provides postural stability by extending the base of support, while allowing voluntary gait with an increased psychological sense of stability [14]. Although use of the conventional cane reduces body sway in the Mediolateral (ML) and Anteroposterior (AP) directions, it causes the COP (center of pressure) of the user to shift forward and laterally towards the cane [15]. In addition, conventional canes are generally used to support the affected lower limb through the unaffected side, which results in the provision of only intermittent support as the user has to lift and move the cane forward repeatedly [16]. This can cause excessive loading of the upper extremity, which has been related to the occurrence of entrapment neuropathies such as carpal tunnel syndrome on the non-paretic side in sub-acute and chronic stroke patients [1], [17]. Moreover, conventional

TABLE I
COMPONENTS AND SPECIFICATIONS OF THE IC

Specification	Value
Total Weight	510 g
Height	77 ~ 99.5 cm
Microcontroller	LilyPad Arduino Board DEV-10274
Load cell	CDFS (Capacity: 100 kg, Bongshin Loadcell Co.)
Load Cell Amplifier	XFW-HX711
Battery	Li-Polymer 3.7V, 2000mAh
Wi-Fi Module	ESP8266



(a) (b)

Fig. 1. The (a) instrumented conventional cane (IC) and (b) the robotic haptic cane (HC).

cane use can cause reduction in the muscle activity of the paretic side [18], and higher energy expenditure [19].

Boonsinsukh *et al.* reported that provision of light touch contact through an instrumented cane improved ML trunk stability and paretic leg muscle activity of stroke survivors [20], [21]. Furthermore, pilot testing of the earlier version of the haptic cane (light-grip device) with a stroke patient showed promising results in terms of gait speed and muscle activations in the paretic leg [22]. Therefore, there is a need for a gait-training device that is able to encourage a hemiplegic patient to use the affected lower limb more actively and increase their gait speed while providing continuous balance support during overground gait. Some robotic systems have been suggested for overground walking assistance [23], [24]. However, these walker/cane type devices usually require the users to bend their torso forward to transfer their weight to the device. This limits arm movement during walking, which is important for stabilization of the rotational motion of the body that occurs during locomotion [25]. Qingyang Y. et al reported a robotic cane capable of providing stable vertical support [26]. However, due to its wide base, it requires the user to extend their arm to a relatively large extent and can cause hindrance during walking.

In this paper we present the development of a robotic Haptic Cane (HC) for hemiplegic patients that induces fast overground walking with higher stability, and evaluate its effects on the gait, balance and muscle activation parameters of stroke survivors. The HC is designed to be economical and compact, in order to provide robotic benefits to the therapist such as sensor-based training with overground walking. Thus, the developed robotic device may replace the conventional cane for gait training of sub-acute and chronic stroke patients. In order to compare the effectiveness of the developed device with that of a conventional cane, we have also developed an Instrumented Conventional Cane (IC) and performed experiments with stroke survivors to compare the two devices.

In the presented work, stroke survivors walked overground with each device at preferred and fast walking speeds (speeds determined during IC walking). The data recorded in these experiments are used to compare the ground support forces applied by the users on the two devices, their spatio-temporal gait parameters and their electromyographic (EMG) data of muscle activity, in order to determine the benefits that the developed system may have over the commonly used conventional cane. Additionally, the users' preferred speed while walking with the HC is determined and compared with their preferred walking speed with the IC. This work was done to evaluate four hypotheses: First, walking with the HC involves the generation of continuous but lesser supporting force at the hand as compared to the intermittent force generated during walking with an IC. Second, walking with the HC can produce greater improvement in the gait parameters (gait symmetry ratio) than walking with the conventional cane due to the continuous provision of lesser magnitude ground support force. Third, use of the HC can result in a higher preferred speed with stable balance than that observed with the IC. Fourth, use of the HC can cause the muscles on the paretic side to be more active than with the use of the conventional cane. To evaluate these hypotheses, we carried out the presented study with two objectives, first was to compare the IC and HC at the IC preferred and as fast as possible speeds in order to compare the effects of the two devices, and second was to determine the comfortable speed with the HC and to compare it with the IC preferred speed.

II. SYSTEMS & METHODS

A. Instrumented Conventional Cane (IC)

The IC used in this research is shown in Fig. 1 (a). In order to keep the weight and feel of the IC close to that of a conventionally used cane, it was built by adding a force sensor (load cell) to a conventional cane made out of lightweight and rigid aluminum alloy tubing. The entire apparatus, consisting of the conventional cane, load cell and allied electronics weighs 510g, and its length can be adjusted between 77cm and 99.5cm. The load cell (100kgf CDFS, Bongshin Loadcell Co, Republic of Korea) attached to the tip of the cane converts the force applied by the user to an electrical signal that is amplified and digitized by the load cell amplifier (XFW-HX711, Sparkfun, USA). Only the compressive load acting on the load cell is of interest in this study. Therefore, the load cell is attached at the bottom of the cane so that the cane's own weight also causes compression of the load cell, which is corrected by zeroing the cane at the start of the experiment. This method of attaching the load cell to the tip of the IC is similar to that reported in previous studies related to instrumented supportive devices [27]–[30]. However, the IC that we have used here is much lighter than that which was reported previously (1kg) [27]. The microcontroller (LilyPad Arduino Board DEV-10274, Sparkfun, USA) processes the amplifier output to obtain the actual value of the applied load. This value is then wirelessly transmitted to the PC via the WIFI module (ESP8266, Espressif Systems, China). The device is powered by an onboard 3.7V, 2000mAh Li-Polymer battery through a 3.3V voltage regulator (S7V8F3, Pololu, USA). At the PC, a custom developed software running in the LabVIEW environment (National Instruments, USA) receives the load data and records it at a rate of 50Hz. The IC software is setup to calibrate the load value to zero at the time of startup so that the recorded values represent only the load applied by the user.

B. Robotic Haptic Cane (HC)

The HC, shown in Fig. 1 (b), is a modified version of the previously developed simple and intuitive robotic rehabilitation device for partially ambulant stroke patients [22]. In the presented work, in order to make the previously developed device a free-standing system and to add the capability of providing ground support, we have modified its structure by adding a freely rotating supporting wheel. This modification has been made while ensuring that it does not interfere with the free movement of the user's legs [26]. With this modification, the system can help to improve walking stability by providing continuous contact with the ground to generate kinesthetic haptic feedback while moving with a regulated speed in the horizontal direction. Addition of the third wheel also means that the wheels support the entire weight of the cane and no load due to it is applied to the user's hand. We expect that this continuous transfer of ground conditions to the user's hand can provide somatosensory augmentation [20]. Additionally, kinesthetic haptic feedback with regulated gait speed should encourage the user to increase the employment of their affected lower limb in supporting their body weight rather than putting more load on their healthy upper limb.

As shown in Fig. 1 (b), the HC frame tilted at 45° is constructed using rigid aluminum tubing (diameter, 25 mm) with a thin aluminum plate at its base used to attach the electronic components. The supporting caster wheel assembly is attached to the system structure to provide rigid ground support similar to a conventional cane when the patient cannot stand without external support. The cane's drive motor transfers power through a bevel gear mechanism to the rigid rubber driven wheels that were selected to minimize deformation under load. All hardware components, with the exception of the force sensor, are located as close to the wheels as possible to keep the system's center of gravity as low as possible. The two wheeled solid axle design for the driven wheels helps to maintain straight walking while the freely rotating support wheel allows easy turning of the system according to user applied forces. Total weight of the system is 9.85 kg and it can support a maximum load of 18 *kgf* applied at the handle, which is adequate to provide support to stroke subjects during gait [31].

Fig. 3 shows the system block diagram of the HC while Table II presents its components and specifications. A Sb-RIO (National Instruments, USA) is used for hardware interface and control. The motor control is implemented at the FPGA level of the Sb-RIO with a feedback loop frequency of 1 kHz. The device connects to a PC via a WIFI interface. At the PC, software running in the LabVIEW environment (National Instruments, USA) is used by the operator to control the device's speed and to record operational data. To guarantee safe interaction between the user and the device, a switch is incorporated into the handle of the system. When the user grips the handle, the system speed controller is activated, and when the switch is released, a brake function is activated by setting the controller's speed command to zero. For added safety, the motor operation can also be controlled by the operator through the PC software. The system can be operated for more than 3 hours with a full charge of its battery, which is sufficient for several sessions of gait training. The HC is equipped with a force sensor mounted near its handle to avoid any reading errors due to the inertia of the system. The force measured by this sensor is communicated by its digital indicator to the Sb-RIO at 50 Hz via a RS-485 interface. This force data represents the axial force acting along the cane structure. This axial force is a combination of the support

TABLE II
COMPONENTS AND SPECIFICATIONS OF THE HC

Specification	Value
Maximum speed	2 m/s
Weight	9.85 kg
Wheel radius	136 mm
Encoder resolution	38 pulses/rev
Motor	DC motor (48.6 W, gear ratio 19:1, 4000 rpm)
Motor driver	Escon 50/5 (Maxon DC motor controller)
System controller	Sb-RIO 9636 (National Instruments, USA)
Force sensor with digital indicator	CDES(Capacity: 100 kg), BS-205 digital indicator (Bongshin, Korea)
Battery	Lithium Ion battery (4/3FA 22.2V 13000mAh)

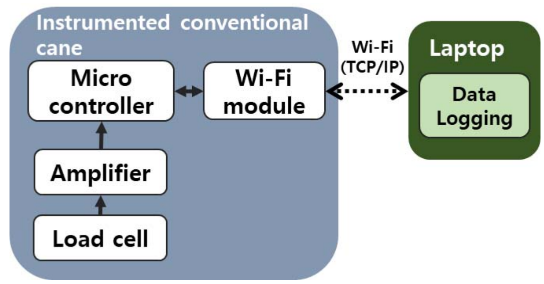


Fig. 2. Block diagram of the IC.

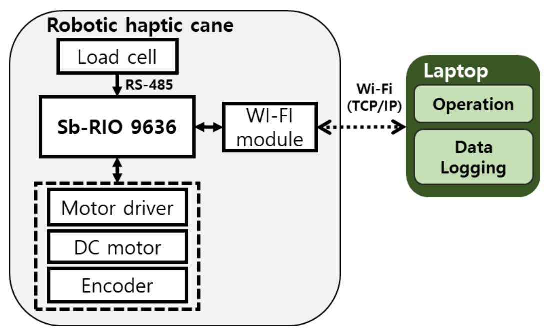


Fig. 3. Block diagram of the HC.

TABLE III
DEMOGRAPHIC DETAILS OF PARTICIPANTS

Demographic	Value
Number of subjects	10
Gender	Male=5, Female=5
Age (year)	62.0 ± 12.1
Height (cm)	161.5 ± 7.3
Weight (kg)	64.2 ± 19.0
Days since onset	45.6 ± 47.9
Modified Barthel Index Score	63.1 ± 10.0
Mini-Mental State Examination Score	24.9 ± 3.7
Cause of Stroke	Infarction = 7, Hemorrhage =3
Side of Hemiplegia	Right=6, Left=4

being provided by the cane and the propulsive force exerted by the motor. Therefore, in order to consider only the support being provided by the HC, the cosine (cos 45°) of the axial force is taken, and the values thus obtained, termed as ground support force (GSF), are recorded for subsequent data analysis.

C. Participants

Ten individuals with stroke took part in this study, which was carried out with the aim of achieving two objectives. First, we wanted to compare the effectiveness of the HC with that of the IC in assisting the subjects to improve their gait parameters, balance and muscle activation while also comparing the interaction force behavior of the two devices. The second objective was to determine and compare the user's preferred gait velocities while using each of the devices. The study participants had suffered a single onset of unilateral hemiparetic stroke and were in the sub-acute phase of recovery. All participants were able to walk more than 10 m with the conventional cane. Individuals who suffered from any additional neurological or musculoskeletal conditions were excluded from the study. Demographic details of the participants are shown in Table III. This study was conducted at the Rehabilitation Center of Gyeongsang National University Hospital (Jinju, Republic of Korea) following the principles of the Declaration of Helsinki, and was approved by the Institutional Review Board of Gyeongsang National University Hospital (2018-10-018-002). All subjects gave written informed consent for research participation and publication of their data.

D. Protocol

Prior to the start of trials, the participants were briefed about the operation of the HC and IC, and about the different sensors that they were required to wear for data collection. Training was not provided to the participants; however, they were allowed to familiarize themselves with the devices by walking a few steps (less than the length of one trial walk) with them. In each trial, the participants walked 8 m in a straight line on a flat surface with either the IC or the HC and the data obtained during the middle 6 m were analyzed. Two trials under each condition were performed with 1-minute seated rest between trials and about 3-minute seated rest between each condition. The IC and HC trials were carried out in separate blocks with the IC trials being done first as their recorded speeds were required for the HC trials that followed them after a 5-minute seated rest. The extensive breaks were provided to limit the training and fatigue effects that may arise due to the use of a non-randomized protocol. Therefore, first, the subjects walked with the IC at their preferred normal speed and then at the fastest safely possible speed which they selected by themselves. The researcher operating the HC control software calculated the average walking speeds for both these conditions. Then, he set the HC velocities according to the trial conditions and the subjects walked with the HC at the preferred and fast speeds recorded during the trials with the IC. The user's intentional speed was not used in the HC trials, as it would have led to a mismatch between the IC and HC speeds, which would prevent us from doing a fair comparison between the two devices. While performing the HC trials, if the stroke patient was unable to stably control the device operation switch, the operator remotely controlled the starting and stopping of the HC. Lastly, in order to determine the maximum speed at which the subjects can comfortably use the HC, starting from the fast walking speed with the IC, the operator increased or decreased the speed in steps of 5 – 10 % and asked the subjects to walk a few steps. Once the comfortable speed was determined, we carried out two trials of waking with the HC at this comfortable speed. None of the trials included any cognitive loading or performance of concurrent tasks. The overall trial conditions were as follows:

ICPS: IC with Preferred Speed
ICAF: IC with As Fast as possible speed
HCPS: HC with speed recorded during ICPS
HCAF: HC with speed recorded during ICAF
HCCS: HC with maximum Comfortable Speed

As shown in fig. 4, the subjects wore soft Velcro belts holding IMU (Inertial Measurement Unit) sensors for evaluation of gait parameters (MyoMOTION, NORAXON, USA), while wireless electromyography (EMG) sensors (Ultium EMG, NORAXON, USA) to monitor muscle activity were stuck directly to their skin. The IMU sensors were attached to the thigh, shank, foot and pelvis, and their motion capture data were wirelessly recorded at a rate of 100 Hz. EMG data of four muscles (vastus medialis obliquus (VMO), semitendinosus (SMT), tibialis anterior (TBA) and gastrocnemius medialis (GCM)) of the healthy and paretic legs were wirelessly recorded at a rate of 2 kHz. As an additional safety measure, a rehabilitation medicine doctor walked behind the subjects during all trials.

E. Data Analysis

This study focuses on comparing general characteristics of the support that the users take from HC and IC, and the effects of the use of these devices on the gait parameters and muscle activations of the users. In the force comparison, the GSF exerted by the HC and IC are compared in terms of their peak values and duration of application. For this purpose, the HC GSF is calculated as described in section II B and the total force measured by the IC load sensor is considered as the IC GSF. This has been done based on the observation

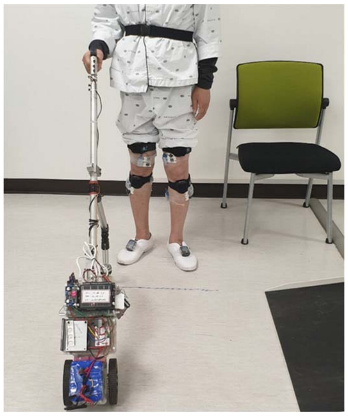


Fig. 4. A stroke patient wearing the IMU and EMG sensors with the HC.

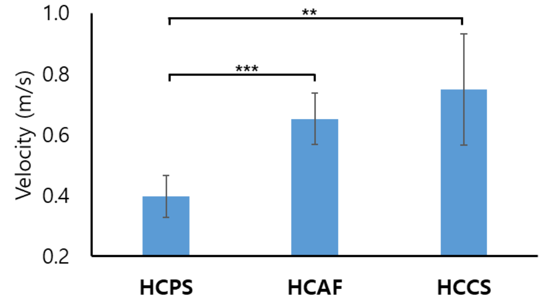


Fig. 5. Mean and SD values of gait speed of all subjects during various HC trials. Statistically significant differences are marked based on post-hoc pairwise comparisons ($**$ = P-value $<$ .01, $***$ = P-value $<$ .001). The error bars represent the SD.

that while using a conventional cane, the peak support force occurs when the cane is in the vertical orientation [32]. During processing, the IC and HC GSF values stored in the PC were normalized according to the subjects' bodyweight and the portion of data recorded during the middle 6m of the walking path was used to determine the peak and duration of GSF. The duration of application of positive force was considered as the duration of GSF, which was then converted to percentage of gait cycle. The gait speed was calculated using time recorded by a stopwatch and the average value from the two IC trials under each condition (ICPS and ICAF) was used in the corresponding HC trials.

Stride length, step length and gait phase were extracted from the myoMOTION IMU data using the MyoResearch software (MR3 3.14, NORAXON, USA). Furthermore, from the pelvis IMU data, RMS (Root Mean Square) values of Mediolateral (ML) and Anteroposterior (AP) tilts were calculated to determine the level of balance during the different gait conditions. The swing-time and step length symmetry ratios (SR) were also calculated for each condition. The SR for stroke patients is defined as the ratio between the value of the paretic leg and the value of the non-paretic leg while using the smaller value as the denominator [21]. These parameters for each trial were extracted from the MR3 report and average of the two trials was taken using a software running in the MATLAB® environment (Mathworks, USA).

The EMG signals were recorded after band pass filtering (10–500 Hz). During post-processing, the data was smoothed using an RMS filter with a 100 ms window [33]. To identify the increase in muscle activity of the paretic side, the percentage of non-paretic peak activity (%NPA) was calculated, where integrals of the EMG values of the paretic muscles during stance and swing phase were normalized to the corresponding peak EMG values of the same muscles on the non-paretic side [20], [21], [34].

A one-way repeated measures analysis of variance (RMANOVA) was done to observe the difference between the three HC gait speeds (factor, levels: HCPS, HCAF and HCCS). Paired t-tests were conducted to evaluate the difference in balance (RMS of ML and AP tilts) of the subjects between ICPS and HCCS conditions. Furthermore, a two-way RMANOVA was used to identify the effects of Device (factor, levels: IC and HC) and increase in Speed (factor, levels: PS and AF) on GSF, stride length, SR (swing time and step length), RMS of ML&AP Tilts, and %NPA of the four paretic muscles. Mauchly's test of Sphericity was used to check the validity of the RMANOVA results, and Greenhouse Geisser corrections were applied in case of its violation. Post hoc tests were conducted with application of the Bonferroni correction method. Partial eta squared (η_p^2) was calculated as a measure of the effect size for one- and two-way RMANOVA. All statistical analyses were performed using SPSS V20.0 (IBM Corp., USA).

III. Results

Table IV presents the mean and SD values of all the observed parameters and the gait speeds during all the trials.

A. Gait Speed Dependent RMS of Pelvic Tilt With HC

Results of one-way RMANOVA comparing the different gait velocities (m/s) are shown in Fig. 5. The PS and AF velocities are those that were measured during the IC trials and the HC was regulated according to those measured values, whereas the CS is the HC comfortable speed selected through user feedback. The results show that there are significant differences in the velocities (F (1.161, 10.445) = 29.195, p < .001, η_p^2 = .764) between conditions HCPS and HCAF (p < .001), and HCPS and HCCS (p <.005). As shown in Fig. 6, paired t-tests revealed statistically significant difference only in RMS of AP tilt between ICPS and HCCS (p < .001).

B. Gait Speed and Device Dependent Ground Support Force (GSF), Symmetry Ratio (SR), Stride Length, Pelvic Tilt and Muscle Activation

Fig. 7 presents the GSF of a representative subject during the different speed trials with both IC and HC. Results of two-way RMANOVA of GSF, stride length, SR, pelvic tilt (ML and AP) and muscle activation, carried out to investigate the effects of device and speed variation are presented in Table V. For peak and duration of GSF, no significant interaction led to post-hoc analysis of device and speed separately. The results of all these post-hoc tests are presented in Fig. 8. Also, for stride length, swing SR and step length SR, no significant interaction led to post-hoc analysis of Device and Speed separately. The results of these post-hoc tests are presented in Fig. 9 and 10. No significant interaction of Device

TABLE IV
DETAILS OF THE MEASURED PARAMETERS DURING VARIOUS GAIT TRIALS

Parameter	ICPS	ICAF	HCPS	HCAF	HCCS
Peak of ground support force (GSF)	0.115±0.029	0.103±0.032	0.020±0.009	0.016±0.009	0.015±0.007
Duration of GSF (%)	60.907±6.117	53.089±9.690	83.362±6.902	66.103±12.485	63.045±12.064
Speed (m/s)	0.396±0.068	0.651±0.085	0.396±0.068	0.651±0.085	0.749±0.183
Stride length (cm)	61.844±8.776	77.310±8.330	54.702±9.890	72.679±12.735	78.965±16.144
Swing Symmetry Ratio	1.111±0.020	1.086±0.025	1.073±0.024	1.048±0.020	1.041±0.014
Step Length Symmetry Ratio	1.086±0.049	1.069±0.033	1.066±0.033	1.034±0.025	1.029±0.011
RMS of ML tilt (degrees)	1.836±0.367	1.809±0.422	1.472±0.355	1.506±0.408	1.639±0.355
RMS of AP tilt (degrees)	3.850±0.926	3.419±0.793	2.766±0.657	2.959±0.623	2.658±0.798
NPA of Paretic VMO in Stance (%)	18.427±7.757	21.571±19.001	19.001±6.604	20.567±6.553	22.253±9.339
NPA of Paretic VMO in Swing (%)	5.238±4.148	8.894±4.010	6.657±4.984	12.875±4.349	13.045±5.273
NPA of Paretic SMT in Stance (%)	20.276±8.420	22.893±7.280	19.430±11.019	23.276±8.445	25.957±9.587
NPA of Paretic SMT in Swing (%)	7.410±3.794	12.993±4.669	9.660±5.627	17.049±4.863	16.722±3.849
NPA of Paretic TBA in Stance (%)	26.971±8.135	27.936±7.231	26.462±9.392	27.693±7.060	28.763±7.624
NPA of Paretic TBA in Swing (%)	10.889±2.712	15.953±2.807	11.287±2.772	15.282±2.844	17.276±3.868
NPA of Paretic GCM in Stance (%)	26.245±10.852	28.820±9.424	25.268±6.907	31.686±7.666	32.898±9.375
NPA of Paretic GCM in Swing (%)	5.794±3.937	8.442±5.029	5.902±3.615	11.281±4.183	11.585±4.304

ICPS: IC with Preferred Speed, ICAF: IC with As Fast as possible speed, HCPS: HC with same speed as ICPS, HCAF: HC with same speed as ICAF, and HCCS: HC with Comfortable Speed. Peak of GSF is normalized to subjects' bodyweight while duration of GSF is calculated as percentage of gait cycle.

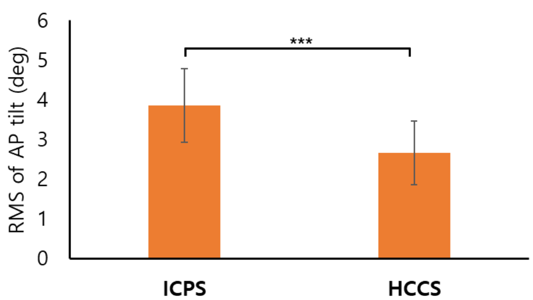
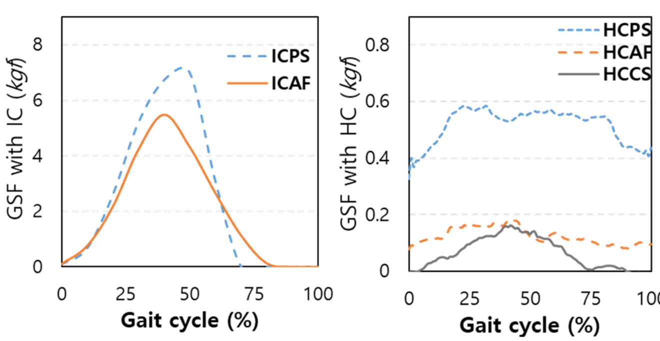
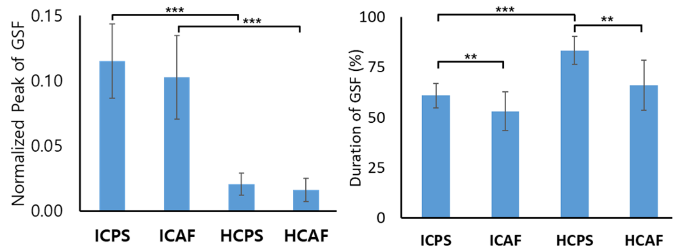


Fig. 6. Mean and SD values of RMS of AP tilt of all subjects during ICPS and HCCS trials. Statistically significant difference is marked based on paired t-test ($*** =$ P-value $<$ .001). The error bars represent the SD.



(a) (b)

Fig. 7. GSF with (a) IC and (b) HC of a representative subject. Values represent average of all gait cycles during one trial under each condition.



(a) (b)

Fig. 8. Mean and SD values of (a) peak of GSF, (b) duration of GSF of all subjects during various gait trials. GSF is normalized to the subjects' bodyweight. Duration of GSF is calculated as percentage of gait cycle. Statistically significant differences are marked based on post-hoc pairwise comparisons ($* =$ P-value $<$ .05, $** =$ P-value $<$.01, $*** =$ P-value $<$.001). The error bars represent the SD.

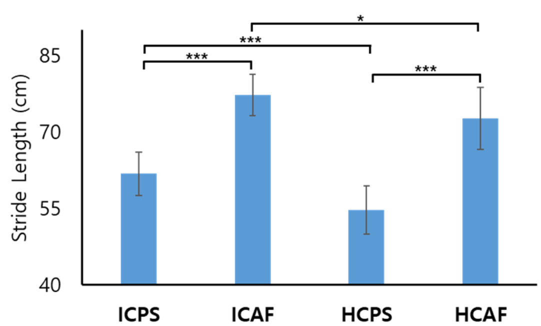


Fig. 9. Mean and SD values of stride length all subjects during various gait trials. Statistically significant differences are marked based on post-hoc pairwise comparisons ($* =$ P-value $<$.05, $** =$ P-value $<$ .01, $*** =$ P-value $<$ .001). The error bars represent the SD.

and Speed on RMS of ML and AP tilt, led to post-hoc analysis of Speed separately. The results of these post-hoc tests are presented in Fig. 11.

There was no significant interaction between Device and Speed on %NPA of Paretic VMO, TBA and SMT during swing, which led to post-hoc analysis of Device and Speed separately (Fig. 12. (a), (b) and (c)). There was also no significant interaction between Device and Speed on %NPA of Paretic GCM during both the stance and swing phases, which led to post-hoc analysis of Device and Speed separately for both values. Results of these post-hoc tests are presented in Fig. 12 (d) and (e).

IV. DISCUSSION

The objectives of the presented work comprise studying the effects of the developed system (HC) on the user's gait related outcomes and comparing the HC with the IC during overground walking. The HC has a simple and intuitive design that may provide the user with relatively lower magnitude and more continuous supporting force than the IC. This may not only improve walking stability, but also increase the paretic side muscle activity. To evaluate these hypotheses, we designed the presented experiment to allow the comparison of IC and HC in terms of ground support force (GSF), gait

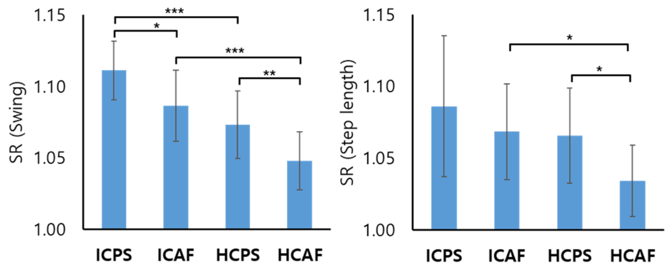


Fig. 10. Mean and SD of (a) swing and (b) step length SR of all subjects during various gait trials. Statistically significant differences are marked based on post-hoc pairwise comparisons ($*$ = P-value $<$ .05, $**$ = P-value $<$ .01, $***$ = P-value $<$ .001). The error bars represent the SD.

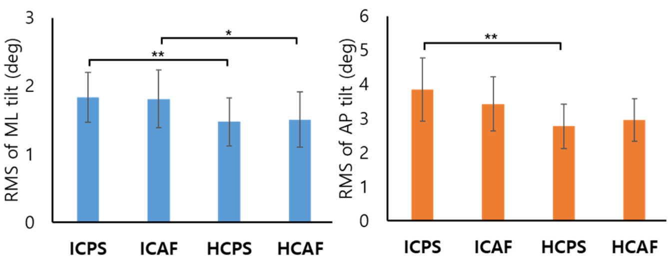


Fig. 11. Mean and SD values of RMS of (a) ML and (b) AP pelvic tilts of all subjects during various gait trials. Statistically significant differences are marked based on post-hoc pairwise comparisons ($*$ = P-value $<$ .05, $**$ = P-value $<$ .01). The error bars represent the SD.

parameters (stride length, SR and RMS of pelvic tilt), and %NPA of paretic leg muscles of stroke survivors during overground walking at preferred and fast gait velocities. Furthermore, to compare the comfortable speeds at which the subjects can utilize each device, we had them select their comfortable speed with the HC and compared that with the IC preferred speed. We also evaluated the differences in subjects' balance during walking with IC at preferred speed and with HC at comfortable speed.

A. Gait Speed Dependent RMS of Pelvic Tilt

Improving gait speed and symmetry is one of the important goals in gait rehabilitation because these measures are representative of dynamic stability [5], [10], [11]. Polese, Janaine Cunha, *et al.* reported that walking speed of stroke survivors increased with the use of a conventional cane [35]. Results of the current study also show that walking speed can be increased with the use of a conventional cane (ICPS (0.396 $\pm$ 0.068), ICAF (0.651 $\pm$ 0.085) (Table IV). However, since HCPS (same speed as ICPS) and HCCS also showed significant differences (p $<$ .005), it means that a higher range of walking speed can comfortably be achieved while using the HC, which indicates that the system has high usability for people with a wide range of gait abilities. In addition, as shown in Fig. 6, despite the higher speed in the HCCS condition, there was no significant difference in the RMS value of ML tilt, and the RMS value of AP tilt showed a significant decrease (p $<$ .001). From these observations, it can be concluded that use of the HC can result in a higher preferred speed than with the IC while maintaining stable balance. This is useful for

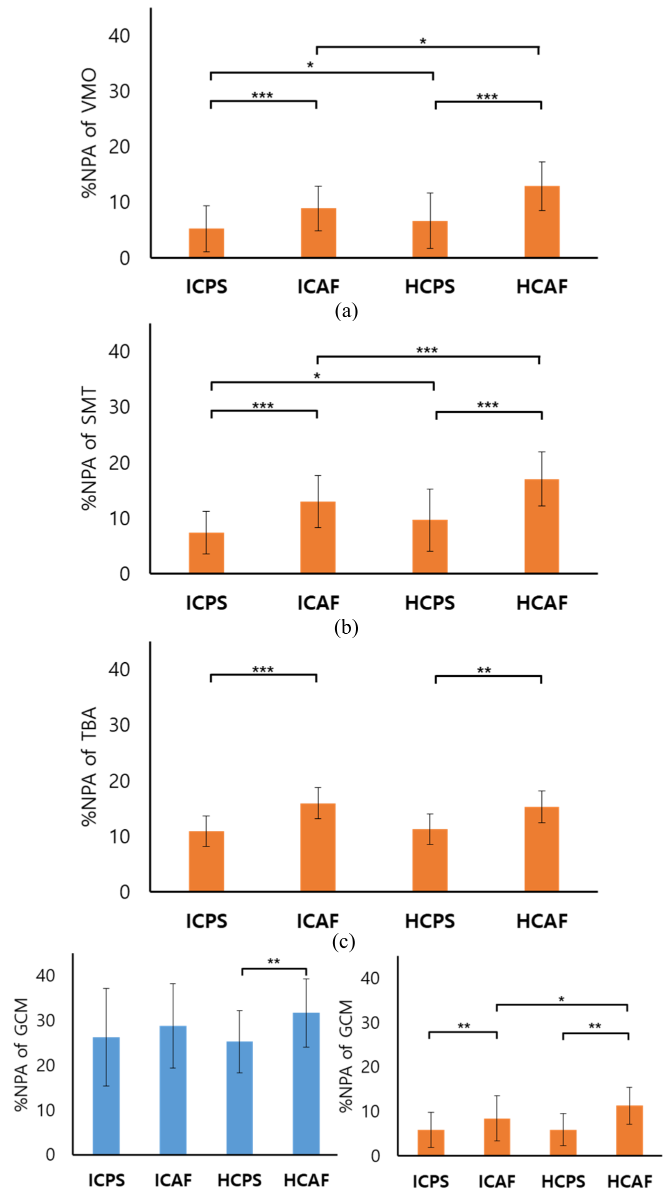


Fig. 12. Mean and SD of %NPA of paretic leg muscles. (a) VMO in swing, (b) SMT in swing, (c) TBA in swing, (d) GCM in stance and (e) GCM in swing. Statistically significant differences are marked based on post-hoc pairwise comparisons ($*$ = P-value $<$ .05, $**$ = P-value $<$.01, $***$ = P-value $<$.001)). The error bars represent the SD.

stroke survivors as increased gait speed is usually accompanied by improved balance [10].

B. Gait Speed and Device Dependent Ground Support Force (GSF), Symmetry Ratio (SR), Stride Length, Pelvic Tilt and Muscle Activation

While looking at the peak GSF values, it can be observed that at the preferred speed (PS), while using the IC, about 11.5% ($\pm$2.9) of the user's body weight was supported by the cane. Whereas, while using HC at the same speed, only 2% ($\pm$0.9) of the user's body weight was supported by the cane. Similarly, at fast speed (AF), the IC supported 10.3% ($\pm$3.2) of the body weight, while the HC supported only 1.6% ($\pm$0.9) of the body weight. The peak GSF observed here with the IC

TABLE V
RESULTS OF TWO-WAY RMANOVA

Parameter	Factor	F	P-value	η_p^2
Peak of GSF (normalized by body weight)	Device	(1,9) =78.497	**.000**	**.897**
	Speed	(1,9) =2.441	.153	.213
	Interaction	(1,9) =1.971	.194	.180
Duration of GSF (calculated as percentage of gait cycle)	Device	(1,9) =23.168	**.001**	**.720**
	Speed	(1,9) =39.412	**.000**	**.814**
	Interaction	(1,9) =3.273	.104	.267
Stride length	Device	(1,9) =26.154	**.001**	**.744**
	Speed	(1,9) =92.692	**.000**	**.911**
	Interaction	(1,9) =1.894	.202	.174
Swing Symmetry Ratio	Device	(1,9) =80.656	**.000**	**.900**
	Speed	(1,9) =16.793	**.003**	**.651**
	Interaction	(1,9) =0.005	.202	.001
Step Length Symmetry Ratio	Device	(1,9) =9.133	**.014**	**.504**
	Speed	(1,9) =7.240	**.025**	**.446**
	Interaction	(1,9) =1.096	.322	.109
%NPA of Paretic VMO in Stance	Device	(1,9) =0.021	.888	.002
	Speed	(1,9) =4.873	.055	.351
	Interaction	(1,9) =0.448	.520	.047
%NPA of Paretic VMO in Swing	Device	(1,9) =12.943	**.006**	**.590**
	Speed	(1,9) =69.185	**.000**	**.885**
	Interaction	(1,9) =5.057	.051	.360
%NPA of Paretic SMT in Stance	Device	(1,9) =0.023	.882	.003
	Speed	(1,9) =4.621	.060	.339
	Interaction	(1,9) =0.241	.635	.026
%NPA of Paretic SMT in Swing	Device	(1,9) =36.971	**.000**	**.804**
	Speed	(1,9) =297.285	**.000**	**.969**
	Interaction	(1,9) =3.863	.081	.300
%NPA of Paretic TBA in Stance	Device	(1,9) =0.105	.753	.012
	Speed	(1,9) =0.672	.434	.069
	Interaction	(1,9) =0.013	.912	.001
%NPA of Paretic TBA in Swing	Device	(1,9) =0.050	.829	.005
	Speed	(1,9) =56.624	**.000**	**.863**
	Interaction	(1,9) =1.219	.298	.119
%NPA of Paretic GCM in Stance	Device	(1,9) =0.266	.619	.029
	Speed	(1,9) =16.543	**.003**	**.648**
	Interaction	(1,9) =2.381	.157	.209
%NPA of Paretic GCM in Swing	Device	(1,9) =9.958	**.012**	**.525**
	Speed	(1,9) =48.090	**.000**	**.842**
	Interaction	(1,9) =3.270	.104	.266
RMS of ML tilt	Device	(1,9) =17.721	**.002**	**.663**
	Speed	(1,9) =0.008	.932	.001
	Interaction	(1,9) =0.155	.703	.017
RMS of AP tilt	Device	(1,9) =23.647	**.001**	**.724**
	Speed	(1,9) =1.362	.273	.131
	Interaction	(1,9) =1.968	.194	.179

Statically significant p-values and effect sizes are shown in boldfaced font.

is comparable with that reported in a previous work [31]. The peak values of GSF showed a significant difference ($p < .001$) between devices (IC and HC) during walking at the same speed, which confirms that less amount of ground support is taken by the participants while using the HC than while using the IC. This lower amount of force may help reduce the disturbance imposed on the upper body due to use of a cane. However, such lower magnitude forces can still help to bring about gait related improvements, as revealed by R. Boonsinsukh *et al.*, who reported that provision of light touch through an instrumented cane improved ML trunk stability and muscle activity in the paretic leg of stroke survivors [20], [21].

The duration of GSF results are presented in Fig 8(b). At the relatively slower preferred walking speed, the duration of GSF with the HC was significantly greater than that with the IC ($p < .001$). This shows that at this speed the HC provides more continuous support to the user. However, while using either device, walking at the as fast as possible speed resulted in significant reductions in the duration of GSF (ICPS vs. ICAF ($p < .01$), HCPS vs. HCAF ($p < .01$)). This may be indicative of speed dependent variations in the behavior of the force interaction between the user and the HC, where an increase in the gait speed causes an increase in the intermittence of the force. Thus, it can be concluded from these results that at relatively lower gait speeds the HC can provide lesser magnitude more continuous support than the IC, which may be more beneficial for stroke survivors. However, the intermittence of the support appears to increase with the increase in gait velocity.

The exertion of repetitive stresses on the upper extremity joints due to chronic conventional cane use can contribute to pathologies like tendonitis, osteoarthritis and carpal tunnel syndrome [1], [17]. The greater continuousness and lesser magnitude of supporting force observed with the HC can reduce these repetitive stresses, which may help to mitigate the occurrence of such pathologies.

As shown in Fig. 10 (a), the swing SR showed significant improvements between speeds while using the same device (ICPS vs. ICAF ($p < .05$), and HCPS vs. HCAF ($p < .01$)), and between the two devices at the same speed (ICPS vs. HCPS ($p < .001$), and ICAF vs. HCAF ($p < .001$)). These results indicate that while using either the IC or the HC, an improvement in swing SR can occur with the increase in gait speed, but they also show that the use of the HC can be more efficient than the IC in improving swing SR. However, when looking at the step length SR (Fig. 10 (b)), it was confirmed that the use of HC can significantly improve it at the higher speed, whereas no significant difference was observed due to the use of the IC (HCPS vs. HCAF ($p < .05$), and ICAF vs. HCAF ($p < .05$)). These results are supported by previous works that have reported that gait speed of stroke survivors has a strong correlation with the symmetry of temporal gait measures [5], and that the step length SR has a relatively weaker relationship with temporal asymmetry [33], [36].

In persons with chronic stroke, compensatory strategies such as increase or decrease of step length of the paretic or non-paretic limb influence symmetry [36]. Improvement of step length SR is meaningful for the gait of patients because it plays an important role in reducing the cost of transport and in improving the gait pattern and balance [5]. Furthermore, in a clinical setting, step length SR has been proposed as an index to evaluate the propulsive force generated by the paretic leg [33], [37]. P. Padmanabhan *et al.* reported that even when stroke survivors improved step length SR with visual feedback on a treadmill, gait kinematics and kinetics remained markedly asymmetric and that the participants significantly lengthened the shorter step and shortened the longer step to improve step length SR [5]. A similar mechanism of step length SR enhancement was also observed in the current study where the step length SR was significantly improved between ICAF and HCAF ($p < .05$) and this improvement was accompanied by a significant decrease in the stride length ($p < .05$). Furthermore, while walking with the HC at comfortable speed (HCCS), the subjects' stride length was longer at the higher gait speed

(speed = 0.749 ± 0.183 m/s, stride length = 78.965 ± 16.144 cm), indicating that the use of HC at various speeds may be beneficial for stroke survivors.

We also observed that stroke patients could walk at preferred and fast speed while utilizing the continuous proprioceptive augmentation provided by the HC. This proprioceptive augmentation aided the participants in balance control during walking [21]. This is evidenced by the statistically significant differences between devices in RMS of ML tilt within each of the speed conditions used in this study (ICPS vs. HCPS ($p < .01$), ICAF vs. HCAF ($p < .05$)). In addition, RMS of AP tilt during preferred speed trials showed statistically significant difference ($p < .01$) between devices, whereas no statistically significant difference in it was observed during the fast walking trials. Additionally, use of HC at the user selected comfortable speed produced a significant improvement in RMS of AP tilt over that observed during use of the IC at its preferred speed ($p < .001$), showing that use of the HC may provide more balance improvement for the user than the IC.

Encouragement of the use of the affected side can be confirmed through %NPA of the paretic side muscles shown in Fig. 12. In the case of fast walking using the IC compared with preferred speed walking with the IC, significant improvements were observed in the %NPA during the swing phase (VMO ($p < .001$), SMT ($p < .001$), TBA ($p < .001$) and GCM ($p < .01$)). Similar significant improvements were also observed during trials with the HC (VMO ($p < .001$), SMT ($p < .001$), TBA ($p < .005$) and GCM ($p < .005$)). In case of the VMO and SMT muscles during swing, use of the HC resulted in significant increases in muscle activation as compared to the use of IC at the same speeds (VMO: ICPS vs. HCPS ($p < .05$), ICAF vs. HCAF ($p < .05$), SMT: ICPS vs. HCPS ($p < .05$), ICAF vs. HCAF ($p < .001$)). In general, increase in gait speed increases joint ROM and, therefore, increases muscle activity [38]. M. R. Afzal *et al.* showed that increasing walking speed (20 %) with kinesthetic haptic input could lead to improved muscle activity in the paretic leg [34]. Therefore, the overall increase in activity observed here can also be attributed to the increase in speed. A previous study involving the use of an instrumented cane reported that light touch interface with the cane increased %NPA of paretic tensor fascia latae (TFL), semitendinosus (ST), and vastus medialis (VM) in stance [20].

Furthermore, in a study that utilized the previous version of the HC (without ground support), stroke survivors walking at increased device regulated speed showed increased VMO and SMT activity in the stance phase [34]. However, results of the current study suggest that combining the provision of continuous GSF with speed regulation had a more significant effect on activation of VMO and SMT in the swing phase. In case of the GCM muscle, use of the HC resulted in a significantly higher activation during swing phase of faster speed walking but not during the preferred speed walking (ICAF vs. HCAF ($p < .05$)).

During the stance phase, the muscle activations did not show any significant differences between devices. While using the IC, there were no significant differences in muscle activations between different trial speeds. However, while using the HC, there was significant increase in the activation of the GCM between preferred speed and fast speed trials (HCPS vs. HCAF ($p < .01$)). Thus, use of the HC with fast walking speed resulted in increased activation of the GCM during both stance and swing phases of gait. The GCM is mainly responsible for the ankle plantar flexion that generates the propulsive force during stance phase of gait [38]. Thus, the increased GCM activity observed during the stance phase may indicate increased generation of propulsive force, which would be beneficial for stroke survivors and may also result in improvement of the step length SR [33], [37].

Use of the HC in the current work increased %NPA of paretic VMO, SMT and GCM during swing, while increasing the %NPA of only the GCM during stance. Thus, it can be inferred that the provision of low magnitude but continuous GSF with constant speed regulation may have a relatively more significant effect during the swing phase of walking. This inference is further supported by the observation of improved swing phase and step length symmetry.

In this study, the IC GSF, which includes the support taken by the user, the braking force and the propulsive force, is assumed to represent only the support taken by the user. This introduces some error in the analysis. However, since this has no effect on the calculation of the duration of GSF, and the peak of GSF has been reported to occur when the cane is in vertical position [32], which corresponds to the vertical component of force considered for the HC, the error caused by this assumption in our analysis is negligible. Nonetheless, this assumption limits our ability to perform in depth comparisons of the forces experienced with the HC and IC. Another limitation of this study is the use of a non-randomized protocol. However, this was unavoidable due to the requirement of IC trial values during the HC trials, and the effects of this have been mitigated through the provision of rest breaks between trials.

V. Conclusion

In this paper, we have endeavored to show the effectiveness of our developed HC. In order to do so, we also developed an IC and performed experiments to compare the IC and HC in terms of their interaction with the user and their effects on the gait parameters, balance and muscle activations of stroke survivors. Ten hemiparetic stroke survivors participated in walking trials, the results of which showed that the IC could be used by the patient to walk at faster speed with improved stride length and swing symmetry ratio, but the step length symmetry ratio did not show significant improvement. However, use of the HC with less but continuous ground support force could improve swing SR more than the use of the IC while also improving the step length SR. Use of the HC also resulted in greater improvements in the paretic muscle activations with more improvements happening in the swing phase than in the stance phase of gait. Furthermore, comfortable speed with the HC was higher than the preferred speed with the IC, and was accompanied by improved RMS of AP pelvic tilt.

Thus, this study showed that the HC provides more continuous and lower magnitude support force to the user and that the use of HC can result in a higher preferred speed than with the IC while maintaining stable balance. The gait parameter results showed that the HC with its constant speed regulation and continuous proprioceptive interaction could bring about greater level of improvements in SR and gait balance. The provision of low magnitude but continuous GSF and constant speed regulation also resulted in significant improvements in paretic muscle activations with relatively more significant effects observed during the swing phase of walking.

Another interesting finding of the study was that the intermittence of the support provided by HC appeared to increase with the increase in gait speed. This can be explored further with more in-depth studies.

The HC with a simple and intuitive mechanism may provide efficient gait training modes for stroke survivors. However, the current study was carried out with a relatively small subject group and the efficacy of this system as a tool for post-stroke gait rehabilitation needs to be evaluated in detail with a larger population in future works. Furthermore, future work is also required to optimize the angle and control method of the HC.

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