Immediate Biomechanical Effects of Providing Adaptive Assistance With an Ankle Exoskeleton in Individuals After Stroke

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Abstract—Recent studies on ankle exoskeletons have shown the feasibility of this technology for post-stroke gait rehabilitation. The main contribution of the present work is a comprehensive experimental analysis and protocol that focused on evaluating a wide range of biomechanical, usability and users' perception metrics under three different walking conditions: without exoskeleton, with an ankle exoskeleton unpowered, and with an ankle exoskeleton powered. To carry out this study, we developed the ABLE-S exoskeleton that can provide time-adapted ankle plantarflexion and dorsiflexion assistance. Tests with five participants with chronic stroke showed that walking with the ABLE-S exoskeleton significantly corrected foot drop by 25 % while reducing hip compensatory movements by 21 %. Furthermore, asymmetrical spatial gait patterns were significantly reduced by 51 % together with a significant increase in the average foot tilting angle at heel strike by 349 %. The total time to don, doff and set-up the device was of 7.86 ± 2.90 minutes. Finally, 80 % of the participants indicated that they were satisfied with their walking performance while wearing the exoskeleton, and 60 % would use the device for community ambulation. The results of this study add to the existing body of evidence supporting that ankle exoskeletons can improve gait biomechanics for post-stroke individuals.

Index Terms—Prosthetics and exoskeletons, rehabilitation robotics, wearable robotics.

I. INTRODUCTION

THE hemiparetic gait observed after stroke is typically slow, metabolically expensive, and unstable, which hinders community ambulation and promotes a sedentary lifestyle [1]. Ankle weakness is considered one of the major contributors to impaired locomotor function as it is responsible for ineffective forward propulsion and lack of foot clearance, i.e., drop foot, during swing [2], [3]. Compensatory pelvis and hip strategies, e.g. hip hiking and circumduction, and asymmetrical walking patterns are used by people with hemiplegia to compensate for ankle weakness.

Ankle-foot orthoses (AFOs) are commonly prescribed to enable safe and independent walking of people post-stroke by providing stability to the ankle joint during stance and preventing foot drop during swing [4]. Although AFOs are a simple and inexpensive solution, they do not address deficits in paretic propulsion, which can encourage compensatory gait strategies. Furthermore, passive AFOs do not provide a training effect beyond their passive support during walking as an assistive aid.

Recent wearable robotic devices that aim at assisting the ankle joint have shown to have a training effect by improving ankle mobility, gait speed, temporal and spatial symmetries, propulsion force and spinal cord excitability in people post-stroke [5]–[10]. The main contribution of the present work is the comprehensive experimental analysis and protocol, as it is one of the few studies on ankle exoskeletons for people with stroke that evaluate a wide range of biomechanical, usability and users' perception metrics under three different walking conditions, i.e, without exoskeleton, with the exoskeleton unpowered, and with the exoskeleton powered [11]. To carry out our study, we developed the ABLE-S, a unilateral powered ankle exoskeleton that can provide time-adapted assistance based on the duration of previous strides and the current walking state (see Fig. 1(a)). The device can provide active ankle plantarflexion (PF) assistance at push off, and ankle dorsiflexion (DF) assistance during swing phase and early stance. As a secondary contribution of this study, we describe in the Methods section a number of distinct design features of the ABLE-S exoskeleton compared to other ankle exoskeletons.
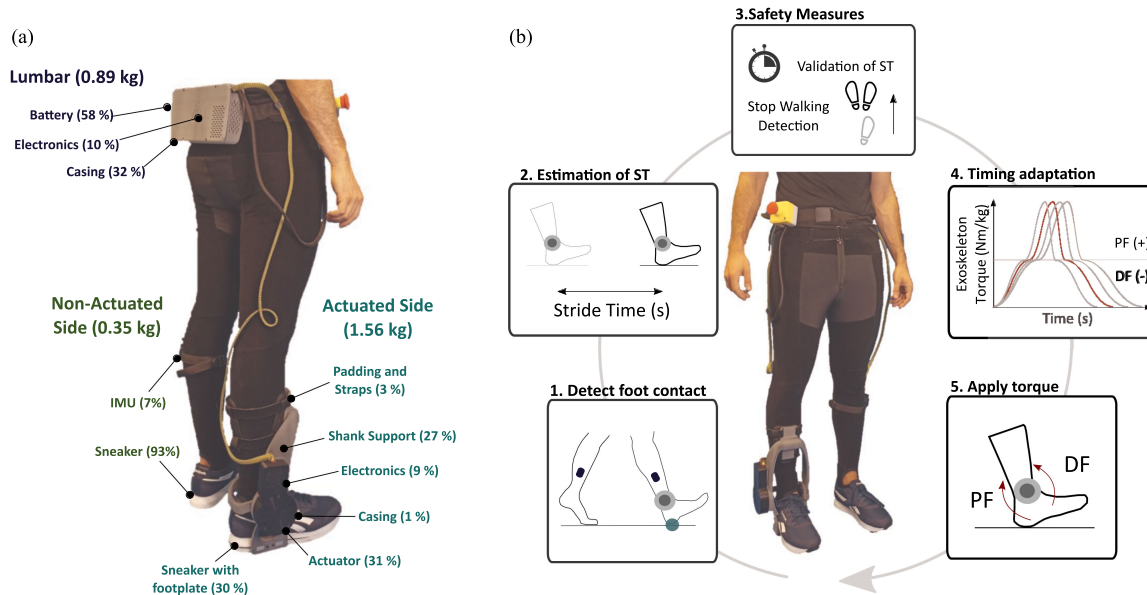
Fig. 1. Overview of the hardware and control of the ABLE-S exoskeleton. (a) Picture of ABLE-S breaking down the device components and mass. (b) ABLE-S control overview: Kinematic information (i.e., angular position and velocity) of the shank of both sides is fed to the exoskeleton controller. The controller is responsible for gait event detection (i.e., heel strike), determination of stride time to adapt the timing parameters of the plantarflexion (PF) and dorsiflexion (DF) torque profiles, and application of measures to ensure a safe interaction between the user and the robot. The controller generates time adapted PF and DF torque profiles in real time that are applied to the ankle joint.

The main objective of this study was to evaluate the immediate biomechanical effects of providing stroke survivors with ankle DF and PF assistance using the ABLE-S exoskeleton. We hypothesized that providing ankle push-off power (i.e., PF assistance) and DF assistance during swing would reduce hip compensatory movements and increase foot clearance. The improvement in gait performance would result in a more symmetrical gait pattern, i.e., longer paretic steps and stance time, and an increased gait speed. We also considered that side-effects, e.g., changes in compensatory pelvis movements, would appear due to the added weight of the device. As a secondary objective, we performed a usability assessment, quantifying the time to don, doff and set up the device, and collected the subjective opinion of the participants by means of questionnaires.

II. METHODS

A. Study Overview

This study was approved by the Medical Research Ethical Committee (MREC) East-Netherlands under the number: NL77959.000.21. Five participants were recruited from Roessingh Centre for Rehabilitation in Enschede, the Netherlands. The current study evaluated differences in gait parameters during three conditions: walking without the ABLE-S (*No Exo*), with the ABLE-S unpowered (*Unpowered*), and with the ABLE-S powered (*Powered*). The study was divided into two measurement sessions for each participant. In the first session, we measured the participants' baseline gait (*No Exo* condition) and, in the second session, we measured participants walking in the *Unpowered* and *Powered* conditions. Donning and doffing the exoskeleton, tuning the exoskeleton parameters and familiarizing with the device took 20 minutes approximately. The time

walking with the device *Unpowered* and *Powered* was 10 min in total. Each walking trial consisted of walking on a 10-meter straight line. Walking trials were practiced by the participants as part of the familiarization process. The number of trials was dependent on the number of valid steps that were made inside the force plates and the participants' level of exhaustion.

B. Study Inclusion and Exclusion Criteria

The main inclusion criteria were the following: (1) age above 18 years, (2) unilateral ischemic or haemorrhagic chronic ($>$ 6 months) stroke, and (3) Functional Ambulation Categories (FAC) score $\geq$ 3. The main exclusion criteria were the following: (1) high levels of spasticity of muscle tone (resistance to passive movement), as represented by modified Ashworth scale scores $\geq$ 3, (2) premorbid disability of lower extremity, and (3) skin lesions or severely impaired sensation at the hemiparetic leg.

C. Exoskeleton Hardware

The ABLE-S device is a wearable, unilateral, powered ankle exoskeleton that provides adaptive assistance to improve forward propulsion and foot clearance in stroke survivors (see Fig. 1(a)). If required, the ABLE-S exoskeleton can be used in combination with a walking aid such as a cane, crutch or walker.

The exoskeleton frame is rigidly attached to a footplate embedded in a commercial sneaker and to the user via straps around the calf. Three different shoe sizes, i.e., 39, 42 and 45 in EU scale, are available and easily interchangeable depending on the participant's foot size. The distal module houses a brushless DC motor, aligned with the ankle joint axis, and several onboard sensors, which measure ankle joint angle and angular velocity, motor current and voltage, as well as the shank acceleration and
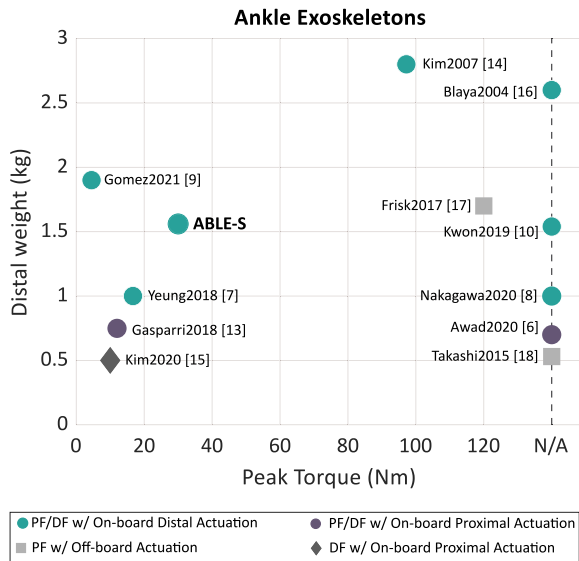
Fig. 2. **Mechanical characteristics of ankle exoskeletons tested on people with hemiplegia.** The graph represents the relation between the distal weight of the exoskeletons and the maximum ankle torques produced by the exoskeletons. The plot also indicates the location (off-board, on-board, proximal or distal) of the actuators and the active degrees of freedom (DF and PF).

angular velocity at 100 Hz (BNO055, Bosch, Germany). The lumbar module contains the battery and control unit, which are housed inside a plastic case and attached to the pelvis using a belt. On the non-actuated side, the user wears an additional IMU sensor to measure kinematic information of the shank in the sagittal plane. ABLE-S weighs a total of 2.80 kg (actuated side: 1.56 kg; non-actuated side: 0.35 kg; lumbar: 0.89 kg). The actuator (AK80-9, T-motor, China) has a maximum angular velocity of 25.65 rd/s and can provide a peak torque of 30 Nm (25 % of the maximum DF-PF peak value for a person who weights 85 kg) [12].

Compared to other ankle exoskeletons, ABLE-S is one of the few untethered ankle exoskeletons tested on people with hemiplegia with on-board actuation that has one of the best characteristics in terms of peak torque and distal weight, while providing both DF and PF assistance (see Fig. 2) [6]–[10], [13]–[18].

D. Exoskeleton Control

The exoskeleton's controller detects user foot contact of both legs following a threshold-based algorithm that uses shank angular kinematics, i.e., angular position and velocity. Stride time is then estimated from previous strides to adapt the timing parameters of the exoskeleton assistance. The exoskeleton generates torque profiles for PF and/or DF assistance, with adjustable timing and magnitude following a pi-shaped function that emulates unimpaired joint dynamics [19]. Note that in the first step the exoskeleton blocks ankle PF with the objective of avoiding foot drop and allows free ankle DF, i.e., zero DF torque, during the first cycle. Moreover, in this study we introduce the concept of *safety layers* within the control loop that has not been introduced by any other study on ankle exoskeletons for people with hemiplegia. Two safety layers ensure a secure interaction

between the user and the exoskeleton: the device is disabled if a non-valid stride time is detected (i.e. stride time outside a predefined range), or if the user stops walking. The stop walking state is detected if the shank orientation and angular velocity are within a defined region during a specific period of the gait cycle. Auditory and visual feedback on the lumbar module inform both the therapist and the user of the system status and operating state.

ABLE-S is the first ankle exoskeleton for people with stroke that does not rely on foot metrics to detect gait events and adapt the robotic assistance [6]–[10], [13]–[18]. The robustness of foot-only related metrics to adapt the exoskeleton assistance in post-stroke subjects might be questionable due to the high gait variability resulting from equinovarus deformity, excessive hip external rotation, ankle weakness, and abnormal muscle activation [20]. On the other hand, metrics from shank kinematics have been shown to be especially robust to detect gait events and provide adaptive assistance in people with hemiplegic gait [21]. Therefore, we have chosen metrics from shank kinematics as input signals for the control of the ABLE-S exoskeleton.

E. Exoskeleton Parameter Tuning

The thresholds of the gait event detection algorithm were manually adjusted during the *Unpowered* condition based on real-time data, i.e., shank angular kinematics and event detection flags. To separately tune the dorsiflexion and plantarflexion magnitude during the familiarization phase, we first applied only dorsiflexion assistance and then added the effect of the plantarflexion assistance. In case the participant did not experience the plantarflexion action as comfortable, only dorsiflexion assistance was applied during the experiments. The control algorithm is adaptive time-wise, but the peak plantarflexion and dorsiflexion torques were manually adjusted based on: (1) ankle kinematics, i.e., ankle joint angle and angular velocity, measured by the exoskeleton sensors and (2) participant's perception, i.e., safety and comfort. The target dorsiflexion peak torque was set at the value that resulted in an ankle dorsiflexion angle between 5–10 $^\circ$ at initial contact. The target level of the applied plantarflexion peak torque was set to 5–20 % (0.0665–0.264 Nm/kg) of the peak plantarflexion torque of unimpaired individuals during walking (1.32 Nm/kg [12]). The chosen values were selected to maximize ankle angular velocity in the terminal stance while ensuring comfort of the participant. These target levels were chosen based on pilot tests performed with stroke individuals.

F. Data and Statistical Analysis

Optical tracking with reflective markers (Vicon Motion System, Oxford, U.K.) was used to record the motion of the participants at a sampling rate of 100 Hz. Markers were placed according to the lower-body plug-in-gait model and additional markers were placed on the exoskeleton. In addition, foot-ground reaction forces and torques were recorded from two in-ground force plates (OR6-5-1000, AMTI, Watertown, MA, USA) with a sampling rate of 100 Hz.

The gait events were determined offline by an algorithm based on the vertical position of the heel and toe markers to

Participant	Side of Paresis	Chronicity (mos)	FMA	FAC	Gender	Age (yrs)	Weight (kg)	Height (m)	Walking Aids
P1	Right	14	15	4	Female	55	108.7	1.70	Cane
P2	Right	9	18	4	Female	55	83.1	1.65	AFO / Cane
P3	Right	41	22	5	Female	19	61.1	1.78	None
P4	Left	9	7	4	Male	72	85.0	1.73	AFO / Quadcane
P5	Left	22	14	4	Female	62	65.0	1.63	AFO / Quadcane

FMA: Fugl-Meyer Assessment; FAC: Functional Ambulation Category; AFO: Ankle Foot Orthosis.

determine initial contact and toe-off events. All the parameters were extracted from the 3D position data of the markers and time-normalized using the initial contact and toe-off events. The data analysis focused on metrics associated to compensatory movements and gait asymmetries, e.g., minimum toe clearance, foot tilting angle at heel strike, step length, step width, circumduction, hip hiking, peak anterior ground reaction force, and spatial and temporal symmetry indexes. Hip hiking has been calculated by measuring the vertical displacement of the marker placed in the trochanter of the ipsilateral side with respect to the contralateral side. Circumduction has been calculated as the maximum medio-lateral displacement of the ankle marker during the swing phase.

A usability assessment was carried out focusing on the time to don and doff, and to operate the device, measured by a digital stopwatch. During the donning of the device, participants inserted their feet into the shoes and tied them tightly. Depending on the upper-extremity impairment level of the participant, a clinical specialist partially assisted in fitting and adjusting the straps on each shank and the lumbar module. During doffing, participants required a a level of assistance similar to the donning. To assess subjective user experience and quantify the usability of the exoskeleton, each participant completed a modified version (only the first eight questions) of the Quebec User Evaluation of Satisfaction with Assistive Technology 2.0 (QUEST 2.0) and answered to open-ended questions about their experience with the ABLE-S device.

Interparticipant means and standard errors are reported for the biomechanical results, while means and standard deviations are reported for the usability results. The percentage of variation between the three conditions and the range across all the participants are also reported. The small-sample Lilliefors correction of the Kolmogorov-Smirnov test was used to check normal distribution of the data, all tests confirmed normality. We used repeated measures one-way ANOVA to compare across *No Exo*, *Unpowered* and *Powered* conditions with a post-hoc paired two-tailed t tests to evaluate statistical significance. The level of significance was set to $p < 0.05$ for this pilot study to allow individual interpretation of p-values. Statistical significance was not validated on dynamic metrics due to the low number of steps on the force plates.

III. RESULTS

Five participants were enrolled in the study and completed the two sessions (see Table I). Three out of the five participants (60 %) were right hemiparetic and all of them had chronic stroke with an average chronicity of 19 months (ranging from 9 to 41

months). Average clinical scores for Fugl-Meyer Assessment (FMA) and Functional Ambulation Category (FAC) were 15.2 (ranging from 7 to 22) and 4.2 (ranging from 4 to 5), respectively. The vast majority (four out of five, 80 %) used passive walking aids (i.e., AFOs, canes and quadcanes) for community ambulation.

Means and standard errors for each outcome of interest for the three conditions (see Fig. 3(a)–(l)), the usability timings divided per processes (see Fig. 3(m)) and the answers to the questionnaires (see Fig. 3(n)–(o)) are presented in this section.

A. Foot Drop Reduction

The paretic foot was significantly more tilted at heel strike when walking with the exoskeleton *Powered* with respect to the *No Exo* (389.8 [144.4, 2793.2] %, $p = 0.007$) and *Unpowered* (134.6 [25.6, 725.3] %, $p = 0.032$) conditions (see Fig. 3(a)).

Significantly higher paretic foot clearance was obtained when walking with the exoskeleton *Powered* (24.6 [3.4, 36.0] %, $p = 0.020$) and *Unpowered* (26.7 [9.3, 41.2] %, $p = 0.009$) in comparison with walking with *No Exo* (see Fig. 3(b)).

B. Paretic Push-Off

We found that the average peak anterior (propulsive) ground reaction force was higher when walking with the exoskeleton *Powered* by 9.9 [−29.4, 107.1] % and 104.0 [46.7, 209.5] % with respect to the *No Exo* and *Unpowered* conditions, respectively (see Fig. 3(d)).

C. Walking Speed

Walking with the exoskeleton *Powered* did not have a significant influence on walking speed (−6.9 [−16.3 – 18.7] %, $p = 0.362$) when compared to walking with *No Exo* (see Fig. 3(c)). Yet, we found that walking speed was significantly increased by 32.8 [6.0, 59.0] % ($p = 0.039$) for the *Powered* condition with respect to the *Unpowered* condition.

D. Hip and Pelvis Compensatory Movements

Although no statistically significant differences were found in terms of circumduction between the three conditions for neither the paretic nor the non-paretic leg, the average circumduction of the paretic side was lower when walking with the exoskeleton *Unpowered* (−21.0 [−49.2, 30.9] %, $p = 0.236$) and *Powered* (−19.7 [−47.8, 0.3] %, $p = 0.126$) in comparison with the *No Exo* condition (see Fig. 3(e)).

Hip hiking of the non-paretic side was not significantly different for any of the three conditions. Yet, we found that walking
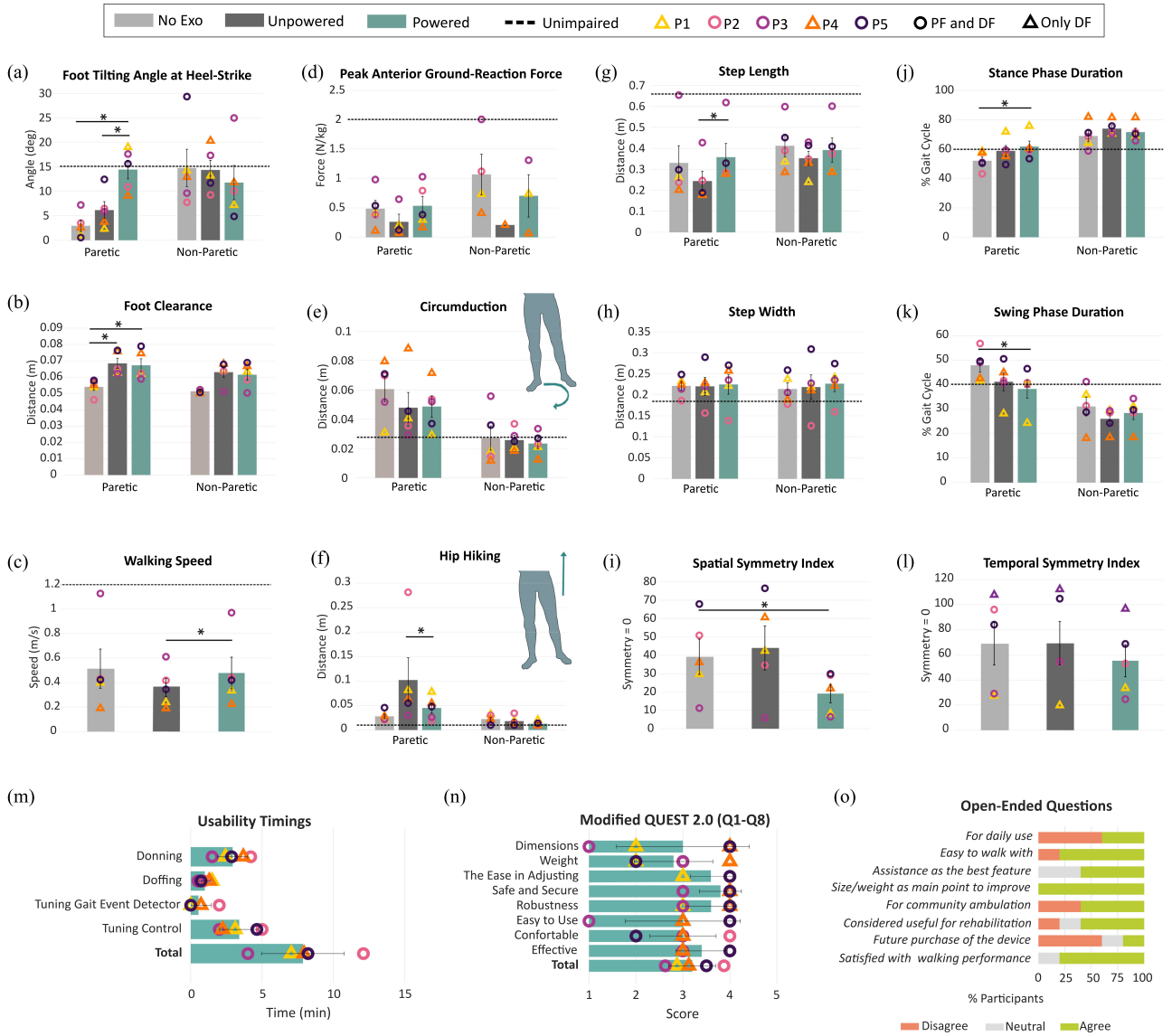
Fig. 3. Biomechanical and usability outcomes: (a-l) Mean and standard error of the biomechanical outcomes of interest for the three conditions *No Exo*, *Unpowered* and *Powered*. Dashed lines indicate values expected for unimpaired people extracted from [12]. (m) Duration of all the preparation phases involved in using the ABLE-S. (n) Scores for the QUEST 2.0 (score of 5 indicates "very satisfied" while 1 indicates "not satisfied at all"). (o) Main ideas collected from the open-ended questions. The mean value of each participant is shown for all metrics. Note that the participants are divided between the ones that received PF and DF assistance (circle), and only DF assistance (triangle).

with the exoskeleton *Powered* significantly reduced the hip hiking of the paretic side with respect to the *Unpowered* condition by -55.9 $[-93.3, -4.3]$ % ($p = 0.016$) (see Fig. 3(f)).

E. Spatial Gait Pattern

Steps with the paretic leg were significantly longer when walking with the exoskeleton *Powered* (47.0 $[15.6, 74.7]$ %, $p = 0.011$) with respect to the *Unpowered* condition (see Fig. 3(g)). The spatial symmetry index was significantly improved in the *Powered* condition with respect to the *No Exo* condition by 51.15 $[39.4, 72.4]$ %, ($p = 0.02$) (see Fig. 3(l)). Step width with the paretic and non-paretic legs remained similar for all three conditions (see Fig. 3(i)).

F. Temporal Gait Pattern

The paretic swing time was significantly reduced and paretic stance time was significantly increased when walking with the exoskeleton *Powered* with respect to the *No Exo* condition by -20.2 $[-41.4, -6.2]$ % ($p = 0.034$) (see Fig. 3(j)) and by 18.6 $[5.3, 37.1]$ % ($p = 0.034$) (see Fig. 3(k)), respectively. Despite these significant improvements on gait phase durations, the temporal symmetry index did not show significant differences between any of the three conditions ($p = 0.175$) (see Fig. 3(l)).

G. Usability Timings

The usability assessment results (see Fig. 3(m)) showed that the majority of the preparation time was spent tuning the control

parameters (3.39 ± 1.36 minutes) and donning the device (2.93 ± 1.05 minutes). The tuning of the event detection algorithm required the least amount of time (0.54 ± 0.87 minutes) and the doffing time (0.98 ± 0.41 minutes) was approximately three times faster than the donning time. The total time including all processes was of 7.86 ± 2.90 minutes.

H. Participant's Perception

The participant's perception of the ABLE-S exoskeleton according to the modified QUEST 2.0 (see Fig. 3(m)) showed a moderate level of acceptability (average score 3.2 ± 0.5). The wearable exoskeleton was perceived to be safe and secure (3.8 ± 0.4), easy to adjust (3.75 ± 0.5), robust (3.6 ± 0.5), and effective (3.4 ± 0.5). The weight of the device was the item with the lowest score (2.8 ± 0.8).

Regarding the answers to the open-ended questions (see Fig. 3(o)), 4 out of the 5 participants were satisfied with their performance walking with ABLE-S and found it easy to walk with it. Three out of the 5 participants thought that the best feature of the exoskeleton was the assistance it provided. All the participants pointed out that the size and weight of the device were the main points to be improved. Finally, 3 out of the 5 participants declared that they would use it for community activities and considered that ABLE-S could be useful for their rehabilitation. However, only one participant had the intention to purchase the device.

IV. DISCUSSION

The purpose of this study was to test the immediate effects of providing PF and DF assistance with a unilateral ankle exoskeleton on a wide range of biomechanical metrics in individuals post-stroke. Our findings supported the hypothesis that the assistance provided with the ABLE-S exoskeleton (*Powered* condition) increased gait performance when compared to the *Unpowered* and *No Exo* conditions. Additionally, positive effects of the *Powered* condition over the *Unpowered* condition pointed out the effectiveness of the control strategy. Altogether, the increase in the walking performance, the low setup time, and the positive perception of the participants with the wearable exoskeleton are encouraging outcomes that indicate the potential of the ABLE-S exoskeleton as an assistive device for individuals post-stroke.

While walking with ABLE-S *Powered*, the participants showed values of step length, foot clearance and foot tilting angle closer to the ones of unimpaired individuals, potentially reducing risk of fall [22]. The reductions of temporal and spatial asymmetries might be related to improvements in paretic push-off and foot clearance [23], [24].

Reductions in the average paretic circumduction for the *Unpowered* and *Powered* conditions might be associated with the higher foot clearance when walking with the exoskeleton compared with the *No Exo* condition [2], [3].

Unexpectedly, foot clearance was significantly increased in the *Unpowered* condition with respect to the *No Exo* condition. We believe that the observed increase in foot clearance in the *Unpowered* condition, as compared to the *No Exo* condition, was not due to a correction of the foot orientation (i.e., foot tilting angle), but to the substantial increase in hip hiking (see Fig. 3(f)).

The active assistance of the exoskeleton contributed to the reduction of hip hiking with respect to the *Unpowered* condition, but not with respect the *No Exo* condition. The asymmetrical weight added to the distal limb might have increased pelvic compensations in the frontal plane for the *Unpowered* and *Powered* conditions. As hip hiking occurs due to a lack of knee flexion angle, we did not expect to see a relevant positive effect on this metric as we were not targeting the knee joint [25].

With the ABLE-S *Powered*, we have not seen substantial differences in walking speed with respect to the *No Exo* condition, as we did not detect significant differences in paretic step length and paretic propulsion either [26]. This low variation in walking speed is coherent with results from the literature [6], [9], and might be attributed to initial hesitation and short familiarization time with the device. Moreover, the participants had a limited space to accelerate, decelerate and turn 180 degrees due to the laboratory settings. For the *Unpowered* condition, the added weight of the device might have been translated into slower gait pace. Walking faster with the exoskeleton *Powered* than with the exoskeleton *Unpowered* is however a good indicator of the effectiveness of the implemented control strategy.

The usability assessment revealed interesting aspects to improve in future design iterations of the device. The total usability time was 3 minutes higher than the expected value of 5 minutes. The time to don and setup was considerably lower than the one in other exoskeletons, which required up to 20 minutes to don [27]. We found that the majority of the setup time (43 %) was needed to tune control parameters. Strategies to automatically adapt the magnitude of the assistance might thus improve the usability of the device.

The results obtained in this study regarding the participant's perception contribute to the increasing evidence on the acceptance of robotic devices for gait assistance. However, lower scores in satisfaction were obtained in comparison to other novel exoskeletons for people with stroke [5]. The device weight and low familiarization time might have led to this result.

While these exploratory findings should be interpreted with caution due to the small sample size and low training time with ABLE-S, our results contribute to a growing body of evidence that powered adaptive ankle assistance can improve walking performance in individuals after stroke. The main limitation found while analyzing the results was the low training time with ABLE-S. Considering that the time for unimpaired people to adapt to robotic ankle assistance is between 20 and 90 min [28], the training time of this study (15 min) was likely insufficient for people with stroke to fully adapt to ABLE-S. These limitations also hindered comparing the effects of receiving both dorsiflexion and plantarflexion assistance or only dorsiflexion assistance.

To explore the maximum performance of the ABLE-S ankle exoskeleton, future work will focus on carrying out studies with larger cohorts and longer familiarization and training times. Future studies will also investigate the motor learning process when walking with the ankle exoskeleton. Design iterations also have to be carried out to reduce the mechanical interference when the device is *Unpowered*.

V. CONCLUSION

In conclusion, the assistance provided by the ABLE-S exoskeleton can improve gait biomechanics of individuals with mild-to-moderate gait impairments due to stroke. After only 15 minutes of walking, we found that participants showed relevant benefits in terms of spatial and temporal symmetry together with higher foot clearance and reduction in hip compensatory movements. Design iterations focusing on weight reduction and control adaptability have to be implemented to increase the performance, usability and acceptability of the ABLE-S exoskeleton. Our study results add to the existing body of evidence supporting that ankle exoskeleton assistance can safely and effectively improve gait biomechanics for individuals post-stroke.

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