Changes in Muscle Activity with Assistance Level During Exoskeletal Gait in Acute Stroke

Ben P.F. O'Callaghan (ben.o-callaghan@ucdconnect.ie)
University College Dublin https://orcid.org/0000-0002-5550-6901
Matthew W. Flood
University College Dublin
Michele Tonellato
University of Padova Department of Neuroscience: Universita degli Studi di Padova Dipartimento di Neuroscienze
Caitríona Fingleton
Dept. of Physiotherapy, Mater Misericordiae University Hospital
Madeleine M Lowery
University College Dublin
Olive Lennon
University College Dublin

Research

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Abstract

Background

The level of assistance provided to the user is an important decision in rehabilitation training using robotic devices. Both fully assistive and assist-as-needed paradigms have shown benefits in functional outcomes in healthy individuals and users with chronic stroke and spinal cord injury. The effect of assistance level on muscle activity and kinematic gait parameters has not yet been directly examined during overground exoskeletal gait in a stroke population. Furthermore, it is not clear whether an assist-as-needed approach could elicit increased voluntary activity in individuals in the acute stages of stroke. The aim of this study was to examine the effect of assistance level on muscle activity and kinematic parameters during exoskeleton gait in individuals in the acute stage of stroke care.

Methods

Nine individuals in the acute stage of post-stroke care performed walking tasks in the EKSO GT^{TM} exoskeleton using both maximal assistance and adaptive assistance control paradigms. Temporal gait parameters and muscle activity were recorded using accelerometers and surface EMG on the lower limb muscles.

Results

Shorter swing times and longer double support times were observed on the non-paretic side during adaptive assist mode than with maximum assist mode ($p<0.0065$). No significant effect of exoskeleton mode was observed on the remaining temporal gait parameters. On the paretic side, proximal lower limb muscles (RF and ST) and plantar-flexors (SO) ($p<0.00125$) exhibited greater activation in adaptive assist mode than in maximum assist mode. On the non-paretic side however, the lower limb distal muscles (TA and SO) displayed greater activity during maximum assist mode than adaptive assist mode ($p<0.00125$).

Conclusions

The level of assistance provided by an exoskeleton in the acute stages of stroke care is an important clinical decision. The results indicate that an adaptive or assist-as-needed approach elicits higher levels of activation in muscles acting around the knee joint and plantar-flexors on the paretic side than a maximal, fixed assistance paradigm, in the acute stage post-stroke. Increased activity around the ankle joint during maximum assistance mode was also noted. Improved understanding of the effect of assistance level can help inform future control paradigms for exoskeleton gait in acute stroke.

Background

Early, repetitive, task specific practice is recommended to increase the likelihood of positive functional outcomes and independence following stroke (1, 2). Traditional gait re-education training is labour intensive, requiring manual positioning of the lower limb/s throughout the gait cycle (3, 4). Body-weight supported treadmill training became increasingly popular in practice due to the reduction in therapist burden and relative increase in the number of repetitions of the gait cycle that users are able to perform during training sessions (4–7). However, a recent systematic review of treadmill training (with or without body weight support) after stroke by Mehrholz et al. concluded that while benefits in outcomes of walking speed and endurance may result when compared with other interventions, the studies identified do not support treadmill training for restoration of independent walking for individuals after stroke who require assistance or are non-ambulatory (8).

Gait rehabilitation using robotic exoskeletons has been shown to be equivalent to conventional methods involving manual input from therapists, with sub-acute, non-ambulatory patients showing added benefits (5, 9, 10). A recent Cochrane review of electromechanically assisted gait training after stroke identified strong evidence supporting its use in combination with physiotherapy to increase the odds of participants becoming independent in walking (OR 2.01; 95% CI 1.51 to 2.69) (9). When compared with treadmill-based methods, robotic overground exoskeletons have been shown to improve trunk muscle activation, which may improve postural control, while also having the added benefit of providing users with multi-sensory feedback such as somatosensory visual and vestibular feedback that can help re-integrate the cortical and cognitive mechanisms of gait, providing a more comprehensive and intensive rehabilitation strategy (11–14).

A common target of exoskeletal use is to reduce muscle recruitment, however, this is often in the context of military or labour settings (15–17). In a rehabilitation setting, particularly conditions such as stroke, it is desirable to have increasing levels of task-specific muscle recruitment during daily tasks such as locomotion (18, 19). The level of assistance provided by an exoskeleton is an important consideration in the clinical setting, and should be considered alongside the impairment level after stroke and the aims of the therapy. Recent gait rehabilitation paradigms involving robotic devices use one of two approaches, the first where the individual is largely passive and the device controls the user’s gait and the second being an assist-as-needed approach, where the robotic device provides assistance to the user as necessary during the gait cycle (20–22).

The assist-as needed approach has been shown to reduce passiveness in healthy, chronic stroke and spinal cord injury users and elicit positive temporal and functional recovery outcomes while also introducing an element of variability to the movement, thought to be a critical factor in motor learning (20, 23–28). Lin et al. have shown that a reduction in body weight support and guidance force while walking in the Lokomat can increase the activity of specific muscles in chronic stroke survivors (29). Although assist-as needed approaches have shown positive functional outcomes in individuals with chronic stroke (> 6 months) (20, 24, 27), the differences between fully assistive and assist-as-needed robotic assisted overground gait training on neuromuscular function in the acute phase following stroke has yet to be elucidated. Furthermore, it is not clear how differences in the level of assistance provided by the device influences muscle activity across the lower limb muscles. This information is critical for therapists in making informed decisions on how to plan and programme robotic assisted rehabilitation programmes in acute stroke care.

To address this, the aim of this study was to compare the effect of a fully assistive and assist-as-needed approach on gait timing and muscle activation during exoskeleton gait in individuals after stroke in an acute hospital setting. Surface electromyography (EMG) activity was examined in individuals in the acute/subacute phase of stroke during gait training using the Ekso GT® (Ekso Bionics, Inc. Richmond, CA, USA) overground exoskeleton. Two assistance levels were examined, a maximum assist mode and an adaptive assist mode. In the maximum assist mode, the motors at the hip, and knee of the exoskeleton guide the lower limb through the pre-programmed spatial trajectory at all points during the gait cycle, while in the adaptive assist mode, the motors only provide assistance to the joints where deviation from the pre-programmed spatial trajectory is identified. It was hypothesised that muscle activity and gait timing parameters would vary with the level of assistance provided by the exoskeleton and that greater levels of muscle activity would be elicited during gait in the assist-as-needed approach.

Materials And Methods

Participants

Nine clinically stable individuals (64.6 ± 13.7 years, 4 female) were recruited following stroke in an acute hospital setting (Table 1) (Mater Misericordiae University Hospital, Dublin). The study was approved by the hospital's Research Ethics Committee and written informed consent was obtained from all participants. A summary of participant demographic and clinical information is provided in Table 1.

Table 1: Patient information (MRS: Modified Rankin Scale, FAC: Functional Ambulation Category, NR: Not Recorded, PCA: posterior cerebral artery, MCA: middle cerebral artery, ACA: anterior cerebral artery)

ID	**Age (years)**	**Sex**	**Stroke Type**	**Stroke Location (Artery / Territory)**	**Time since Stroke (days)**	**MRS Pre-Stroke**	**MRS Post-Stroke**	**FAC Pre-Stroke**	**FAC Post-Stroke**
1	76	F	Ischemic	PCA	19	0	4	5	2
2	90	F	Ischemic	PCA	28	4	5	NR	2
3	67	F	Ischemic	MCA	14	1	2	NR	2
4	61	M	Ischemic / Haemorrhagic	MCA	42	NR	NR	NR	0
5	47	F	Ischemic	MCA	50	0	4	5	3
6	61	M	Ischemic	ACA, MCA	51	0	4	5	1
8	60	M	Ischemic	Basilar large vessel occlusion	57	0	5	5	0
9	73	M	Ischemic	PCA	10	0	3	5	3
10	47	M	Haemorrhagic	Frontal lobe	39	0	NR	5	0
Mean	**64.67 ± 13.7**				**38.9 ± 21.5**	**0.5 ± 1.3**	**4 ± 1.06**	**5 ± 0**	**1.3 ± 1.25**

To satisfy the dimensions of the exoskeleton, recruited participants had to measure between 1.58 and 1.88m in height with a maximum hip width of 0.46m and maximum weight of 100kg. Exclusion criteria included impaired cognition, other neurological conditions or co-existing conditions that limited device use, severe spasticity (modified Ashworth Scale of 4) (30) and acute medical illness. Participants were familiarised with the EKSO device prior to the day of data collection. Anthropometric measurements were taken and the exoskeleton dimensions were adjusted accordingly for each participant.

Experimental Protocol

With the assistance of an EKSO certified physiotherapist to programme the device and provide stability, participants were brought from sitting to standing in the device and robotic assisted walking was commenced in either maximum assistance or adaptive assistance mode, using a pre-prepared randomised schedule to minimise learning/training effects. In both modes, the exoskeleton was set to use the “ProStep” mode, by which forward and lateral movement of the users’ bodyweight would trigger the next step (31). Once walking efficiently, experimental trials commenced along a 10m length. Data were collected using repeated 10m walks. Participant reported perceived exertion and fatigue were used to dictate the number of walking trials in each mode. A rest period of up to 15 minutes was provided between modes. Each participant on average performed two 10m repetitions in each mode (minimum 1 rep; maximum 3 reps).

EMG and Accelerometry Data Acquisition

Bipolar surface EMG and accelerometer data were recorded bilaterally (Delsys Trigno®, Natick, MA, USA) from the tibialis anterior (TA), soleus (SO), rectus femoris (RF) and semitendinosus (ST) muscles, with the sensors placed following SENIAM guidelines (32). The EMG signals were bandpass filtered between 20-450Hz and sampled at 1925.93Hz, while the accelerometer data was sampled at 148.14Hz. The skin at the electrode interface was lightly abraded and cleaned using alcohol prior to application of the electrodes. Correct electrode placement was confirmed by asking the participants to contract each muscle separately. An example of the raw EMG recorded from a sample participant is shown in Figure 1, below.

Data Analysis

Bouts of continuous gait were automatically determined from the shank accelerometer data using a custom developed algorithm to distinguish regions of activity from regions of rest (33). To summarize briefly, the signal vector magnitude, SVM, of the tri-axial accelerometer data was calculated as

$$SVM[n] = \sqrt{\sum_{k=1}^{K} x_k[n]^2}$$

Where $X_k[n]$ denotes the n^{th} sample of the k^{th} dimensional acceleration signal. The SVM accelerometer signals were then filtered using a 4^{th} order Butterworth band-pass filter (0.5-2Hz). To ensure that movement was present in both legs simultaneously, the dot product between the filtered SVM signals was calculated and the resultant was rectified to form the activity signal.

$$activity[n] = \left|Filtered\_SVM_{right}[n] \cdot Filtered\_SVM_{left}[n]\right|$$

Where activity[n] is the n^{th} sample of the activity signal and $Filtered\_SVM_{side}[n]$ is the n^{th} sample of the filtered SVM signal for each side. Before determining bouts of gait, the rectified accelerometery data was smoothed using a moving average window. The power spectrum of the activity signal was calculated and the frequency of the first prominent peak was determined, with its reciprocal being used as the window length for smoothing the signal. Finally, an adaptive thresholding method based on Otsu's image thresholding algorithm was employed to detect the bouts of activity from the accelerometer data (34).

Initial and final contacts of the foot during gait were then estimated from accelerometer data recorded at the shank using an algorithm based on the Teager Kaiser Energy Operator (TKEO) (35,36), and where necessary, manually edited. The initial and final contact times were used to segment the data in each detected walking bout into individual gait cycles. Each gait cycle was then further separated into phases of the gait cycle. Stride times were calculated as the time taken for a complete gait cycle to be completed (time from heel strike to the next successive heel strike on the ipsilateral leg). The stride times were then divided into the stance and swing times, where the stance times were calculated from an initial heel strike to the next toe-off on the ipsilateral leg while the swing times were calculated from the toe-off to the next heel strike for the ipsilateral leg. Finally, double stance times for each leg were defined as the time between the heel strike to the next toe-off on the contralateral leg, Figure 2.

Using the gait cycle timing information, the root mean square (RMS) amplitude of the EMG data for each muscle during each gait cycle was estimated using a moving window of duration of 125ms with 75% overlap. Each gait cycle was normalised in time from 0-100% of gait cycle and the EMG RMS amplitude during each gait cycle was interpolated to 5000 samples using a Fourier transform interpolation to obtain the EMG envelope. The EMG envelopes for each muscle were then normalised with respect to the peak EMG RMS amplitude recorded for that muscle during a single recording session in the maximum assist mode (37). Muscle activation was determined using an adaptive thresholding method, based on Otsu's image thresholding method, applied to the linear envelope of the muscle activity for each muscle, participant and gait cycle (34). The adaptive threshold was applied to the EMG envelope for each muscle using a moving window approach. The length of the moving window was determined by the mean stride time as determined from the accelerometer data previously.

A number of features of the accelerometer and EMG data were examined to quantify changes in muscle function and timing of gait events with exoskeleton assistance level. Following estimation of the heel strike and toe-off events during gait, stride, swing, stance and double support times were estimated. To assess changes in muscle function, the integral of the EMG RMS envelope over the active portions of the gait cycle, the maximum amplitude and location of the peak amplitude of the EMG RMS envelope during the stance and swing phases of the gait cycle, were estimated. All analysis was performed using MATLAB R2019b.

Statistical Analysis

The relationships between exoskeleton mode and the EMG and accelerometer features were examined using linear mixed effects models. Separate models were fitted to assess each feature in all muscles. Exoskeleton mode was chosen as a fixed effect, while participants were chosen to have a random intercept. Visual inspection of residual plots did not reveal any obvious deviations from homoscedasticity or normality. Statistical significance using p-values was assessed by conducting likelihood ratio tests of the full model with the fixed effect of exoskeleton mode against a null model that excluded the factor of interest as a fixed effect. In these tests, multiple comparisons were accounted for by Bonferroni adjustment as in (26,38) (α_{ACC} = 0.00625, α_{EMG} = 0.00125). All statistical analysis was performed in R version 4.0.2 (R Core Team, 2020) using the lme4 (39) library.

Results

Kinematic Assessment

Average stride times in the adaptive assist mode on the paretic and non-paretic side, were 3.68 ± 0.65s and 3.67 ± 0.65s respectively, and in the maximum assist mode were 3.81 ± 0.82s and 3.80 ± 0.77s. No significant effect of exoskeleton mode was observed for the stride times ($p = 0.08$), stance times ($p = 0.42$), swing times ($p = 0.01$) or double support times ($p = 0.25$) on the paretic side, Fig. 3.

On the non-paretic side, swing times were significantly shorter in adaptive assist mode then in maximum assist mode ($p < 0.00625$) and double support times were significantly longer in adaptive assist mode ($p < 0.00625$), Fig. 3. There was no significant effect of exoskeleton mode on stride times ($p = 0.04$) or stance times ($p = 0.75$).

Comparing across sides, there were no significant differences in stride ($p = 0.733$), stance ($p = 0.944$), swing ($p = 0.433$) or double support times ($p = 0.182$) in either mode.

Neuromuscular Assessment

Effect of Side

Differences in the integral of the EMG RMS amplitude and the maximum EMG amplitude were first compared between the paretic and non-paretic sides. The integral of the EMG RMS amplitude was significantly higher on the non-paretic side than on the paretic side for all muscles in both adaptive assist ($p < 0.00125$) and maximum assist modes ($p < 0.00125$), Fig. 4. Similarly, the maximum EMG amplitude was significantly higher on the non-paretic side than on the paretic side for all muscles and in both assistance modes ($p < 0.00125$), Fig. 4.

Effect of Exoskeleton Mode

Paretic side

On the paretic side the normalised integral of the EMG RMS amplitude during the stance phase was significantly greater in the adaptive assist mode than in the maximum assist mode for the RF ($p < 0.00125$), ST ($p < 0.00125$) and SO ($p < 0.00125$). There was no significant difference in the normalised EMG RMS amplitude between modes for the TA on the paretic side during the stance phase ($p = 0.53$), Fig. 5. During the swing phase of gait, the normalised integral of the EMG RMS amplitude was significantly lower in adaptive assist mode than in maximum assist for the ST ($p < 0.00125$), while there was no significant difference with respect to exoskeleton mode for all other muscles; RF ($p = 0.025$), ST ($p = 0.189$) and TA ($p = 0.323$).

The maximum value of the normalised EMG amplitude on the paretic side was also significantly greater during the swing phase in the adaptive assist mode than in the maximum assist mode for the RF ($p < 0.00125$) and SO ($p < 0.00125$). After Bonferroni correction, there were no significant differences with respect to mode for the ST ($p = 0.005$) or TA ($p = 0.035$) muscles for the maximum amplitude in the swing phase of gait.

Examining the maximum EMG RMS amplitude during the stance and swing phases, during the stance phase, the maximum EMG amplitude normalised with respect to the maximum value recorded from that muscle in maximum assist mode on the paretic side, was significantly greater in the adaptive assist mode than in the maximum assist mode for the ST ($p < 0.00125$) and SO muscles ($p < 0.00125$). No significant difference in maximum value of the EMG RMS amplitude with respect to mode was observed for either the RF ($p = 0.21$) or TA ($p = 0.17$) muscles. During the swing phase, the maximum normalised EMG RMS amplitude was significantly greater in adaptive assist mode for the RF ($p < 0.00125$) and SO muscles ($p < 0.00125$). There were no significant differences in the maximum EMG RMS amplitude across exoskeleton modes during the swing phase for either the TA ($p = 0.035$) or ST ($p = 0.005$).

To determine whether there were changes in the timing of muscle activation, the location of the peak amplitude of the EMG in each phase of gait was assessed. During the stance phase, the TA ($p < 0.00125$) and ST ($p < 0.00125$) muscles on the paretic side both reached their peak amplitudes at a significantly earlier stage of the stance phase of gait in adaptive assist mode than in the maximum assist mode. There was no significant change in the timing of the peak muscle activity for the RF ($p = 0.077$) or SO ($p = 0.833$) muscles on the paretic side during the stance phase of gait. During the swing phase on the paretic side, the ST reached its peak amplitude at a significantly later stage in the adaptive assist mode compared to the maximum assist mode ($p < 0.00125$). The location of the peak EMG amplitude during the swing phase of gait did not significantly change on the paretic side for the RF ($p = 0.75$), TA ($p = 0.002$) and SO ($p = 0.083$).

Non-paretic side

On the non-paretic side, during the stance phase of gait, the integral of the normalised EMG RMS amplitude was significantly lower in the adaptive assist mode than in the maximum assist mode for the TA ($p < 0.00125$) and SO ($p < 0.00125$). There were no significant differences in the integral of the normalised EMG RMS amplitude for the RF ($p = 0.14$) and ST ($p = 0.45$) with respect to exoskeleton mode, Fig. 5. During swing there was no significant difference in the integral of the normalised EMG RMS amplitude with respect to exoskeleton mode for any muscle; RF ($p = 0.055$), TA ($p = 0.025$), SO ($p = 0.017$) and ST ($p = 0.576$).

Consistent with this, the maximum normalised EMG amplitude on the non-paretic side was significantly lower in the adaptive assist mode than in the maximum assist mode for the TA ($p < 0.00125$) and SO ($p < 0.00125$), while there was no significant difference for the RF ($p = 0.52$) or ST ($p = 0.42$) muscles. Similarly, in the swing phase of the gait cycle, the maximum normalised EMG amplitude was significantly lower in the adaptive assist mode than in the maximum assist mode for the TA ($p < 0.00125$) and SO ($p < 0.00125$) muscles. There was no significant difference between modes in the maximum normalised EMG amplitude for the RF ($p = 0.044$) and ST ($p = 0.005$) muscles during the swing phase for the non-paretic side.

To assess differences in muscle timing on the non-paretic side during the stance and swing phases of gait, the location of the peak normalised EMG amplitude was determined. During the stance phase of the gait cycle, the

RF ($p < 0.00125$) and TA ($p < 0.00125$) muscles reached their peak activity at an earlier point in the gait cycle during the maximum assist mode compared to the adaptive assist mode, Fig. 6. No difference in the timing of the ST ($p = 0.0075$) and SO ($p = 0.118$) peak activity during the stance phase of gait was observed with mode. During the swing phase of the gait cycle on the non-paretic side, the TA reached its peak activity significantly later during maximum assistance ($p < 0.00125$), while the SO reached its peak amplitude earlier during maximum assistance compared with adaptive assistance ($p < 0.00125$). There were no significant differences in the location of the maximum amplitude of EMG for the RF ($p = 0.278$) or ST ($p = 0.69$) on the non-paretic side with respect to exoskeleton mode.

Discussion

This study compared the effects of a maximum assistance and assist as needed paradigm provided by an overground exoskeleton during gait on muscle activity and gait timing in a group of individuals with stroke in the acute hospital phase of recovery. Greater activity of the muscles acting around the knee joint and the plantar-flexors on the paretic side was elicited in the adaptive assistance mode than the maximum assistance mode, suggesting higher levels of engagement in the walking task. Maximum assistance mode however, induced greater muscle activation in the TA and SO muscles on the non-paretic side, in phases of the gait cycle where excessive activity is unlikely to occur during healthy overground gait. Collectively, the results suggest that the mode of control of an overground exoskeleton and the degree of assistance provided can influence muscle activity and may have an important role in the context of motor relearning.

Temporal gait parameters

Stride and stance times were not statistically different regardless of which leg the participants were stepping with or which mode of assistance was used in the exoskeleton, Fig. 3. Faster swing times were observed on the non-paretic side when the exoskeleton was in adaptive assist mode in which participants have increased ability to control the swing motion, Fig. 3. These faster swing times may simply reflect greater voluntary control of swing on the less affected side. Reduced stability during single leg stance on the paretic side however, may also result in patients completing the swing phase in a faster time to regain stability, thereby minimising single leg support time. Given that muscle activity around the knee was seen to increase in the adapt assist mode in the paretic side, this is unlikely to be the case in the current study (40).

In both modes, to trigger a step, the user must transfer their weight in the forward and lateral direction past a pre-programmed threshold. Double support times on the non-paretic gait cycle were significantly longer during adaptive assist mode, reflecting slower weight transfer movement to the non-paretic side. EMG analysis revealed increased activity of the plantar-flexors on the paretic side during adaptive assist which could suggest greater effort on the part of the user in performing the weight transfer movement in the adaptive assist mode. In this case, although there was increased activation of the paretic plantar-flexors during adaptive assist mode, they were slower performing the push-off stage of the weight transfer motion than when the exoskeleton was in maximum assistance mode. An emerging evidence-base suggests that temporal asymmetry of gait may be improved with robotic assistance after stroke (41), with clinicians often opting for a maximum assist setting to ensure a symmetrical pattern is delivered. However, in the current study no significant differences were observed in temporal gait parameters during adaptive assist mode on the paretic side.

Muscle activity

Substantially higher EMG amplitude was observed across all muscles in the non-paretic side when compared to the paretic side in both modes of activation, Fig. 4 (42, 43). Increased activity of the proximal muscles acting around the knee joint (RF and ST) on the paretic side, indicated by significantly higher EMG amplitude over the gait cycle, was observed during adaptive assistance when compared to maximal assistance mode, Fig. 5. These findings were further supported by the increased peak amplitude of the EMG for the ST during the stance phase and RF during the swing phase of gait in the adaptive assist mode on the paretic side. Early training after stroke is reported as critical in promoting recovery of impairment and brain reorganization after injury (44), with robotic assisted devices allowing repetition of a reciprocal, non-compensatory pattern. Rehabilitation paradigms need to distinguish between impairment and function during motor learning for the stroke population (45)and higher levels of knee muscle activation around the hemiparetic knee, in the presence of non-compensatory gait patterns during adaptive assistive mode suggest this mode better targets impairment at muscle level.

Activity levels in the muscles controlling the ankle joint (TA and SO) are not as easy to interpret during robotic gait as the device used has a passive ankle joint and rigid footplate, limiting normal movement of the ankle joint during gait. The plantar-flexors (SO) on the paretic side exhibited increased muscle activity over the entire gait cycle in adaptive assistance than in maximum assistance mode, also reflected in the increased peak amplitude of the EMG for the SO during both the stance and swing phases of gait in adaptive assistance mode. Increased plantar-flexor activity during the swing phase of gait is an abnormal event when compared to healthy overground gait. Lower leg dorsi- and plantar-flexors (TA and SO) on the non-paretic side showed increased peak amplitudes in both swing and stance phases of gait as well as overall increased activity in maximum assist mode, Fig. 5. Participants reported feelings of imbalance and instability, similar to what has been reported previously by exoskeleton users during overground gait (46). Observing excessive activity of the dorsi- and plantar flexors in areas of the gait cycle that would not be expected under healthy normal conditions, we hypothesise that the increased activity in the muscles acting around the ankle joint resulted from a resistance against the pre-programmed movements of the exoskeleton in the maximum assistance mode and restriction of movement at the ankle joint due to the fixed foot plate. Similar increases in muscle activity during the swing phase of exoskeleton gait were reported in healthy controls and also hypothesised to be due to the mechanical constraints of the device (10, 47, 48). This altered activity is also evident in the adaptations to the time to peak activity in both the swing and stance phases of gait for the TA muscle and in the swing phase only for the SO during maximal assisted gait on the non-paretic side, Fig. 5. Across the population, the TA on the paretic side reached its peak activity at a later point in the stance phase of gait, but at an earlier point of the swing phase of gait during maximum assistance, while the SO during the swing phase reached its peak activity at a significantly earlier point of the gait cycle, Fig. 6.

Collectively, the results indicate changes in EMG activity (Cohen's d values ranging from 0.05 to 0.42) between adapt assist and maximum assist exoskeleton modes during overground walking in acute stroke survivors, with the largest effects seen in the RF (d = 0.36) and SO (d = 0.42) muscles on the paretic side. Reducing the level of assistance tended to increase muscle activity on the paretic side. Similar changes were not consistently observed on the non-paretic side, though increased muscle activity around the ankle joint was observed in the maximum assistance mode. Although all participants in this study were recruited from the acute hospital stage with an acute stroke related gait impairment, in a relatively small cohort, profiles varied widely in terms of age, stroke severity and resultant disability (highlighted by modified Rankin Scale and Functional Ambulatory category scores, Table 1). Heterogeneity was also evident in the EMG profiles during gait and limits within group comparisons.

A number of study limitations should be considered when interpreting the results. The number of walking trials for each subject was low overall. This was determined by the participants' reported rate of perceived exertion and fatigue levels. Low levels of cardiorespiratory fitness are reported early after stroke (49)and overground robotic gait has previously been identified as conferring a moderate intensity aerobic demand in a comparator neurological population (50). Therefore, this limitation may apply with any exoskeleton study in the acute hospital setting. Future study protocols should plan data collection over a number of days to build larger sets of trial data. Trials of overground walking without the exoskeleton device would have allowed the influence of the device on the symmetry of participants' gait and muscle activation to have been further elucidated. However, as evidenced in Table 1, 80% of study participants in this acute hospital setting were physically dependent for walking and trials of overground independent gait were not feasible.

Conclusions

The results of this study demonstrated that in participants in the acute stage of care post-stroke an assist-as-needed paradigm elicited greater muscle activity on the paretic side when compared with a fully assisted paradigm during overground exoskeleton walking in an overground exoskeleton. Increased activity of the distal muscles acting around the ankle joint during the maximum assistance mode was noted, which we hypothesise to be unnatural activity caused by the fixed trajectory paradigm and rigidity of the footplate. Double support and swing times on the non-paretic side were significantly different between modes. Where the therapeutic goal seeks to increase muscle activity in the lower limbs during gait training, notably at the knee, an assist-as-needed approach may be appropriate. The results presented can help inform future control paradigms for exoskeleton gait in acute stroke and exoskeletal design considerations.

Declarations

Ethics approval and consent to participate

Ethics approval was sought and obtained on 23rd May 2019 from Mater Misericordiae University Hospital and Mater Private Hospital Institutional Review Board (IRB): Institutional Review Board Reference: 1/378/2052. Informed consent was sought from all participants prior to their inclusion in the study.

Consent for publication

Consent for publication was sought from all participants at the point of attaining consent for participation in the study.

Availability of data and materials

The datasets used and analysed during the current study are available from the corresponding author on reasonable request.

Competing interests

EKSO Bionics Inc. are a partner in the EU funded PROGAIT project and provided the EKSO Bionics GT device and the relevant training for use of the device for the purposes of research of this study. None of the authors have any affiliations or hold any shares with EKSO Bionics Inc.

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Authors' contributions

BO’C, MF, M.T, CF and OL were responsible for the data collection. MT, CF and OL recruited the participants for the study. BO’C, OL and ML were responsible for the data analysis. BO’C, OL and ML were responsible for writing and editing the manuscript.

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Figures

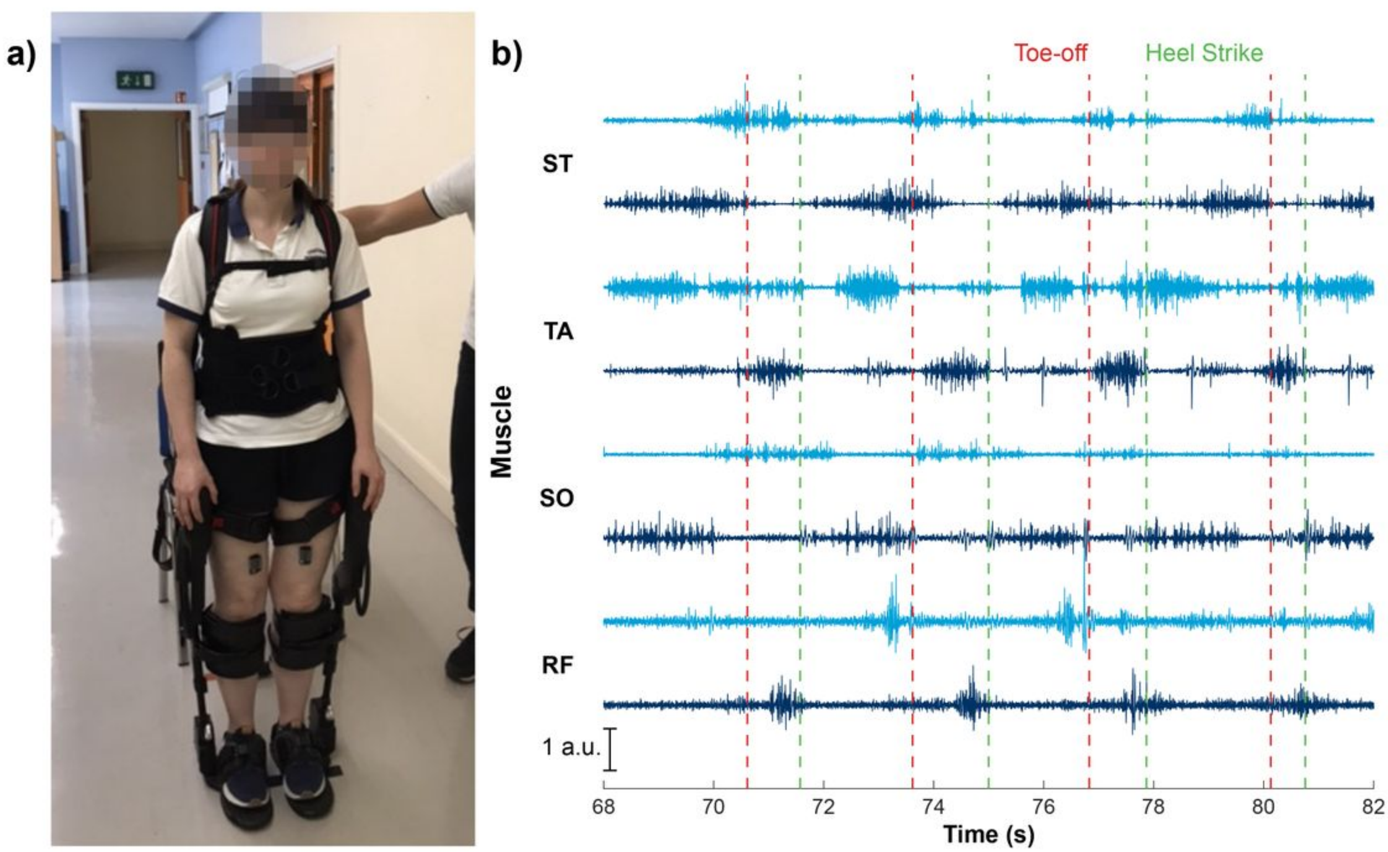

Figure 1

a) Image of a young, healthy volunteer standing in the exoskeleton. The EMG electrodes on the rectus femoris are located below the thigh straps of the device. b) Normalised, raw EMG data for a single leg of a representative stroke participant (Subject 5) during gait in the adaptive assist mode. EMG from the non-paretic side is shown in the bottom (navy) trace for each muscle, while the top (blue) trace shows EMG from the paretic side. Heel strike and toe-off events as determined using the accelerometer data are also displayed.

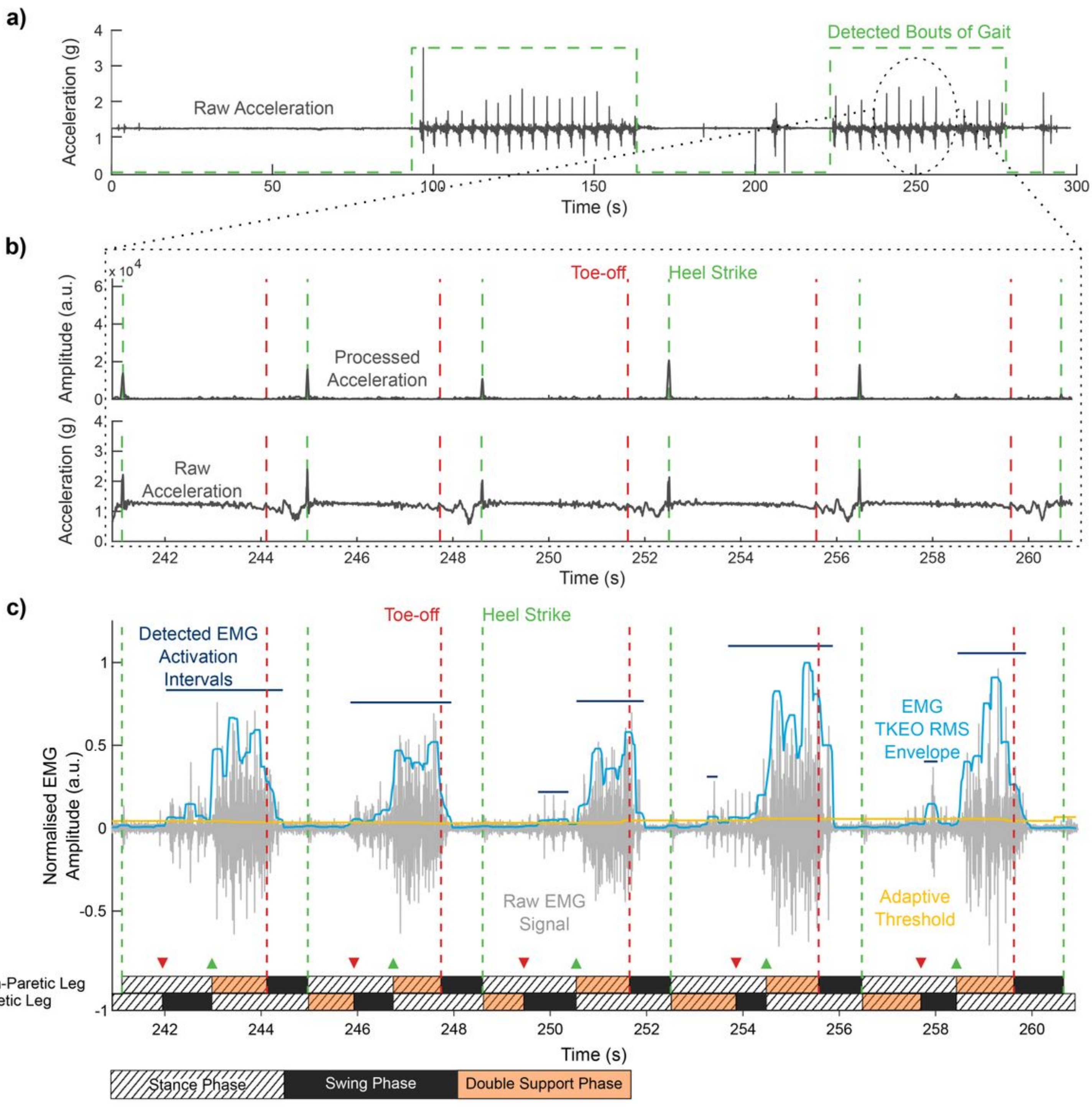

Figure 2

Signal processing steps. a) Detection of walking bouts from raw accelerometer recorded on non-paretic leg.. b) Implementation of the Teager-Kaiser Energy Operator to identify heel strike and toe-off events. c) Normalised EMG signal from the non-paretic soleus muscle for a representative subject with the normalised RMS envelope of the Teager-Kaiser Energy Operator applied to the raw EMG signal. The adaptive threshold was applied to the RMS envelope and the locations where the muscle was detected to be active are shown. Heel strike and toe-off events are displayed for both legs with the contralateral (paretic) heel strike and toe-off events indicated by the triangular markers. The gait phases for both legs are also displayed for reference.

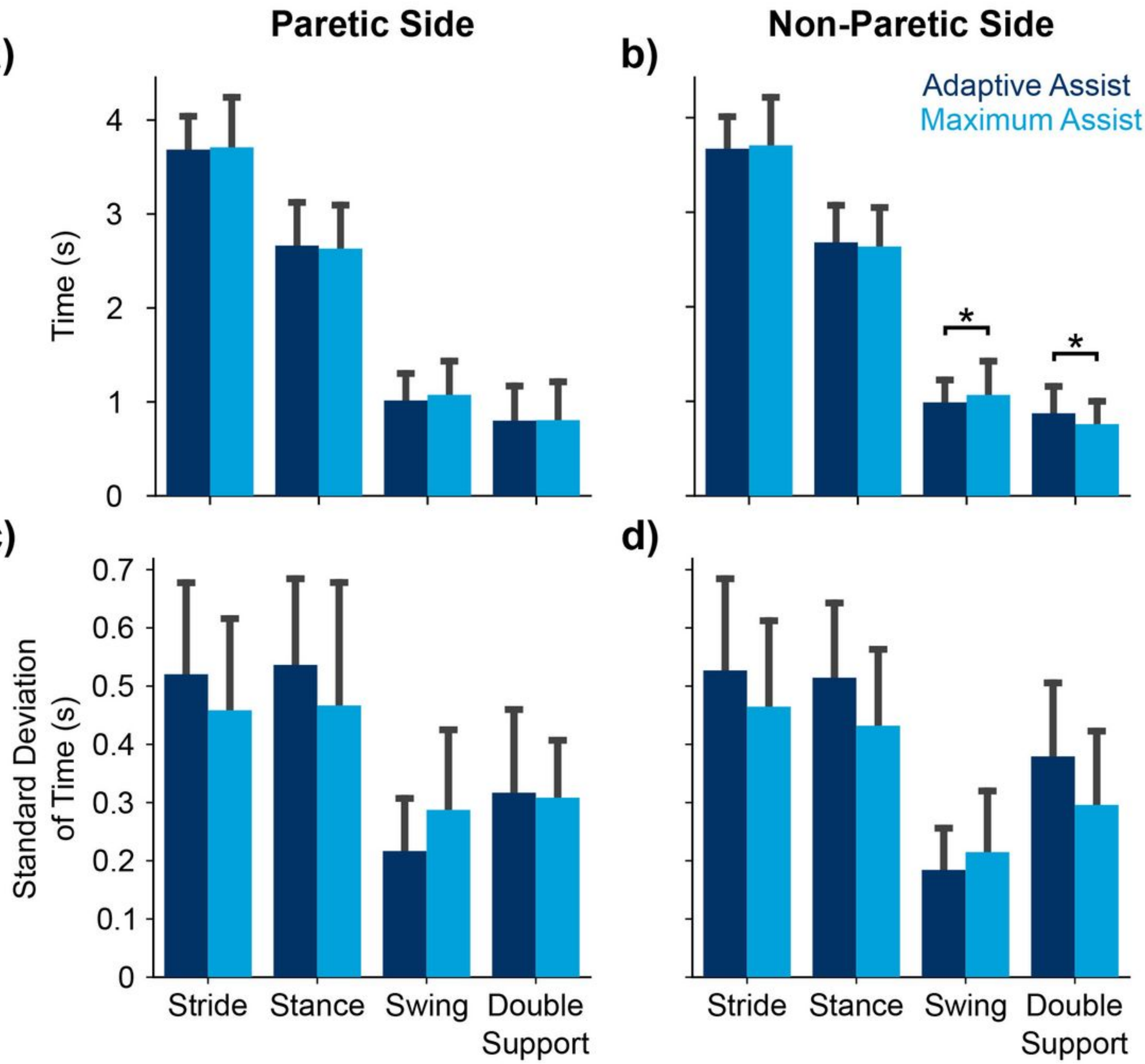

Figure 3

Mean ± standard deviation for the stride, stance, swing and double support times on the a) paretic and b) non-paretic sides. Standard deviation of the respective times are also shown for the c) paretic and d) non-paretic sides to display the variability in the accelerometer data. (* $p < 0.00625$ as indicated by the linear mixed effects model)

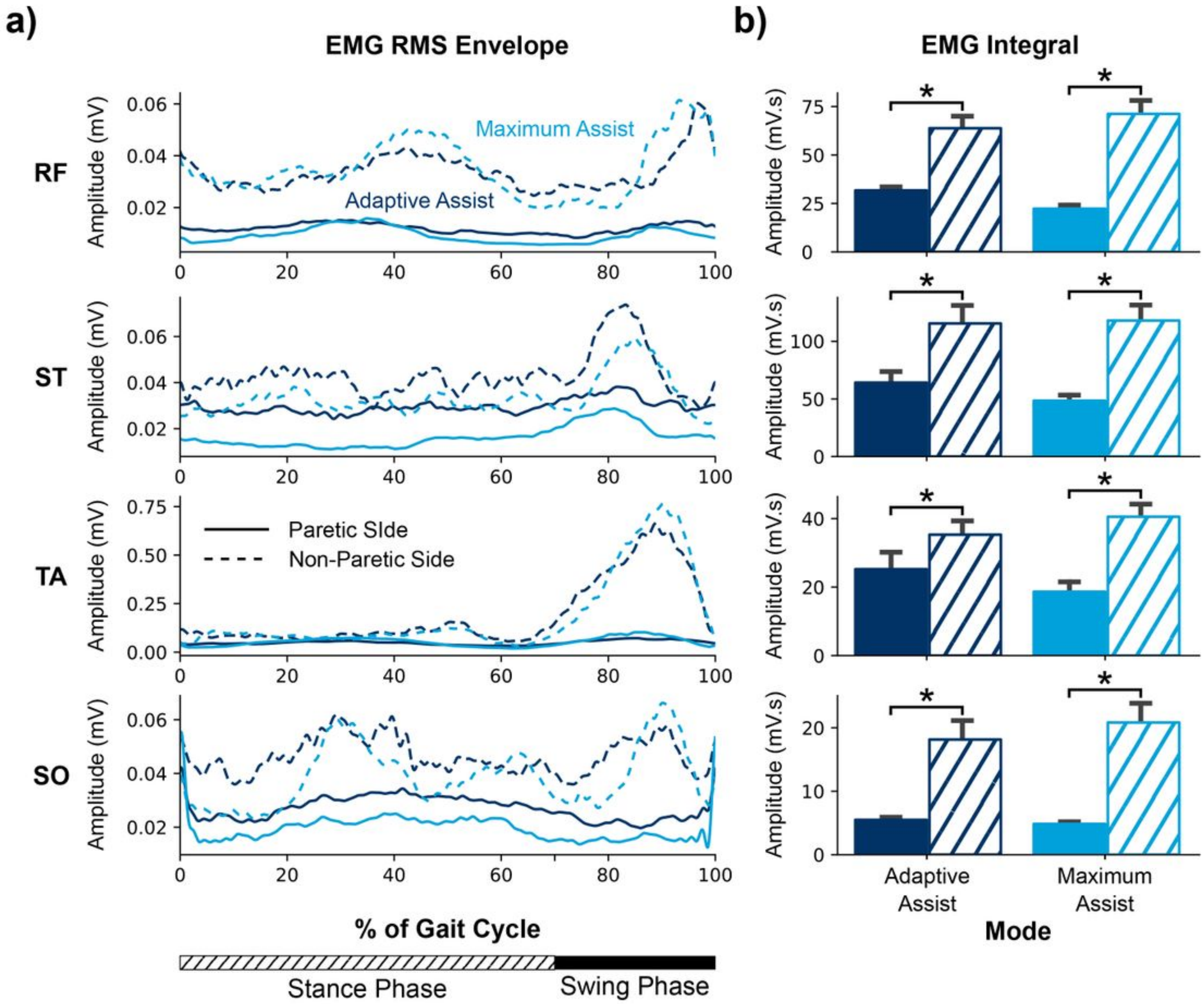

Figure 4

(a) EMG RMS envelopes for the rectus femoris (RF), semitendinosus (ST), tibialis anterior (TA) and soleus (SO) muscles for a representative subject (S1) on the paretic and non-paretic side in adaptive assist mode. Solid lines show activity during adaptive assist mode while the dashed lines show activity in maximum assist mode. (b) Mean ± standard deviation of the integral of the EMG RMS amplitude on the paretic and non-paretic side for each muscle in adaptive assist mode (* $p < 0.00125$ as indicated by the linear mixed effects model)

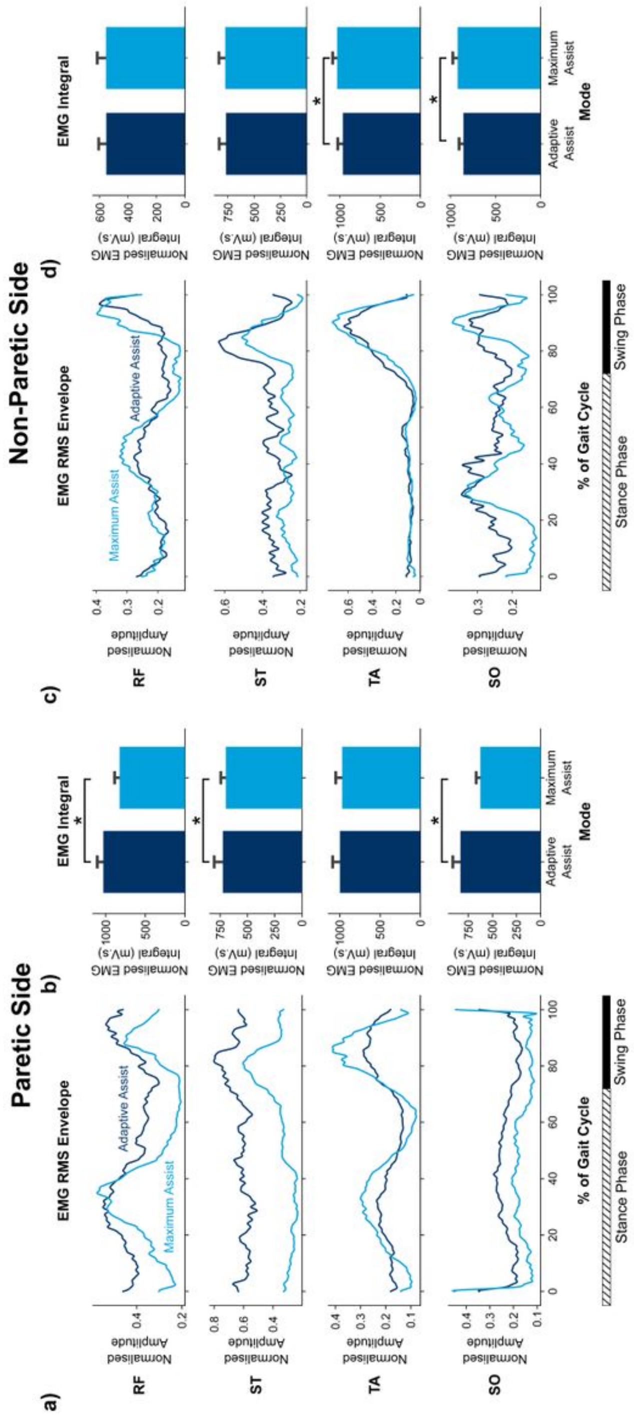

Figure 5

EMG RMS amplitude from a representative subject (Subject 1) on the paretic side (a) and non-paretic side (c) in both adaptive assist and maximum assist mode. Data have been normalised with respect to the maximum value recorded for each muscle in maximum assist mode. Mean ± standard deviation of the integral of the normalised EMG RMS amplitude in adaptive and maximum assist modes on the paretic side (b) and non-paretic side (d) across all participants when the respective muscle was considered active. (* $p < 0.00125$ as indicated by the linear mixed effects model)

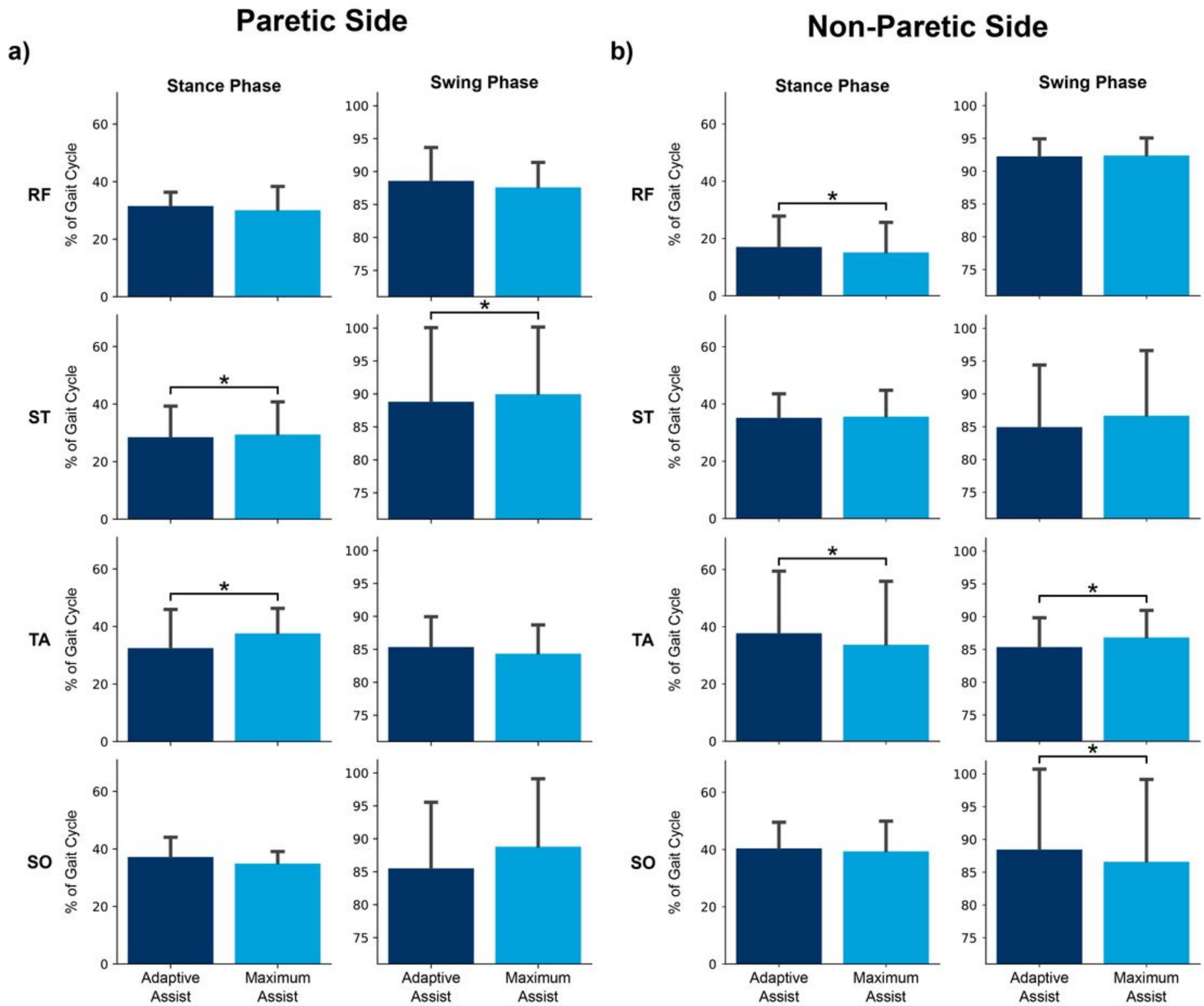

Figure 6

Mean ± standard deviation of the location of the peak EMG RMS amplitude during the stance and swing phases of gait as a percentage of the gait cycle on the paretic side (a) and non-paretic side (b) across all participants. The mean stance phase of the gait cycle for all participants in this study was 0-71% of the gait cycle, while the swing phase comprised of 71-100% of the gait cycle. (* $p < 0.00125$ as indicated by the linear mixed effects model)