Article

Relationships between Body Weight Support and Gait Speed Parameters and Muscle Activity and Torque during Robot-Assisted Gait Training in Non-Neurological Adults: A Preliminary Investigation

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Abstract: Robot-assisted gait training (RAGT) is a promising therapeutic vehicle to maximize active participation and enhance functional neuroplasticity in patients with central nervous system pathology by adequately adjusting gait speed, body weight support (BWS) level, and impedance provided by the exoskeleton. The aim of the present study was to determine the relationship between RAGT training parameters (BWS and speed) and electromyography (EMG) muscle activity torques in the knee and hip joint during RAGT. To analyze the correlation between the joint torques measured in the Walkbot gait rehabilitation system and the EMG signal of the lower limbs (vastus lateralis oblique, biceps femoris, tibialis anterior, and gastrocnemius) and understand the real-time state of the lower limb an experiment involving 20 subjects was conducted. The EMG–torque relationship was evaluated in a general rehabilitation training setting to overcome the limitations of in vivo settings. Pearson correlation coefficient analysis was performed at $p < 0.05$. Moderate relationships between biceps femoris activation data and hip and knee torques were statistically significant, ranging from $r = 0.412$ to $-0.590$, $p < 0.05$). Importantly, inverse relationships existed between hip torques and vastus lateralis oblique, biceps femoris, and tibialis anterior activation, respectively. The present results demonstrated the association between EMG locomotor control patterns and torque generation in the hip and knee joints during RAGT-treadmill under the different BWS and walking speed settings while adjusting the impedance mode parameters in non-neurological adults. Additionally, the EMG locomotor control patterns, concurrent torque generation in the hip and knee joints, and application of different BWS and walking speed parameters in the RAGT were linked to the gait speed and BWS. The outcomes also showed that the amount of BWS supplied had an impact on the effects of treadmill speed on muscle activity and temporal step control. It is essential to adjust RAGT parameters precisely in order to maximize training session efficiency and quality. The results of this study nevertheless call for more investigation into the relationship between muscle activity and torque outcomes in diseased populations with gait impairment.

Keywords: robot-assisted gait training; body weight support; gait speed; muscle activity; torque

1. Introduction

Robot-assisted gait training (RAGT) is a promising therapeutic approach to maximize active participation and enhance functional neuroplasticity in patients with central nervous system pathology by adequately adjusting gait speed, body weight support (BWS) level, and impedance provided by the exoskeleton [1–3]. The contemporary motor learning theory suggests that RAGT is most effective when variable gait parameters are modified throughout RAGT, thereby increasing appropriate muscle recruitment [4]. RAGT can
accomplish the goals of gait training, including offering support, propulsion, stability, and foot clearance, using three key parameters: BWS, treadmill speed, and impedance [5,6]. Thus, patients who are unable to voluntarily accommodate these parameters can elicit neuroplastic changes in the task-specific proprioceptive and somatosensory information necessary for gait rehabilitation as changes occur in the spinal and supra-spinal networks stimulating unaided walking [7–11].

An impedance-based control mode in the Walkbot is an interactive resistive-assistive system that enables a patient to move with or against impedance-based resistive-assistive forces [12]. For instance, the lower the impedance (e.g., 10–100%), the higher is the resistive-assistive force interactively exerted as the patient actively attempts to reach a target in ankle–knee–hip segmental kinematic and kinetic coordination, which is concurrently provided in the real-time presentation system [13]. This allows low impedance control to elicit greater active participation of patients through human–robot interactions. In contrast, high impedance control is designed to guide the lower limbs to follow a predefined trajectory.

Regarding BWS, the locomotor muscles cannot be activated properly if the patient’s weak muscles are overloaded with body weight. However, they could adequately re-learn weight-bearing steps, interlimb coordination, and foot placement with ideal muscle coordination using RAGT with BWS. Similarly, gait training at an improper speed could cause a compensatory muscle activity pattern, which hampers neural restoration [14].

In the evaluation of neurophysiological muscle activity, electromyography (EMG) is frequently used to demonstrate the results of gait studies and investigate the effects of training parameters, such as BWS, treadmill speed, and guidance force, in RAGT studies [5]. However, to date, only two studies have examined the muscle activity patterns at different impedance levels, showing that the modular output of the synergistic muscle groups is invariant over impedance levels [15] and that reductions in impedance force do not increase muscle activity [16]. Recently, van Kammen et al. (2014) [6] showed that the nature and magnitude of the differences between RAGT and treadmill walking depend on complex interactions between BWS and gait speed.

Evaluating EMG biofeedback during RAGT is important when investigating muscle activity and inspecting its active participation in neuroplasticity. However, muscle activity studies utilizing EMG are time-consuming and limited by their in vitro settings, showing poor implementation in real-time RAGT sessions. EMG could be replaced with other biomechanical parameters with similar patterns, such as kinetic (torque) variables. The Walkbot gait training system offers real-time visual and haptic biofeedback for torque, kinematics, and active and resistive stiffness of the hip, knee, and ankle joint movements during gait rehabilitation. To supplement the other parameters, the biofeedback must be accurate and reliable. Nevertheless, most recent studies on force/torque estimation were conducted on the upper limbs [17–19]. Such protocols should be developed in consideration of locomotor control and motor learning and shed light on how training parameters alter gait training task demands and gait control. However, this research has its novelty in the fact that similar studies [5,6,9] have never been carried on a Walkbot RAGT exoskeletal device. Therefore, to gain an insight into how the nature and magnitude of gait speed and BWS affect muscle activity and how this muscle activity pattern correlates with hip and knee torques, Walkbot RAGT, with varied BWS and gait speed, should be studied in non-neurological adults. The aim of the present preliminary study was to determine the optimal RAGT training parameters (BWS and treadmill speed) for EMG muscle activity and the relationship between EMG muscle activity and hip and knee torques during RAGT in non-neurological adults. The hypothesis was that lower-limb EMG muscle activities including vastus lateralis oblique, biceps femoris, tibialis anterior, and gastrocnemius would be correlated with hip and knee torques during RAGT. Clinically, this preliminary information would serve the baseline normative data for clinicians when utilizing RAGT for locomotor recovery in individuals with neurological impairments or conditions.
2. Methods

2.1. Gait Rehabilitation Robot

The experiments were performed using the gait rehabilitation robot Walkbot_G (P&S Mechanics, Seoul, South Korea). The robot automates BWS treadmill training in patients with locomotor dysfunction in the lower limbs, such as spinal cord injury and hemiplegia, after stroke [20,21]. The Walkbot exoskeleton has three actuated orthoses attached to the participant's lower limbs with cuffs and straps. The hip, knee, and ankle joints of the Walkbot are actuated by linear drives that move orthoses through the gait cycle in the sagittal plane [22]. The knee and hip torques can be determined using the force sensors between the actuators and orthoses [12]. Spring-loaded cloth straps are added to induce ankle dorsiflexion during the swing phase. An external BWS system can relieve a definite percentage of body weight using a harness. The monitor in front of the participants offers real-time visual feedback for active and resistive torques and kinematics for the hip and knee joints during gait training.

2.2. Surface Electromyography (EMG)

Muscle activity was measured in bilateral lower limbs using EMG electrodes attached to the thigh (vastus lateralis obliquus [VLO] and biceps femoris [BF]) and lower leg (tibialis anterior [TA] and gastrocnemius medialis [GCM]). The same person obtained all EMG measurements. The skin was shaved, cleaned, and rubbed with an abrasive gel to ensure good signal conduction. Eight self-adhesive Ag/AgCl dual-snap electrodes (10 mm diameter each and 20 mm inter-electrode distance, Noraxon Inc., Scottsdale, AZ, USA) were placed according to SENIAM (surface EMG for a non-invasive assessment of muscles) recommendations [23]. An assistant observed the quality of the EMG signals throughout and immediately eliminated sources of movement artifacts (owing to interference with the harness, contacting cables, or dropping electrodes). The EMG signal was recorded (sampling rate: 1500 Hz) with a wireless TeleMyo DTS system (Noraxon Inc., Scottsdale, AZ, USA; common-mode rejection ratio >100 dB; 10 Hz first-order high-pass hardware filter) and the corresponding software applications MyoResearch 3 and myoMUSCLE (Noraxon Inc., Scottsdale, AZ, USA). The system was time-synchronized with a custom device (16 MHz clock speed) paired via Bluetooth v2.0 with Walkbot_G to trigger a start signal in the EMG software (Figure 1). The EMG set up is shown in Figure 2.

Figure 1. Schematic system configuration for synchronization between TeleMyoDTS and WALKBOT_G.
2.3. Statistical Analysis

The Pearson correlation coefficient was used to evaluate the torque–EMG relationship. Statistical significance was set at \( p < 0.05 \). All data were analyzed using the Statistical Package for the Social Sciences version 25.0.

2.4. Participants

A convenience sample of 20 non-neurological adults (9 females, mean age = 34.7; SD = 12) were recruited from a local community and university settings in Korea. According to power analysis using G*power software (Franz Faul, Kiel, Germany), the total calculated sample size was 18 (\( \alpha \) error probability, 0.05; 1-\( \beta \) error probability, 0.95; correlation \( \rho \) \( H_0 \), 0). The sample size was computed, considering potential drop-out rate of 10 percent, 20 community-dwelling individuals, including 10 (50%) women and 10 (50%) men, with a mean age of 35 years, were included in this study given the result of power analysis. All participants were healthy with no history of lower limb trauma or serious muscular, neurological, cardiovascular, metabolic, or inflammatory diseases. All participants were able-bodied adults who could walk independently without an assistive device. Table 1 shows the baseline characteristics of the study participants. This study followed the guidelines laid down by the Declaration of Helsinki which was approved by Institutional Review Board of Gwangju Institute of Science and Technology (20210416-HR-60-04-0, Approved on 29 April 2021). All participants provided informed consent before participating in the study. Because many anthropometrical and demographical factors, such as height, weight, body mass index, activity level, profession, and origin could affect gait patterns, we have attempted to include a relatively homogeneous ethnicity and representative sample.

Table 1. Personal anthropometric information.

| Participant ID | Sex | Age (Years) | Height (m) | Weight (kg) | BMI  | Activity Level * | Profession      |
|----------------|-----|-------------|------------|-------------|------|-----------------|-----------------|
| SUB 00         | M   | 29          | 1.76       | 86          | 27.8 | Vigorous        | student         |
| SUB 01         | M   | 28          | 1.88       | 88          | 24.9 | Vigorous        | student         |
| SUB 02         | M   | 59          | 1.66       | 60          | 21.8 | Moderate        | office worker   |
| SUB 03         | M   | 52          | 1.83       | 82          | 24.5 | Moderate        | office worker   |
| SUB 04         | M   | 29          | 1.72       | 77          | 26.0 | Vigorous        | student         |
| SUB 05         | M   | 48          | 1.74       | 63          | 20.8 | Moderate        | office worker   |
| SUB 06         | M   | 45          | 1.75       | 65          | 21.2 | Vigorous        | office worker   |
Table 1. Cont.

| Participant ID | Sex | Age (Years) | Height (m) | Weight (kg) | BMI   | Activity Level * | Profession              |
|----------------|-----|-------------|------------|-------------|-------|------------------|-------------------------|
| SUB 07         | F   | 28          | 1.55       | 61          | 25.4  | Vigorous         | student                 |
| SUB 08         | M   | 38          | 1.81       | 120         | 36.6  | Vigorous         | office worker           |
| SUB 09         | M   | 37          | 1.71       | 80          | 41.0  | Vigorous         | office worker           |
| SUB 10         | F   | 27          | 1.63       | 60          | 22.6  | Vigorous         | office worker           |
| SUB 11         | F   | 58          | 1.55       | 62          | 25.8  | Moderate         | activity assistant teacher |
| SUB 12         | F   | 26          | 1.63       | 54          | 20.3  | Moderate         | student                 |
| SUB 13         | F   | 26          | 1.63       | 55          | 20.7  | Moderate         | student                 |
| SUB 14         | F   | 27          | 1.63       | 55          | 20.7  | Vigorous         | student                 |
| SUB 15         | F   | 20          | 1.7        | 55          | 19.0  | Moderate         | student                 |
| SUB 16         | M   | 26          | 1.74       | 61          | 20.2  | Moderate         | student                 |
| SUB 17         | F   | 22          | 1.66       | 63          | 22.9  | Moderate         | student                 |

M, male; F, female; BMI, body mass index; * The following is in accordance with CDC and ACSM guidelines.

2.5. Experiment Setup

The torque–EMG relationship was evaluated in a general rehabilitation training setting to overcome the limitations of in vivo settings. When experimental data are obtained by adhering to one training condition, data bias may occur. Therefore, we set 7 various training conditions by referring to the actual conditions used when operating the gait rehabilitation system. The gait speed and BWS parameters were adjusted for various set values of the gait rehabilitation robot. The two set values are settings that are adjusted in actual gait training such that the rehabilitation is suitable for the patient. Therefore, various rehabilitative training conditions were created by modifying the corresponding values. Thus, the participants underwent the experiments accordingly under nine gait training conditions. As shown in Table 2, three sets of independent experiment conditions were implemented by adjusting walking speeds (1, 1.5, and 2 km/h), and BWS conditions (0%, 30%, and 60%). The order of the experimental conditions was counterbalanced to minimize the potential confounding ordering effects [24]. The maximal gait speed was automatically adjusted by RAGT based on each subject’s anthropometric data entered. The set value of BWS indicated the ratio of support to the participant’s weight and had no upper or lower limit. The order of training conditions was randomly assigned to each participant. Impedance was set at the lowest level of 10%.

Table 2. Experiment settings according to the gait speed and body weight support.

| Experiment Conditions | Body Weight Supports (%) | Gait Speeds (km/h) |
|-----------------------|--------------------------|--------------------|
| 1                     | 0                        | 1                  |
| 2                     | 0                        | 1.5                |
| 3                     | 0                        | 2                  |
| 4                     | 30                       | 1                  |
| 5                     | 30                       | 1.5                |
| 6                     | 30                       | 2                  |
| 7                     | 60                       | 1                  |
| 8                     | 60                       | 1.5                |
| 9                     | 60                       | 2                  |

2.6. Experiment Procedure

The height, weight, age, and leg length of the participants were obtained. Based on the acquired data, a gait rehabilitation system was established to fit each participant. The maximum gait speed was calculated, and the experimental conditions that the participant could perform under were determined. Eight EMG sensors were attached to the participants’ leg at four aforementioned sites on each leg. They were then asked to walk as if they were walking on the ground, and a familiarization session was performed to
enable them to create a natural gait pattern for themselves. Subsequently, the participant proceeded with the experiment in a randomized order. Each time a new experimental condition was applied, participants took approximately 2 min to familiarize themselves with it. After familiarization, data were measured while walking for 2 min. Because the number of measured gait cycles varied with the gait speed, an average of 30–40 gait cycles were measured. Among the data measured for each condition, 30 middle gait cycle data points were selected and used for further analyses. Flow chart of the experimental design is demonstrated in Figure 3.

Figure 3. Flow chart showing the experimental design of the study.

2.7. Data Collection and Processing

A 20 Hz Butterworth high-pass filter was applied to the raw signal to eliminate artifacts. Additionally, the signals were rectified and smoothed by the root mean square (RMS) with a time window of 100 ms. The EMG signal is a measure of the actual muscle voltage, and because the maximum voltage that can be generated varies across individuals, it should be normalized. The most common methods are maximum voluntary isometric contraction (MVIC) and reference voluntary contraction (RVC). In both methods, the reference value is the first set, and the percentage of that value is displayed. The MVIC is based on the RMS
when the participant applies the maximum isometric contraction to the muscle, whereas RVC is based on the RMS when a specific movement is performed. In this study, the EMG signals were normalized using MVIC. To avoid confusion, the “normalized EMG signal” is hereafter referred to as “muscle activity.”

No separate filter was applied to the torque of the gait rehabilitation system. For torque, the range of strength that can be generated by each person is different; therefore, normalization is necessary, as in the raw EMG signal. However, there is no consensus on the method of normalizing strength or torque measurements [6]. Generally, measures related to body size, such as body mass (BM) and body mass index (BMI), are indices used to normalize the strength of the lower limb. Therefore, they were used to normalize the joint torque in this study. The torque was divided by BM and BMI to obtain two types of normalized torques.

Figure 4 shows the data collected under all experimental conditions for all participants. Therefore, the analysis method can be applied to organized data to determine the correlation.

Figure 4. Example of EMG and torque signals corresponding to one gait cycle.

2.8. Correlation Analysis by EMG Window Criteria for Muscle Activity

The muscles used for gait were not activated evenly throughout the gait period, and the activation time for each muscle was determined. Therefore, for a more accurate analysis, muscle activity was obtained by dividing the muscle activity window rather than by simply dividing it into the swing and stance phases. According to these criteria, the muscle activity-related indicators were set as follows [5]:

1. Thigh
   a. WINDOW_VLO: Mean activity of the VLO in the EMG activation window (0–23% and 95–100%)
   b. WINDOW_BF: Mean activity of BF in the EMG activation window (0–19% and 82–100%)

2. Lower leg
   a. WINDOW_TA: Mean activity of the TA in the EMG activation window (0–16%, 67–85%, and 95–100%)
b. WINDOW_GCM: Mean activity of the GCM in the EMG activation window (10–58%)

3. Results

The final EMG data of 18 out of 20 participants were used for further statistical analysis because of incomplete experimental data.

Relationship between EMG and Torque Data during Different RAGT Gait Speed and BWS Conditions

In the Spearman’s rho analysis of the relationship between the EMG signal and torque of the lower limb muscles (Table 3), all data were statistically significant ($p < 0.05$). Most EMG (VLO, BF, and TA) and hip torque correlations showed negative relationships. In contrast, most knee torques and EMG yielded positive relationships. The BF showed the strongest correlation ($r = -0.590, 0.412, -0.412, and 0.241$) with the hip and knee torque data in the stance and swing phases. The TA also showed a relatively moderate correlation with hip torque during the swing phase ($r = -0.240$).

Table 3. Correlation between electromyography and torque data during different robot-assisted gait training conditions.

| Spearman’s rho ($r$) | VLO   | BF     | TA     | GCM   |
|----------------------|-------|--------|--------|-------|
| **Experimental conditions** | **−0.383** | **−0.190** | **−0.072** | **−0.243** |
| **STANCE hip torque**  | −0.108 | **0.412** | −0.168 | 0.116 |
| **STANCE knee torque** | −0.211 | **−0.590** | −0.335 | **0.069** |
| **SWING hip torque**   | −0.214 | **−0.412** | −0.240 | **0.119** |
| **SWING knee torque**  | **0.180** | **0.241** | **0.136** | **0.120** |

** $p < 0.01$, * $p < 0.05$. VLO, vastus lateralis oblique; BF, biceps femoris; TA, tibialis anterior; GCM, gastrocnemius medialis.

4. Discussion

The current preliminary research investigated the relationship between muscle activity and hip and knee torque data during stance and swing phases in non-symptomatic adults. As hypothesized, our results demonstrated that VLO, BF, TA, and GCM muscle activities were related to hip and knee torques. Most importantly, the present baseline normative data would provide insightful information to clinicians when treating gait impairments utilizing RAGT in individuals with neurological impairments or conditions because our novel results revealed optimal or normative (physiological) motor control patterns elicited during Walkbot RAGT by means of adjusting gait speed and BWS parameters.

Moderate relationships between hip and knee torques and biceps femoris activation data were statistically significant, ranging from $r = 0.241$ to $−0.590, p < 0.05$. Importantly, inverse relationships between hip torques and vastus lateralis oblique, biceps femoris, and tibialis anterior activation data correlations existed, respectively. Specifically, the correlation analysis showed a moderate relationship of the VLO with gait speed and BWS ($r = −0.383$). The current results demonstrated that BF EMG activity was related to gait speed and BWS. In a previous study involving 35 healthy women, a moderate ($r = 0.50$) correlation was reported between BF EMG muscle activity and gait speed [25]. Gait phase prolongation may have enforced modifications in neuromuscular control to accommodate the altered task constraints implied by changing the natural relationship between gait speed and BF muscle activity [25]. Another study on the intentional modification of the preferred step-frequency-to-amplitude relationship suggested that such alterations in the temporal step pattern are accompanied by reorganization of the underlying muscle activity, including the quadriceps and hamstrings [6]. Gizzi et al. (2012) [26] showed that modular organization of neuromuscular activity characteristics to bipedal human locomotion was compatible between the overground walking and RAGT conditions and
largely unaffected by changes in gait speed and BWS. Interestingly, the overall temporal patterning of muscle activity was maintained regardless of the different BWS and gait speed conditions, which supports the role of the central pattern generator in RAGT as evidenced in EMG muscle activation patterns [26]. Similarly, Wan and colleagues reported moderate ($r = 0.48$) to strong ($r = 0.71$) correlations between the hamstring muscle activity and torque at different positions in 19 healthy adults [25] because, neuromechanically, hamstrings involve in the hip and knee torques as a two-joint muscle. A possible rationale for these results is related to the neurophysiological effect of decreasing the EMG activity at greater muscle length, together with the mechanical effect of increased force by passive parallel elastic components; this may account for the constant torque and is associated with EMG activity of the lengthened muscle. A previous study reported that promoting early task-specific RAGT with bottom–up integration could improve gait recovery in patients with neurological disorders [2,27]. Additionally, our previous evidence demonstrated that altered EMG activities of lower limbs including TA and GCM in hemiparetic stroke and cerebral palsy were more normalized with the immediate application of RAGT as compared to the normative data [28,29].

Although the observed EMG patterns displayed an increase in amplitude that typically accompanies increases in gait speed for some muscles (VLO and GCM), speed effects were attenuated when BWS increased [30–32]. During the body weight unsupported gait training, increases in gait speed are accompanied by larger impact forces at foot contact, which should be actively counteracted by generating additional ground reaction force, thereby further activating corresponding lower leg muscle groups [33]. In accordance with previous studies [6], leg unloading through BWS reduced the EMG amplitude during the stance phase in the VLO and GCM muscles. In contrast, the amplitude of muscle activity increased in the BF when BWS was provided, particularly at lower speeds. It is crucial to understand how parameter settings affect locomotor control for a purposeful exploitation of training parameters in the Walkbot RAGT.

Taken together, our EMG locomotor control and torque data suggest that optimal, normative locomotor control patterns were associated with hip and knee torque production. Moreover, the BWS and walking speed tend to relate to the EMG and torque production, indicating that lower BWS and increased walking speed favorably generated more normalized locomotor control patterns and force production.

Limitations

The present preliminary research poses a few shortcomings which should be considered in future. The first drawback is that some of the relationships were quite weak. Even though they were statistically significant, they did not fully account for the variation in the connections between the variables. Another limitation is that only healthy adults were included; hence, we should be careful when generalizing the current data into pathological populations due to different neuromechanical characteristics [32]. Nevertheless, the present normative data would render important information to clinicians when mitigating gait impairments utilizing RAGT in individuals with neurological impairments. The other limitation is that the sample size may be an issue, although EMG and torque data were shown to be clinically useful during variant RAGT’s BWS and walking speed conditions.

5. Conclusions

The present results highlighted the relationship between the EMG locomotor control patterns and torque generation in hip and knee joints during RAGT-treadmill under the variant BWS and walking speed conditions while controlling the impedance mode. Furthermore, the application of variant BWS and walking speed parameters in the RAGT, the EMG locomotor control patterns and concurrent torque generation in hip and knee joints were associated with the gait speed and BWS. The results also demonstrated that the effects of treadmill speed on muscle activity and temporal step control depended on the amount of BWS provided. The specific settings of RAGT parameters are crucial in order to maximize
effectiveness and quality of training sessions. Nevertheless, the outcomes of this study warrant further research on the effects of RAGT parameters on muscle activity and torque outcomes in pathological population with gait dysfunction.

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