THE EFFECT OF ASSISTIVE FORCE ON THE AGONIST AND ANTAGONIST MUSCLES IN ELBOW FLEXION

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Abstract

This study investigated the responses of the agonist and antagonist muscles against assistive force during isometric muscle contraction. Participants performed isometric elbow flexion at 90º for 30 seconds under two workload conditions (20% and 40% of the maximal voluntary workload) with three levels of assistive force (0%, 50%, and 100% theoretical effectiveness) for 10 seconds. Electromyography (EMG) of the biceps (agonist muscle) and triceps (antagonist muscle) was measured during the task, and perceived exertion was obtained after the task. Assistive force significantly reduced EMG activity in the agonist muscle and the perceived exertion score only at 40% workload. However, the reduction of EMG activity and perceived exertion score were lower than that for the physical estimated effect. In addition, the EMG activity in the antagonist muscle was not influenced, irrespective of workload conditions and the level of assistive force. These results suggested that although the assistive force during isometric muscle contraction relieves the exertion of the agonist muscle that accompanies the decrease in perceived exertion, their assistive effects are influenced by various human physiological and anatomical factors.

Introduction

Ageing leads to a decline in physical strength and health (Kirkendall & Garrett, 1998; Janssen, Heymsfield, & Ross, 2002; Mayer et al., 2011). These declines may cause physical problems in elderly people and affect their ability to perform daily activities such as walking, bathing, and performing house chores. In order to improve the quality of life in elderly and physically incapable individuals, it is important to assist them to be more active and independent, as stated by Rikli (2005). Given this background, technology, namely assistive technology (AT), has been rapidly developed to assist these individuals (Araullo & Potter; Yusif, Soar, & Hafeez-Baig, 2016). These technologies are expected to be useful in various fields to reduce physical burden (Zhuang, Stobbe, Hsiao, Collins, & Hobbs, 1998) and prevent musculoskeletal disorders (Edlich, Winters, Hudson, Britt, & Long, 2004; Fujishiro, Weaver, Heaney, Hamrick, & Marras, 2005) in elderly and physically infirm people.

AT is defined as any device or system that allows individuals to perform physically difficult tasks, or increases the ease of use and safety in performing tasks (McCreadie & Tinker, 2005). AT devices have been examined from biomechanical, haptic, electrical, and mechanical perspectives to achieve function. Devices to assist the motion of humans with deteriorated muscle and joint capability have developed remarkably, especially in recent years. For example, the ED Walker is a walking aid device that implements a biomechanics system to transfer body weight through the pelvic girdle (Clark, Manini, Ordway, & Ploutz-Snyder, 2004). General Motors and National Air and Space Administration collaborated to develop the Robo-Glove for automotive factory operators. This device enables them to hold heavier objects and manipulate tools for longer periods comfortably. These kinds of devices aim to enhance motion capability and reduce the physical efforts of users.

Some previous studies have proven the effectiveness of assistive devices using the EMG signal of the assisted working muscles. For instance, Miyawaki et al. (2009) reported that the weight support mechanism of spring compression led to 28% and 35% decrease in % integral EMG (%IEMG) of the biceps brachii (BB) and triceps brachii (TB) muscles during elbow extension and flexion, respectively. Clark et al. (2004) observed that increasing the bearing load in an ambulatory device reduced the activity of the leg extensor muscles and provided assistance during walking. Bellar, LeBlanc, and Judge (2014) reported reduction of the biceps femoris and the semitendinosus seen on EMG during an isometric knee extension exercise while participants wore the elastic hamstring assistance device.

However, humans have various functions against external stimuli and stressors such as self-defence and adaptation (St Clair Gibson, Lambert, & Noakes, 2001; Mottram, Maluf, Stephenson, Anderson, & Enoka, 2006). Accordingly, the effectiveness of an assistive force on humans could not be predicted only from the engineering aspect. Furthermore, while assistive force is applied, physical output against motion could be different from that which was expected, based on the feedback of the muscle tension itself and the user's own experiences. This gap might disturb the motor control system. In order to develop motion AT that is preferable for use in humans, specific human physiological responses to external assistive force must be strongly considered.

Against this background, the present study investigated muscle activity against external assistive force at various force levels while a voluntary muscle exercise was performed at various workloads. Simple voluntary exertion (namely, isometric elbow flexion) was investigated as the first step to understanding transient human physiological adaptability against assistive force. This type of exercise can exclude various factors such as joint angle, joint angular velocity and mechanical loss to interpret adaptability, and can identify the main agonist and antagonist muscles (the biceps and triceps, respectively). Accordingly, it is expected to help determine the specific human physiological responses to assistive force. We hypothesised that although the assistive force reduces muscle exertion in both the agonist and antagonist muscles, their reduction is smaller than estimated.

Methods

Participants

Twenty-five young male university students were recruited for the present study (Table 1). All participants were between 20 and 35 years old, and all were right-handed. Right-handedness was confirmed with the Edinburgh Handedness Inventory (Oldfield, 1971). None of the participants reported any arm injuries or musculoskeletal disorders within the past 6 months and were mentally and physically healthy. This study was approved by the Ethics Committee of the Faculty of Design, Kyushu University, Japan, and written consent was obtained from each participant.

Table 1: Characteristics of the participants (n=25)

| Variables	Mean	±	SD
Age (years)	25.1	±	3.5
Body height (cm)	170.9	±	4.8
Body weight (kg)	66.5	±	11.3
Radius length (cm)	26.0	±	0.9

SD: Standard deviation

Experiment set-up

The experiment set-up was designed as shown in Figure 1. Participants sat on an armless chair with their body straightened, while the right arm was positioned at 90° elbow flexion and the palm was placed in the supine position (test posture). A strap was secured at the participant's wrist to connect his forearm to weights that acted as workload. The workload was adjusted for each participant to match the capabilities of the muscle's isometric contraction.

Another strap was firmly secured at the middle of the length of the radius to support the forearm and connect the forearm to the assist load at the other end of the fixed pulley system. As shown in Figure 2, a force that pulled the forearm up was transferred from the gravity of the assist load to the forearm by the fixed pulley system. We believed that assistive force should reduce the effort of the targeted muscles (biceps and triceps). In

addition, a barrier was placed between the participants and the assist load to keep the participant from anticipating the degree of assistance.
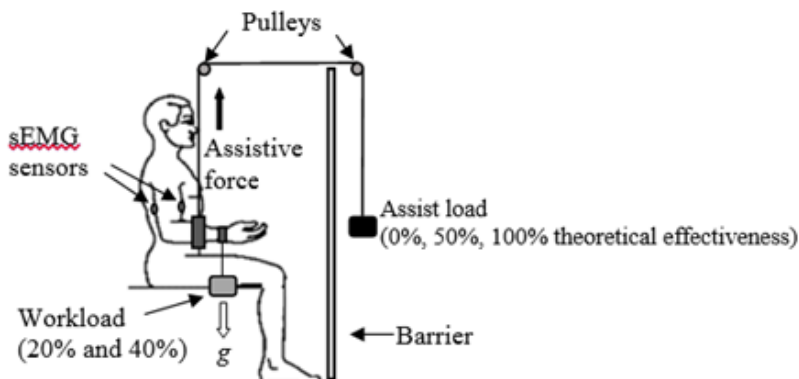
Figure 1: Experimental set-up for isometric elbow flexion and application of assistive force
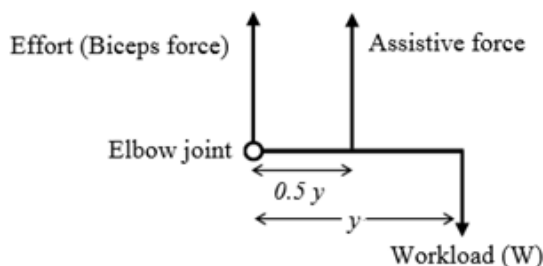
Figure 2: Free body diagram of the elbow joint and related forces that acted on the joint (y is the radius bone length)

Experimental conditions

First, the maximum workload of isometric elbow flexion was identified for each participant. Based on the maximum workload, 20% and 40% of the maximum workload (submaximal workloads) and three assisting load conditions (0%, 50% and 100%) based on the submaximal loads were determined. Next, each participant performed six trials that consisted of the mentioned workload and assistive force in counterbalanced order.

Procedure

All experiments were performed at room temperature (about 22°C). Participants were asked to wear a short-sleeved shirt. One day prior to the experiment, they were requested not to perform any exercise that may cause arm muscle fatigue or pain, and thus affect physical performance during the experiment. The maximal isometric voluntary contraction (MVC) was obtained before the experimental tasks started. The MVC was used to normalise the EMG muscle activity during the exercise at submaximal workloads. For the BB, the participants were asked to sit straight on an adjustable stool with their right hand in a supinated forearm posture at 90° of elbow flexion and the wrist located under a table.

Next, they were asked to lift their wrist up against the fixed bottom surface of the table at maximum effort for 10 seconds. As for the TB, participants were asked to maintain the same posture with their hands and wrists on the table. Participants then pushed down against the top of the table at maximum effort. This procedure was repeated three times for both muscles. A rest time of at least 60 seconds was provided for the participants to relax after completing each trial.

Then, the MVC during isometric elbow flexion at maximum loads was identified. The maximum workloads were exerted against the wrist, starting with 4 kg and then gradually adding 0.2 kg, 0.1 kg, and 0.05 kg until participants claimed that the workload met their expected maximal effort. If the participant's upper limb shook considerably and he was unable to maintain the 90° elbow flexion position, this indicated that the participant had met their maximal effort. The mean and standard deviation of the maximum load for all participants was 6.66 ± 1.15 kg, ranging between 4.7 kg and 8.2 kg. Two submaximal workloads (20% and 40%) based on the maximal workload were then determined.

In the experimental trial, participants were requested to maintain the 90° elbow flexion position for 30 seconds. Between 0 and 10 seconds, no assistive force was applied; between 11 and 21 seconds, assistive force was applied at three levels of theoretical effectiveness; and between 22 and 30 seconds, the assistive force was removed (Figure 3). The EMG signals at the 10th and 11th seconds and 21st and 22nd seconds were not included in the data analysis because these were transition periods between levels of assistive force. A minimum rest period of 60 seconds was provided after each trial was finished. Each participant was required to perform a set of six trials (20% workload with 0%, 50%, and 100% assist load and 40% workload with 0%, 50%, and 100% assist load). The order of trials was randomised.
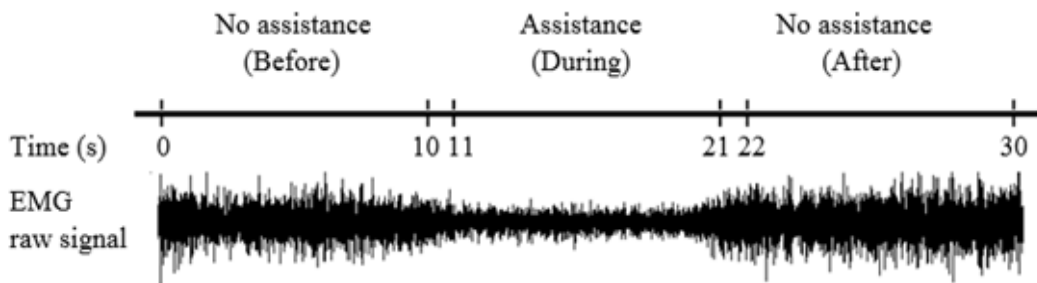
Figure 3: Timing for assistive load application for one trial and the related EMG raw signal pattern

Measurements

Surface EMG (sEMG) of the BB and TB was recorded in real time using a multichannel telemetry system (WEB-7000, Nihon Kohden, Tokyo, Japan) at a sampling frequency of 1 kHz with a band-pass filter of 15-500 Hz and built-in electrode of ZB-150H. Three EMG sensors were firmly attached to the participants' right arms to measure BB (A1), TB (A2), and the wrist flexor (WF) (A3). The EMG signal of the WF muscle was measured to ensure that participants used the BB muscle more often than the WF in the trial. The sensors were

attached to the skin by using disposable adhesive tapes. Participants practised the isometric contraction for both muscles before the actual experiment started, and palpation was done to identify the locations of the sensors. Based on the palpation, the sensors were located at the middle of the muscle belly as stated by Gerachshenko and Stinear (2007). The participant's skin was cleaned with alcohol before the sensors were attached. In addition, a goniometer was attached to the elbow to monitor the 90° elbow flexion position. For MVC, measurements were taken for 8 seconds in each trial, whereas the duration of the EMG recording for the trial was 30 seconds. The EMG raw data were imported, synchronised, and analysed using KineAnalyzer software (Kissei Comtec, Nagano, Japan). All signals were rectified with full waves. The EMG amplitude was normalised with respect to the MVC, which was measured before the trials. The mean values of the EMG before, during, and after the assist load application periods (Figure 3) were calculated, and the percentage of MVC was determined as:

$$\text{Mean EMG} = \frac{Mean\ EMG_{exp}}{Mean\ EMG_{MVC}}$$

The perceived exertion during the exercises was subjectively analysed using Borg's CR-10 Scale (Borg, 1982; Shibata, Takizawa, & Mizuno, 2015). The Borg Scale consists of rating values that range between 0 and 10, which reflects the perceived level of exertion: 0 indicates no feeling of muscle effort and 10 indicates maximal effort. Participants rated the exertion immediately after each trial was completed.

Statistical analysis

IBM SPSS (version 22.0 software, Cary, NC, USA) was used for the statistical analysis. All data are presented as the mean ± standard deviation. Two-way repeated measure analysis of variance (2 x 3 factorial) was conducted to evaluate the influences of the levels of workload (20% and 40%) and assist load (0%, 50%, and 100%). The assumption of sphericity was violated, as indicated by Mauchly's test; thus, the Greenhouse Geisser correction was used in the analysis of variance (Field, 2013). A post-hoc pair-wise Bonferroni-corrected comparison was used to examine the mean differences in the factors. Rated perceived exertion (RPE) and workload levels were analysed using a non-parametric Wilcoxon signed-ranks test. Statistical significance was accepted at $p < 0.05$.

Results

Muscle activity of the BB and TB

The relationship between workload and assistive level and %MVC is shown in Figure 4(a) for the BB and Figure 4(b) for the TB. For the BB, the analysis of variance showed that both workload (F [1,24] = 65.67, p<0.001) and assistive level (F[1.22, 29.32] = 13.16, p<0.01) had a significant effect and a significant interaction (F [1.68, 40.27] = 12.53, p<0.001). Post hoc comparisons indicated that there was a significance difference among assistive levels in the 20% workload condition. In contrast, at 40% workload, the 50% and 100% assistive level conditions showed significantly lower %MVC than 0% assistance.

This can be seen in the Figure 4(a) for the same assistance condition; the reduction in %MVC value at 40% workload was larger than that at 20% workload.

Contrary to BB, in TB, workload (F [1, 24] = 4.98, p<0.05) and assistive level (F [1.36, 32.58] = 5.06, p<0.05) had significant main effects, but no interaction between workload and assistance was found (Figure 4(b)). The post hoc comparison did not suggest that there were significant differences among assistive levels.
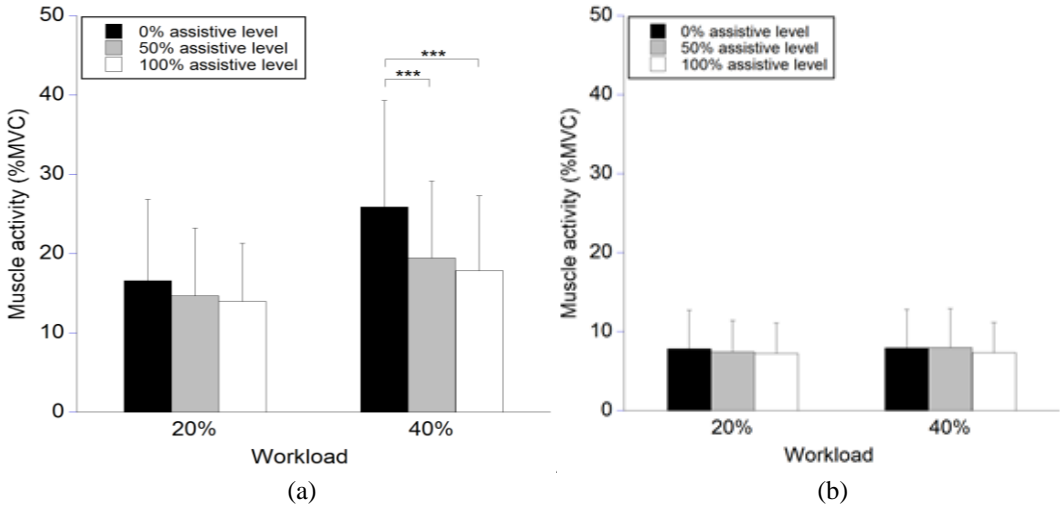
Figure 4: Mean muscle activity (%MVC) for (a) the biceps brachii and (b) the triceps brachii for different levels of assistance

Table 2: Workload x assistive force analysis of variance of the biceps brachii.

Source	Df	F	η	p
(A) Workload	1	65.67	1339.53	.000
(B) Assistive force	1.22	13.16	653.46	.001
A x B (interaction)	1.68	12.53	128.45	.000
Error (within groups)	24			

Table 3: Bonferroni comparison of assistive force levels of the biceps brachii.

Workload	(I) Assistive level (%)	(J) Assistive level (%)	Mean Diff (I-J)	Std. Error	95% Confidence Interval Lower Bound	95% Confidence Interval Upper Bound
	0	50	6.43*	1.42	2.78	10.09
		100	8.08*	1.75	3.58	12.59
40%	50	0	-6.43*	1.42	-10.09	-2.78
		100	1.65	.76	.30	3.59
	100	0	-8.08*	1.75	-12.59	-3.58
		50	-1.65	.76	-3.56	.30

* p < 0.05

Table 4: Workload x assistive force analysis of variance of the triceps brachii.

Source	Df	F	η	p
(A) Workload	1	4.98	2.48	.035*
(B) Assistive force	1.36	5.06	6.86	.022*
A x B (interaction)	1.97	2.89	1.16	.075
Error (within groups)	24			

* $p < 0.05$

RPE

According to the Borg Scale ratings, a significant effect was detected for workload level (F [1,24] = 71.93, p<0.001) and assistive level (F [1.77, 42.45] = 28.98, p<0.001). A significant interaction (F [2,48] = 13.062, p<0.001) was also observed (Table 6). Pairwise comparison using the post-hoc test revealed significant differences among the three assistive levels only at 40% workload, and the higher assistive level decreased the %MVC (Figure 5).
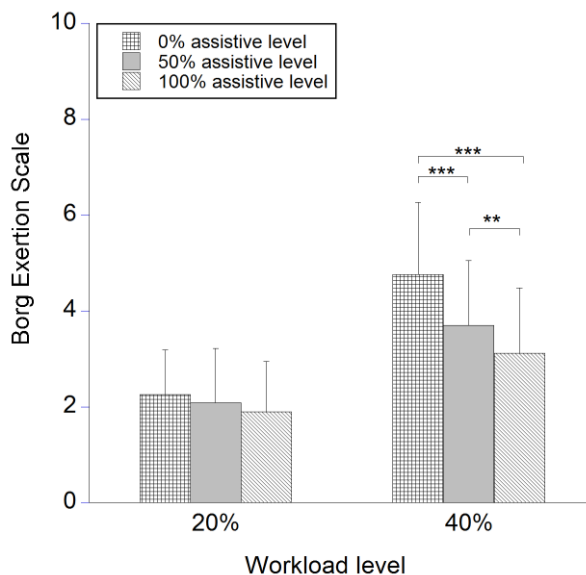
Figure 5: Borg perceived exertion ratings at different workload levels.

Table 5: Workload x assistive force Analysis of Variance of rated perceived exertion.

Source	Df	F	η	p
(A) Workload	1	71.93	118.82	.000
(B) Assistive force	1.77	28.98	14.41	.000
A x B (interaction)	2	13.06	5.36	.000
Error (within groups)	39.6			

Table 6: Bonferroni comparison of assistive force levels.

Workload	(I) Assistive level (%)	(J) Assistive level (%)	Mean Diff (I-J)	Std. Error	95% Confidence Interval Lower Bound	Upper Bound
40%	0	50	1.060*	.196	.554	1.566
		100	1.640*	.223	1.067	2.213
	50	0	-1.060*	.196	-1.566	-.554
		100	.580*	.140	.218	.942
	100	0	-1.640*	.223	-2.213	-1.067
		50	-.580*	.140	-.942	-.218

* $p < 0.05$

Discussion

Effects of assistive force at 40% workload (EMG of main working muscle [BB])

EMG is an electrical activity that is induced by depolarisation of the muscle fibres, depending on the number of active muscle fibres and their firing frequency (Clarys, 2000; Robertson, Caldwell, Hamill, Kamen, & Whittlesey, 2004; Gabriel, Kamen, & Frost, 2006). When the exertional demands increase, the firing rate of muscle fibres moves from lower to higher frequencies, thus resulting in higher EMG amplitude (Cram, 2004; Locks, Santos, Carvalho, Stolt, & Ferreira, 2015). During isometric contraction, EMG amplitude linearly increases with contraction intensity (Linnamo, Moritani, Nicol, & Komi, 2003), and peaks at the MVC (Galganski, Fuglevand, & Enoka, 1993; Rochette, Hunter, Place, & Lepers, 2003; Hunter, Lepers, MacGillis, & Enoka, 2003; Kuriki et al., 2012). Accordingly, the percentage of EMG (i.e. %MVC) reflects the relative intensity of muscle contraction against the MVC. In addition, Borg's Scale is often used to subjectively assess the relative exercise intensity, since values that are ten times the score are roughly equivalent to the percentage of maximal muscle exertion (Pincivero, Coelho, Campy, Salfetnikov, & Bright, 2002; Li & Yu, 2011; Morishita, Yamauchi, Fujisawa, & Domen, 2013).

In theory, the %MVC and the scores of the Borg Scale should be almost equal to workload intensity (percentage of examined workload to maximal workload, i.e., 20% and 40%). In the present experiment, the %MVC of the main working muscle (BB) had lower values than theoretical values (Figure 4), even though the Borg score was nearly the same as the theoretical values. During isometric elbow flexion, although the BB was the main working muscle, other synergist muscles such as the brachialis or pronator teres (Buchanan, Rovai, & Rymer, 1989) could be activated to disperse the load of the biceps.

As shown in Figure 4, the EMG activity of the BB tends to decline as the assistive force increases, which proves that it effectively contributed to the reduction of the exertion of the main working muscle during isometric elbow flexion. However, although previous investigations clarified the effectiveness of a targeted assistive device, they did not clearly

explain how the main working and antagonist muscles adapt to external assistive force (Miyawaki et al., 2010; Bellar, LeBlanc, & Judge, 2014). In the present study, different combinations of the exercise load and assistive level were applied to understand specific muscle responses and adaptability to the assistive force, using a simple experimental device.

Effects of the level of assistive force (50% vs. 100% assistive force)

The assistive force at both the 50% and 100% levels contributed to reduction of the %MVC. From the physical viewpoint, 50% and 100% assistance would decrease by 50% and 100% of the %EMG in the main working muscle (BB), respectively. Nonetheless, the expected reduction was not seen in either assistance level. For the 40% workload, the reduction of EMG was 24.8% with 50% assistance and 31.2% with 100% assistance. The effectiveness of assistance is defined as:

The effectiveness of assistance at the 50% and 100% assistance level was 0.23 and 0.16 in 20% workload, and 0.50 and 0.31 in the 40% workload, respectively. These lower effectiveness values could be due to several reasons such as torque generation, co-contraction of the antagonist muscle, the origin-insertion site of the BB and proprioception.

As reported in previous studies (Murray, 1988; McGuire, Green, & Gabriel, 2014), co-contraction of the BB and TB was observed during isometric elbow flexion. In our study, an assistive force was applied to the mid-forearm during the isometric contraction of the BB. The higher the assistive force that is applied, the stronger the generated upward force is through the supporting strap on the mid-forearm. In order to maintain and stabilise the elbow at 90º flexion, participants would need to resist the upward force to prevent passive elbow flexion movement. A torque at the elbow joint is created along with resistance to the assistive force, which causes BB to increase muscle activity, thus reducing the effectiveness of the assistive force. Furthermore, during assistance, the participants themselves should improve the stability and stiffness of the elbow joint to prevent trauma due to sudden joint movement changes due to experiment equipment trouble and the experimenter's error. Joint stability and stiffness are improved due to co-contraction of the agonist and antagonist muscles (McGuire, Green, & Gabriel, 2014; Bazzucchi, Sbriccoli, Marzattinocci, & Felici, 2006), which should require further contraction against the BB. Thus, these specific muscle activities that respond to assistive force would reduce the effectiveness of assistance.
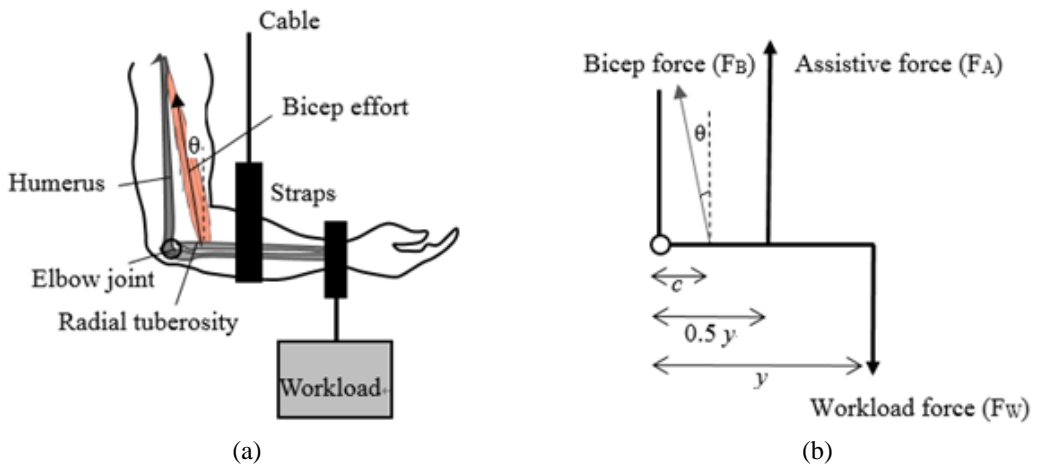
Figure 6: (a) Physiological position of the BB origin-insertion (b) Free body diagram of the elbow joint and acting forces on the radius (y = radius bone length, c = distance between elbow joint and radial tuberosity, θ = tilt angle of the biceps)

The origin-insertion of the BB could also be related to the resulting EMG signal. The origin of insertion determines the position of the muscle based on the tendon muscle's insertion in the bone. Figure 6(a) shows that because the insertion point of the BB is at the radial tuberosity, the muscle is tilted at a certain angle, with the forearm supinated (Hutchinson, Gloystein, & Gillespie, 2008). The tilted position of the BB is not parallel with the direction of the assistive force (Figure 6(b)). The tilted force produced higher muscle activity than if it was in the parallel position, thus decreasing the effectiveness of assistance.

Proprioception is the perception of the body's position and movements in three-dimensional space, and its performance depends on the individual's ability to process sensory signals from the proprioceptors to the brain (Han, Waddington, Adams, Anson, & Liu, 2016). The proprioceptors are sensitive to conscious sensations, such as sense of tension, force, effort, and balance. When assistive load is applied, the sense of effort and heaviness trigger kinaesthetic sensations, providing proprioceptive information about the joint's angle and muscle tension. The brain processes the information, and the output signal modifies the motor control of the muscles, thus coordinating the muscles (Riemann & Lephart, 2002). However, external assistive force should generate a gap between the expected output from proprioceptive information and real output (lifting a specified weight with assistive force). Due to this gap, the participants would overestimate the necessary exerting force to maintain the elbow posture (Proske & Gandevia, 2012), thus reducing the effectiveness of assistance.

On the other hand, the higher assistive level (100%) showed lower effectiveness than the 50% assistance in both workload conditions. At the higher assistance level, the necessary muscle force becomes lower, and its own force is far from the required output. Similarly, low muscle tension should control the required larger output. Therefore, relaxed and light muscle tension might not adequately and accurately control the force. On the other hand, 100% assistance would give greater upward force, hence causing greater flexion of the forearm and creating greater changes in the angle of the elbow. Thus, some muscle tension

may be necessary to control the muscle force accurately to maintain the required 90º elbow position (Kang et al., 2013). The other reason is that participants may maintain necessary muscle activity to prevent damage of the muscles and joints due to unexpected accidents. For instance, if the wire to the belt that is attached to the forearm suddenly snaps, rapid joint movement is forced, which is very dangerous for the muscle and joint. To prevent such damage, individuals may maintain a minimal muscle force even at a higher assistive level.

Effects of assistive force at 20% workload

No significant differences among assistance conditions were observed at 20% workload. Since the 20% workload value was lower than that of the 40% workload, less strength was required for the participants to perform the task. On one hand, the task also required them to maintain the elbow joint angle at 90º. Participants could easily maintain the lower workload by controlling their own muscle tension, rather than utilising assistive force to relieve muscle effort.

Effects of assistive force on workloads in the TB

During isometric contraction, the TB muscle (the antagonist muscle) had small increases seen on EMG, as reported by previous studies (Graves, Kornatz, & Enoka, 2000; Doheny, Lowery, Fitzpatrick, & O'Malley, 2008; Rudroff, Justice, Holmes, Matthews, & Enoka, 2010). This phenomenon is called 'co-contraction' (Tilney & Pike, 1925; Levine & Kabat, 1952). During the unassisted condition, the %MVC was 7.69% and 7.95% at 20% and 40% workload, respectively. However, the %MVC that was obtained in this study was slightly higher than that of Holmes and Keir (2012), who found that the triceps muscle activity was approximately 3% to 5%. It was lower than the result obtained in this study, and the difference could be due to the difference in the position of the forearm; it was natural in their study and supine in this study.

In the assisted condition (Figure 4(b)), the assistive force did not significantly decrease the %MVC of the TB, although the assistive level showed a significant effect. During isometric contraction, the antagonist muscle group maintains joint stability (Rasch, Pierson, & Logan, 1961; Baratta et al., 1988). These results suggest that activities of the antagonist muscle, namely, co-contraction, are essential to maintain elbow joint stability during isometric contraction, irrespective of the presence of assistive force.

Effects of assistive force on RPE

The Borg RPE Scale score was 2.3 and 4.8 at 20% and 40% workload, respectively, for the unassisted condition (Figure 5), showing that as exercise intensified, the RPE and muscle activity increased. This finding was in line with that of previous studies on RPE (Pincivero, Coelho, & Erikson, 2000; Pincivero, Coelho, Campy, Salfetnikov, & Bright, 2001; Lagally, McCaw, Young, Medema, & Thomas, 2004; Troiano et al., 2008; Jakobsen, Sundstrup, Andersen, Aagaard, & Andersen, 2013). During assistance, decreases in RPE scores were observed only in the 40% condition, which was similar to the results

of %MVC for the BB. Accordingly, our results indicated that relief of perceived effort was closely linked to reduction of the main agonist muscle's effort.

Limitations and future study

The present study found unique physiological responses to assistive force, which strongly implies that a physiological perspective is essential for the development of AT. However, to detect specific physiological effects of muscle work against external assistive force, the present study's limitations included the participants' ages (young), sex (male) experience of assistive force (no/little experience), property (isometric contraction), body part (arm), duration (10 sec), intensity (20 and 40% workload), joint angle (90º) of the exercise, level (theoretical 50% and 100%), position (centre of the forearm), direction (upward) and feed-forward (no instruction of the level of assistive force before the task). If these factors are changed, different physiological responses might be apparent. Further investigation should clarify the specific responses against external assistive force in more detail.

Conclusion

During elbow isometric contraction, the muscle activity of the agonist muscle (BB) and perceived exertion decreased with the level of external assistive force, especially at a higher workload (40%). However, their decreases were much smaller than that of the theoretical values, which suggested that unique physiological responses are generated against the assistive force. On the other hand, the activity of the antagonist muscle (TB) was not influenced by the assistive force, since it would help maintain joint stability. These findings strongly imply that a physiological perspective is essential for the development of AT.

Acknowledgments

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