Performance estimation of the lower limb exoskeleton for plantarflexion using surface electromyography (sEMG) signals

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Abstract
In this paper, we evaluate the performance of the rigid and soft exoskeleton by measuring electromyography (sEMG) signal of human lower limb muscles. sEMG represents the degree of muscle activation and the higher sEMG level can be measured if the greater muscle force generated. We compared the sEMG activation level whether wearing the rigid exoskeleton or soft exoskeleton. First, we manufactured the rigid inspired by ‘Berkeley Lower Extremity Exoskeleton (BLEEX)’ and soft exoskeleton motivated by ‘Exosuit’ respectively. After developed the systems, sEMG signals on VM, HAM, GAS, and TA with the rigid lower limb exoskeleton were measured during walking. As a result, up to 150 % muscle activation level increased and it implies that the resistance occurred between human and the rigid lower limb exoskeleton and the user should make an effort to generate more force. After validate the limitation of the rigid lower limb exoskeleton, we did isometric experiment with the soft lower limb exoskeleton, there was 3.4 % normalized MAV decrease at GAS muscle. From this result, we concluded that developed soft lower limb exoskeleton assisted the subject with lower muscle activation level. In addition, the density of the sEMG signal was lower when the subject was assisted by the developed system. It implies that lower fatigue human can feel to maintain isometric condition. Therefore, soft lower limb exoskeleton can assist human more effective than the rigid lower limb exoskeleton.

Key words: Surface electromyography (sEMG), Rigid lower limb exoskeleton, Soft lower limb exoskeleton, Mean absolute values (MAV), Normalized MAV

1. Introduction
There have been abundant rigid lower limb exoskeletons not only for disabled people or patients, but also for soldiers and ordinary people. For rehabilitation and therapy, they assist the patients’ insufficient muscle strength and correct abnormal walking (Ha et al., 2016; Kawamoto and Sankai, 2005). Soldiers can receive exoskeleton’s help to move war supplies and weapons easily and exoskeletons could offer assistance to the workers in the industrial site doing repeated batch operation (Chu et al., 2005; Dollar and Herr, 2008). However, ultimate system does not exist because of some problems. These problems necessitate technology for comprehending the interaction between human and exoskeleton. In addition, complex kinetic constructions of the exoskeleton have to be proposed to align human’s and robot’s lower limb joints. For example, Joint angle change measured by encoder installed in the exoskeleton was different with the human’s natural angle change pattern (Park et al., 2015). To be specific, Schiele found that more joint torques can be generated by joint misalignments and it cause time delay on exoskeleton’s moves relative to the user (Schiele, 2009). Although inertial measurement unit (IMU) can measure the acceleration of linkage easier than other sensors, there are offset drift problem of the sensor and sensor signal delay occurred by the filter (Giansanti et al., 2003). In addition, the weight of the robot and the misalignment of joints can disturb the natural walking of human and generate resistance. Specifically, human’s metabolic rate should be raised when mass added to the body. The leg inertia of human could be changed by additional mass to the legs and it increases the metabolic cost during moving the legs with acceleration. Rose and Gamble (2006) and Browning et al. (2007) showed that the metabolic cost increases up to 7.5% to 8.5% with additional mass (1% to
2%) at the foot during walking (Browning et al., 2007; Rose J and Gamble JG, 2006). In order to solve these problems, soft wearable exoskeletons were developed nowadays. Park et al. developed active soft orthosis to assist ankle movements (Park et al., 2011), Stirling et al. made ankle and knee assist device, and Kawamura et al. proposed hip flexion assistant robot (Kawamura et al., 2013; Stirling et al., 2011). In addition, Asbeck et al. (2015) developed a soft exoskeleton called ‘Exosuit’ utilized wire-driven mechanism to move center of mass (COM) of device up to the upper body (Asbeck et al., 2015). Unlike traditional exoskeletons, this system was lightweight and portable to reduce the resistance between human and robot. Especially, wire-driven mechanism has been issued by other researchers previously due to advantages of the mechanism. One group designed the device which helps stroke patients’ flexion and extension of elbow (Dinh et al., 2016). In et al. (2011) and Vanoglio et al. (2013) developed soft robotic gloves to assist grasp movements and applied devices for rehabilitation. One of advantages using wire-driven mechanism is the weight of the system producing lower inertia and it can reduce the user’s efforts to lift the device. Another advantage is that humans’ kinematics is not restricted due to the flexibility and it can solve the misalignment problem. To evaluate the performance of the rigid and soft exoskeleton, we measured the surface electromyography (sEMG) signal. sEMG is the electrical signal generated by the activation of numerous muscle fibers and sum of motor unit action potentials (MAUP). Moreover, sEMG represents the degree of muscle activation. For example, the higher sEMG level can be measured if the greater muscle force generated (Disselhorst-Klug et al., 2009). Above all, the force estimation of the leg muscles using sEMG has been studied for many applications in these days (Bogey et al., 2005; Fleischer and Hommel, 2008; Lloyd and Besier, 2003). We compared the sEMG activation level whether wearing the rigid exoskeleton or soft exoskeleton. First, we developed the rigid and soft exoskeleton respectively to compare sEMG level with wearing each of them. The rigid exoskeleton was inspired by ‘Berkeley Lower Extremity Exoskeleton (BLEEX)’ which is the most typical lower limb exoskeleton for soldiers (Chu et al., 2005). We developed a lower limb system for monitoring lower limb movements of human during diverse movements. It has Mini-PC and external battery to be operated at the outside of the laboratory environment, such as corridor and outdoor. For soft exoskeleton, our system was motivated by ‘Exosuit’ which has wire-driven mechanism (Asbeck et al., 2015). The design of the system is introduced in Fig.1 and it consists of activation part and control part. In activation part, the cable was attached at the motor mounted in the backpack and heel of the shoe. The motor was activated by the controller in the backpack. Second, we compared the sEMG activation level about both developed exoskeletons. To be specific, the sEMG level at gastrocnemius (GAS), which is activated most during plantar flexion motion at ankle (Neumann, 2013). This muscle is a powerful muscle in the back part of the lower leg and its function is plantar flexing at the ankle joint and flexing at the knee joint.

In this research, we compared the activation level of the EMG signal to compare the efficiency of human assist. The targeted muscles were vastus medialis (VM), hamstring (HAM), gastrocnemius (GAS), tibialis anterior (TA). After sEMG signal from four muscles, we calculated the normalized mean absolute value (MAV) which is MAV divided by force generated by human to compare how much each developed system support the human. If either system has low normalized MAV, it means that the support efficiency of the lower system is larger. Based on these results, we identified the advantage of the soft exoskeleton compared to the rigid exoskeleton widely studied. In addition, we evaluate the performance of the soft lower limb exoskeleton during walking using sEMG activation level.

2. Methods

2.1 Design of the rigid Lower limb exoskeleton

Figure 1 shows the overall design of the rigid lower limb exoskeleton. It has 3 degree of freedoms (DOFs) at the hip, 1 DOF at the knee, and 3 DOFs at the ankle. To fit the joint of human, a roller bearing is installed at center of each joint; rotation, abduction/adduction, and flexion/extension for hip joint, flexion/extension for knee joint, abduction/adduction, plantar flexion/dorsiflexion, and rotation for ankle joint. The joint frames were manufactured by 3D-printer (Dimension 1200es, Stratasys, U.S.A.). The material of them was acrylonitrile-butadiene-styrene (ABS) which is a low cost plastic. The linkage between each joint was made with aluminum (AL1060) because links have to bear the loads. The weight of the foot was 2 kg, shank was 1 kg, and pelvis was 1.5 kg. Total weight was 13 kg including the backpack. Moreover, the length of linkage was chosen according the length of lower limb of the standard of 19-24 aged man and all linkage can
be adjusted at thigh, shank, and hip. A Mini PC (Mini PC: Intel® NUC D54250WYKH, Intel, U.S.A) and battery (NotePack UR60000, URIM KOREA, South Korea) was installed in a backpack of system and it is possible to operate the system at outdoors. The data of sEMG sensors is collected by the Mini-PC directly with 2 kHz sampling rates. This system has not actuation system because it was comparison system with soft lower limb exoskeleton and it was used for investigating the effect of the weight loaded on human lower limb with the system.

2.2 Design of the soft lower limb exoskeleton

The overall design of the soft lower limb exoskeleton is introduced in Fig. 2. The system consists of the actuation part activating the plantar flexion of ankle joint and the control part sensing the human walking pattern and controlling the motor directly.
2.2.1 Actuation

For the actuation part, the end of the cable is connected to the heel under the shoe, and the another end is connected to DC motor. Figure 2 (a) shows cable following rigidly along the leg. Cable only assists ankle’s movement, movements of hip and knee are not assisted by the cable. We used the Bowden cable. The sheath covering the Bowden cable reduces the friction along the cable. Figure 3 shows delay time between motor output and load-cell measurement time when the controller makes the step input. Figure 3 (b) indicates the delay between the motor output generated by step input and step input generated by the controller. Moreover, there is delay between the motor output and the force when motor pulled the wire attached to the motor. As a result, the delay between step input and motor output was 0.6 sec and between motor output and load cell output was 0.12 sec. Totally, 0.72 sec delay occurred immediately after controller generated step input to the motor. In other words, actual time delay to pull the cable or to assist plantar flexion right after controller detects the foot contact sensor input was 0.72 sec. Because this delay was smaller than one second, the user can adjust the system easily.

2.2.2 Control

In control part which is type of backpack, motor is operated and controlled in order to drive the cable. Fig. 2 (c) shows motor, power supply, mini PC and foot contact sensor. In motor part, pulley is attached to fix the cable. Motor driver and power supply are in charge of motor operation. Mini computer controls the motor in response to the input sensor signal. Foot contact sensor measures ground reaction force (GRF) and discriminate gait stance based on fuzzy logic and input into the motor. The input value is step function. It continues 10% of gait cycle which is just before and after heel-off adjusted to the gait velocity. All these components are tightly fixed to hard form and inserted in backpack.

2.3 Data measurement and analysis

sEMG signals were measured by wireless sEMG sensors (Trigno, Delsys, Inc., USA). The signals were filtered between 20 and 450 Hz and then sampled at 2000 Hz. In order to emphasize the significant information in the measured sEMG signals, it is required feature extraction of the signals. First, we calculated the mean absolute values (MAV) (Zecca et al., 2002) of SEMG as follows:

$$\overline{X} = \frac{1}{N} \sum_{k=1}^{N} |X_{ik}|$$

where \(\overline{X}_i\) is a MAV in the ith segment, N is the window length, and \(X_{ik}\) is the kth raw EMG signal in a segment i, respectively. The MAV window length (N) is an important parameter because a large N increases the signal smoothness. However, a large N reduces the responsiveness of SEMG. In contrast, a small N reduces the signal smoothness and it
means that there is some noise on the signal (Morita et al., 2001). In this study, the length of the window \( N \) was set to 200, and the overlap time between adjacent windows was 40 ms.

In addition, sEMG signals for isometric condition as shown in Fig. 5 (a) were different with each trial. We measured GAS muscle activation level. The forces measured by load cell when knee contacted with sensor were different with each trial as shown in Fig. 6 (b), (c). Therefore, we calculated the normalized MAV which is MAV divided by force generated by human for each trials as follows:

\[
\text{Normalized MAV} = \frac{1}{N} \sum_{k=1}^{N} \frac{\text{MAV}_i}{f_i} \text{ (mV/N)}
\]

(2)

where \( f_i \) is a force measured by load cell in the ith trial, \( N \) is the trial number, and \( \text{MAV}_i \) is the MAV of ith trial, respectively.

For experiments, three healthy subjects were recruited who do not have a history of lower limb related surgeries. Mean±standard deviation (SD) for age, height, and body mass of the entire group were 26.83±2.89 yrs, 171.58±7.42 cm, and 64±13.13 kg, respectively.

### 2.4 sEMG for walking with rigid lower limb exoskeleton

As shown in Fig. 4 (a), sEMG sensors were attached to the muscles measured in this research. sEMG signal was attained from vastus medialis (VM), hamstring (HAM), gastrocnemius (GAS), tibialis anterior (TA). It was sampled at 2 kHz using sEMG sensor (Trigno, Delsys, USA). This experiment compared to sEMG signal between normal walk before and after wearing the exoskeleton. Fig. 4 (b) shows sEMG signal during normal walk before and after wearing the rigid lower limb exoskeleton. sEMG signal was normalized based on the maximum value of each muscle signal’s the root mean square before wearing the exoskeleton. As shown in Fig. 4 (b), all signal increased from VM, TA, and HAM after wearing the exoskeleton. Especially, the largest increase in muscles was in GAS whose signal was up to 150%. All results above considered, all muscles made work during walk when wearing the rigid lower limb exoskeleton. This result shows that wearing exoskeleton imposes the load on the user. It implies that there is a limitation in respect of assisting using the rigid lower limb exoskeleton.

### 3. Experiments

In this section, we introduced the experiments about validation of the soft lower limb exoskeleton. We conducted
experiment under the isometric condition to investigate whether the effort of human to lift his leg increases or not. Second, the subject walked on the ground with developed exoskeleton. Specifically, the subject sit on the chair and did plantar flexion at the ankle joint as shown in Fig. 5 (a). When subject did plantar flexion, the knee contacted with the load cell which measured the force generated by the knee. There were three actuation steps as shown in Fig. 5 (b), (c), and (d): heel contact: no actuation, heel-off: actuation, and swing: no actuation. As mentioned in section 2.2.2, motor was actuated immediately after heel-off and the actuating duration was last for one second. Because the actual moment when the system assists the user’s motion is similar with isometric condition, we conducted isometric experiment first and the subject did repetitive plantar flexion for 30 times. During the experiment, the force was measured by the load cell and the data was transferred to the computer through ADC converter with 1000 Hz sampling rates. At the same time sEMG signal of GAS muscle was measured with 2000Hz. As a result, Fig. 6 (a), (b), and (c) indicated measured sEMG signal and force without any assistance, and Fig. 6(d), (e), and (f) implies measured sEMG signal and force with assistance from the soft lower limb exoskeleton. In addition, (a) and (d) shows the raw sEMG signal measured by sEMG sensor, (b) and (e) indicates the MAV of sEMG calculated by Eq. 1, and (c) and (f) represents the force measured by load cell. Finally, Fig. 6 (g) means the normalized MAV calculated by Eq. 2. As shown in Fig. 6 (g), normalized MAV decreased 3.4 % when the system assisted the subject.

4. Discussion

Normalized MAV means MAV per unit force mathematically. In addition, it has a physical reference about how much human should generate muscle force to match the same force measured by load cell whether there is assistance or not. Because it is impossible to exist perfect isometric condition (the subject cannot generate exactly same force), we referred normalized MAV implied in Eq. 2. As a result, there was 3.4 % normalized MAV decrease. It seems that the decrease rate was quite small because knee extension can also be assisted using the soft lower limb exoskeleton although GAS muscle is most related with plantar flexion. In addition, the assistant force may not transfer to the subject because this system was developed for assisting human walking. In other words, the length of Bowden cable and fixing points of cable were optimized to real walking, not for sitting. To investigate the support efficiency of the lower limb exoskeletons, the test with isokinetic condition should be conducted. However, the density of the raw sEMG signal was lower when the system assisted the human as shown in Fig. 6 (a) and (d). It means that human felt lower fatigue to maintain isometric condition. In Fig. 6 (b) and (e), the width of the valley was larger with no assistance.
Conclusion

To evaluate the performance of the rigid and soft exoskeleton, we measured the surface electromyography (sEMG) signal of human lower limb muscles. sEMG represents the degree of muscle activation. For example, the higher sEMG level can be measured if the greater muscle force generated. We compared the sEMG activation level whether wearing the rigid exoskeleton or soft exoskeleton. To compare two systems, we manufactured the rigid and soft exoskeleton respectively. The rigid exoskeleton was inspired by ‘Berkeley Lower Extremity Exoskeleton (BLEEX)’ and we developed soft exoskeleton which was motivated by ‘Exosuit’ which has wire-driven mechanism. Second, we measured sEMG signal on VM, HAM, GAS, and TA with the rigid lower limb exoskeleton. As a result, the largest increase in muscles

![Fig 6. (a) Isometric experiment environment. Load cell was attached at the fixed bar and above the knee, and sEMG sensor attached at gastrocnemius muscle to measure the muscle activation level during plantar flexion of the ankle whether the soft lower limb exoskeleton assists or not. (b) Root mean squared MAV of sEMG signals divided by force whether the soft lower limb exoskeleton assists or not was represented.](image-url)
was in GAS whose signal was up to 150%. It means the rigid lower limb exoskeleton adds the load on the user and the user have to generate more force to operate the system. Next, we measured the sEMG level at gastrocnemius (GAS), which is activated most during plantar flexion motion at ankle with the soft lower limb exoskeleton. For isometric experiment, there was 3.4 % normalized MAV decrease. It means that the muscle activation level can decrease when the user moves plantar flexion with developed system. In addition, the density of the raw sEMG signal was lower when the system assisted the human. It means that human felt lower fatigue to maintain isometric condition. Therefore, we concluded that soft lower limb exoskeleton can assist human more effective than the rigid lower limb exoskeleton.

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