Advanced Robotics

ISSN: 0169-1864 (Print) 1568-5535 (Online) Journal homepage: https://www.tandfonline.com/loi/tadr20

Design of an Under-Actuated Exoskeleton System for Walking Assist While Load Carrying

S. N. Yu , H. D. Lee , S. H. Lee , W. S. Kim , J. S. Han & C. S. Han

///VSP///
Advanced Robotics 26 (2012) 561–580
ADVANCED ROBOTICS
brill.nl/ar

Full paper

Design of an Under-Actuated Exoskeleton System for Walking Assist While Load Carrying

S. N. Yu [a], **H. D. Lee** [b], **S. H. Lee** [b], **W. S. Kim** [b], **J. S. Han** [c] **and C. S. Han** [b,*]

[a] Korea Atomic Energy Research Institute, 1045 Daedeok-daero, Yuseong-gu, Daejeon 305-353, South Korea

[b] Hanyang University, 1271 Sa-3dong, Sangnok-gu, Ansan, Kyeonggi-do 426-791, South Korea

[c] Hansung University, 389 Samseon-dong 2-ga, Seongbuk-gu, Seoul 136-792, South Korea

Received 25 February 2011; accepted 12 April 2011

Abstract

This study proposes an under-actuated wearable exoskeleton system to carry a heavy load. To synchronize that system with a user, a feasible modular-type wearable system and its corresponding sensor systems are proposed. The design process of the modular-type exoskeleton for lower extremities is presented based on the considered requirements. To operate the system with the user, human walking analysis and intention signal acquisition methods for actuating the proposed system are developed. In particular, a sensing data estimation strategy is applied to synchronize the exoskeleton system with a user correctly. Finally, several experiments were performed to evaluate the performance of the proposed exoskeleton system by measuring the electromyography signal of the wearer's muscles while walking on level ground and climbing up stairs with 20- to 40-kg loads, respectively.

Keywords

Wearable robot, exoskeleton, constant force mechanism, muscle stiffness sensor, backpropagation algorithm

1. Introduction

Wearable or exoskeleton systems for human extremities can be utilized in a wide range of applications. As exoskeletons have strong advantages given their unique features, such as their outstanding physical performance, which exceeds that of humans and which is utilized by the operators' nervous system, attempts to adopt them in the industrial field, especially in laborious working sites, are feasible approaches to factory automation.

* To whom correspondence should be addressed. E-mail: hymecer@gmail.com

 DOI:10.1163/156855311X617506

Many institutions around the world have carried out research and development on exoskeletons and assistive devices to empower or aid human lower limbs. MIT's Media Lab developed exoskeletons that would reduce the burden of carried heavy backpacks. This system is composed of a lower-leg device and its weight is about 11.7 kg. The exoskeleton has no actuators, only ankle and hip springs and a knee variable-damper. They tested the developed system and verified the significant reduction of load that was carried by the wearer, but the user consumed 10% more oxygen than normal due to the extra effort required to operate that system because of enduring gait interference. Moreover, the climbing ability of this system was not reported [1, 2]. The University of California at Berkeley has also been developing an exoskeleton system referred to as the Berkeley Lower Extremity Exoskeleton (BLEEX). This featured more than 40 sensors and hydraulic actuators, which were attached on the hip, knee and ankle joint. They carried a 75-kg payload, but this payload included the weight of the exoskeleton. BLEEX 1 was relatively large and bulky — each of its robotic legs was a tangle of hydraulic actuators [3]. Berkeley Bionics has developed three exoskeletons: the Human Universal Load Carrier (HULC), the ExoHiker and the ExoClimber. In contrast to the MIT prototype, wearers of the HULC consume 5–12% less oxygen when walking at 2 miles/h and 15% less when carrying an 81-lb load at the same speed. This performance is suited to military personnel involved in long-duration missions [4]. HAL-5, which was developed by Sankai Yoshiyuki of Tsukuba University, goes a step further by incorporating an additional upper-body system that helps wearers to lift up to 40 kg more than they normally could. Wearing the suit, a healthy adult male can lift 80 kg, roughly double his typical 30–40 kg capability. This system applied an electromyography (EMG) system to communicate with the wearer, but the EMG signal measured from the EMG sensor cannot maintain its signal quality for a long time. Therefore, this system can continue holding a heavy weight for 5–10 min only. Recently, a new version of the HAL series that has only the lower extremity part has been released and tested [5, 6]. Honda developed a walking assist device using a compact motor system to support the wearer's own body weight, especially the lower extremities. The system reduces the load on the wearer's legs while walking and climbing stairs, and while in a semi-crouching position. They developed two types of system — one is a stride management assist system for those with weakened leg muscle who are still able to walk and the other is a body weight support assist system for activities requiring extended standing or repetitive lower-body tasks [7, 8]. eLEGS of Berkeley Bionics is designed to be worn by paraplegics, providing the power and support to walk out of a wheelchair, and standing and walking alone. This system, however, is not for load carrying by normal persons, but for restrictive walking for the handicapped [9]. Northeastern University's Active Knee Rehabilitation Device (AKROD), Yobotics Inc.'s RoboKnee and the NTU-LEE rehabilitation prototype are some of the state-of-art developments in the area of assistive devices to aid human limbs [10–12]. In addition to these systems, many kinds of knee assistive systems are focused on medical services or rehabilitation. The purpose of this device is to share the load or pressure acting on the knee so as to relieve pain or speed up the healing process without disrupting normal daily activities. This is likely to be a potentially useful research area due to the increasing numbers of sports-related injuries and the increasingly aging world population [13]. Obviously, this concept can be applied to assist daily life walking and laborious working in industrial areas. For the purpose of industrial usage, however, operational convenience and compactness of the system is strongly considered. This means that the system has to be designed as wieldy and easy-synchronized as humans. Innovative sensor suits have been developed, which can be put on by an operator to detect his or her motion intention by monitoring his or her muscle conditions (e.g., the shape, stiffness, density, etc.). These sensors are made of soft and elastic fabrics embedded with arrays of micro-electromechanical system sensors, such as muscle stiffness sensors, ultrasonic sensors, accelerometers and optical fiber sensors, to measure different kinds of human muscle conditions [14]. The developers of these sensor systems emphasized their convenience to adapt to humans easily. These sensors, however, are too complicated to manufacture easily or only verified by their performance for a restricted part of the human body. The EMG sensor is one of the most accurate measurement tools to measure the intensity of human motion. The approach using this sensor as an indicator of the wearer's intention, however, is not considered in this study, but only for performance measurement means because of its inconvenient preparation to measure the signals and incongruence for daily life working condition. In particular, energy consumption efficiency is becoming a major issue in this area. A few kinds of systems developed earlier applied a stand-alone-type energy supplier, but they did not satisfy acceptable requirements of operating in field conditions. This is an on-going issue to be solved. Kim *et al.* tried to analysis and minimize the energy consumption of the exoskeleton system [15].

In this study, an under-actuated lower exoskeleton system to carry a heavy load is proposed. To synchronize that system with the wearer, a feasible modular-type exoskeleton system and its corresponding sensor systems are newly proposed. This paper consists of the following issues. (i) The design process of the modular-type exoskeleton for the lower extremities is presented based on the considered requirements. (ii) Human walking analysis and intention signal acquisition methods for actuating the proposed system are introduced. (iii) An idea for sensing data estimation is presented to synchronize the exoskeleton system with the wear correctly. (iv) Several experiments were performed to evaluate the performance of the proposed exoskeleton system through measuring the EMG signal of the wearer's muscles.

2. Design Criteria

2.1. Background

The concept of the proposed system is dedicated to walking and climbing with heavy material. To propose the wearable exoskeleton system, the following conditions are assumed:

- Force augmentation for human walking is not considered.
- Velocity of human walking does not exceed 1.5 km/h while carrying the heavy load.
- Load shape is fixed as a normal backpack. Abnormal shapes are not considered, such as long or large-shaped backpacks.

To concentrate on reducing any additional burden, this study is approached as a motion-synchronized exoskeleton system, not an advanced one. Both of these concepts are different. In case of the first, the performance of simultaneous motion of the exoskeleton for the human's motion is important. In the latter case, active motion at advance based on the estimation of the wearer's intent is required. The proposed system is composed of a human–machine interface (HMI) sensor that measures the status of the wearer's motion, and an exoskeleton system that combines the controlled electric power motion for the knee joint and constant force mechanism system for hip joint. This study suggests the combination of active and passive joint systems. Actually, the power of the wearer is not augmented from this system, but saved while his/her is walking.

2.2. Design Concept

Basically, for load carrying, the proposed exoskeleton has a lower exoskeleton and a ground support structure. The total degree of freedom is 3 for each leg, and one active joint and two passive joints are integrated. The possible position of the active joint can be considered as the hip, knee and ankle joint, and also two or three active joints can be considered. This research focused on the load carrier that performs the anti-weight function of the load while walking level or walking on steps; hence, other motions such as crouching, ducking or bowing motions are not considered. According to this scope, when the exoskeleton is operated on level ground, the leg in the stance phase is only required to support the load bearing along the z-axis direction, which is normal to the ground. During the phase from heel strike to the stance phase, an energy conservative ankle brace can perform load-bearing functions against the weight of the total system and gait propulsion function simultaneously. The leg in the swing phase is supported by the actuated knee joint of the exoskeleton. If this system is operated on a step or inclined ground, this actuated knee joint additionally plays a supporting role of the knee joints of the wearer (see Fig. 1).

3. Mechanical System

3.1. Structure Design

Figure 2 is a view of the exoskeleton for assisting muscular strength of the lower extremities. In the exoskeleton of this embodiment shown in Fig. 2, a drive unit is only provided to the knee joints to rotate an upper-side outer frame and a lower-side outer frame, and thus a hip joint part is constituted by a passive joint to which

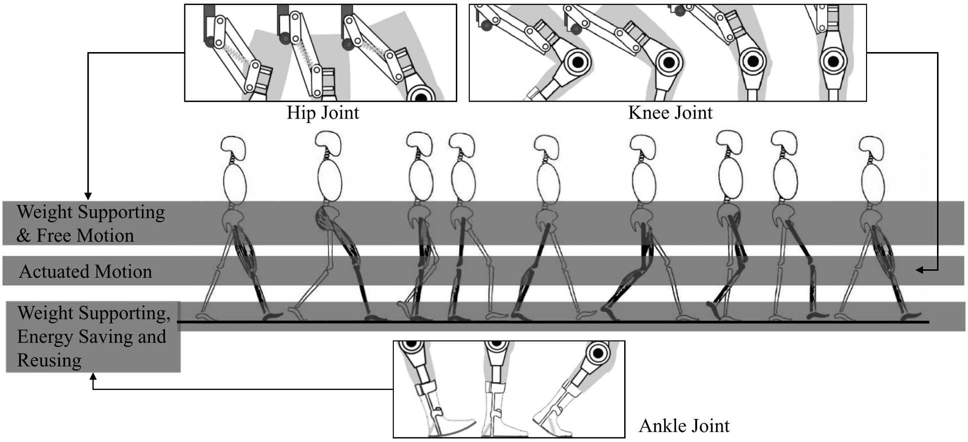


Figure 1. Human's walking cycle and mechanical concept of the proposed system for the corresponding phase.

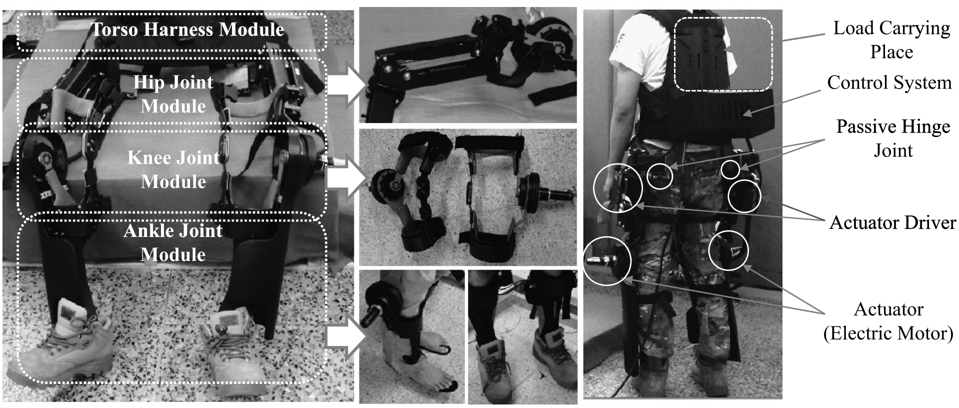


Figure 2. Mechanical design of the proposed system.

such a drive unit is not provided. In Fig. 2, the exoskeleton includes a central torso harness module, the hip joint module, a knee joint module and an ankle joint module. The torso harness module serves to bridge the exoskeleton with the wearer. In order to allow the load to be exerted not on the wearer's body but on the exoskeleton for assisting the muscular strength of the lower extremities, the torso module should be connected to the lower components of the exoskeleton directly. Then the wearer just holds the exoskeleton by connecting the torso harness module of the exoskeleton with his/her upper body to prevent the exoskeleton falling down. The upper-body wearing portion of the torso harness module is provided with a shoulder strap that allows the upper-body wearing portion to be carried on the wearer's upper body like a backpack and further includes a belt to improve the stability of the exoskeleton. The torso harness module is connected to the hip joint module through the backbone, and the hip joint module bears the total weight of the upper

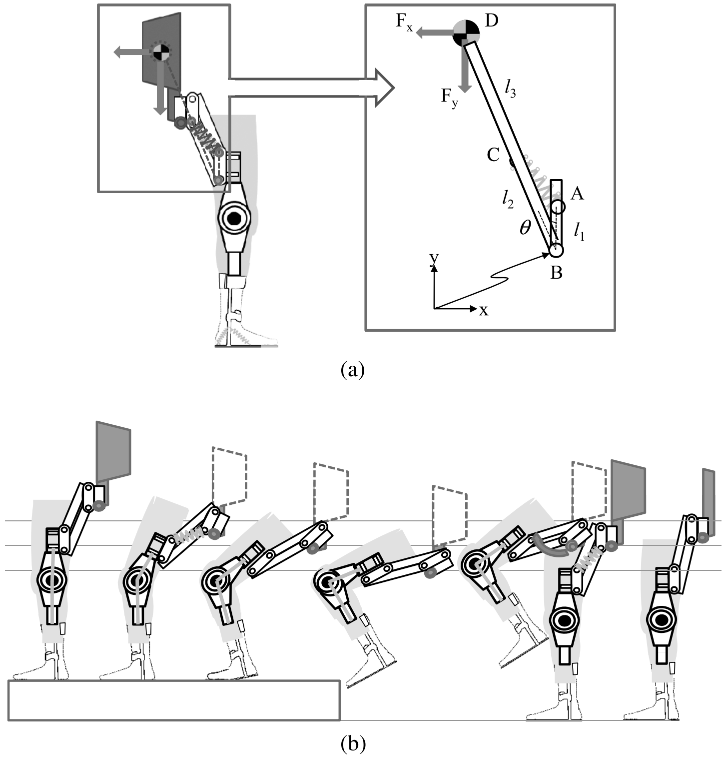


Figure 3. Kinematic structure of the hip joint mechanism.

exoskeleton system and its loading weight by a constant force mechanism. Then the hip joint module is combined with actuated knee joint module and ankle joint module in order. The ankle joint module includes a flexible hyper carbon structure that can bear the total weight of the exoskeleton system and provide a propulsion force for walking by the potential energy of its own elastic deformation.

Figure 3 shows the concept of the designed system's operation while stepping up. The swinging of leg is activated by the human's own effort, but after the stance phase, load bearing and propulsion (supported by the hip joint mechanism and actuated by the knee joint motor) is performed by the exoskeleton. The performance of the proposed system is simulated while walking with an attached weight. The variation of oscillation for a loaded 40-kg weight is under 10 cm while walking at 1.5 km/h and this variation can be held by a torso harness attached to the exoskeleton system. Moreover, the stiffness of the hip joint mechanism can be adjusted mechanically for the wearer's own body traits.

3.2. Design of the Constant Force Mechanism for the Hip Joint

Figure 3 shows the mechanical assumption and modeling. The gravitational force induced by the mass of the backpack is applied at $D(x, y)$ and the base frame is located at point B. The position is expressed as:

$$x = (l_2 + l_3)\sin\theta \tag{1}$$

$$y = l_1 - (l_2 + l_3)\cos\theta. \tag{2}$$

A linear spring is connected between A and C, and its length is expressed as:

$$r^2 = l_1^2 + l_2^2 - 2l_1 l_2 \cos\theta. \tag{3}$$

The performance of point D can be expressed using r for each axis. Differentiate (3) with time and arrange the equation as:

$$r\dot{r} = l_1 l_2 \sin\theta\dot{\theta}. \tag{4}$$

From the derivative of (1):

$$\dot{\theta} = \frac{\dot{x}}{(l_2 + l_3)\cos\theta} \tag{5}$$

$$\sin\theta = \left[\frac{(2l_1 l_2)^2 - (l_1^2 + l_2^2 - r^2)^2}{4l_1^2 l_2^2}\right]^{1/2}. \tag{6}$$

Substituting (5) and (6) into (4) and arranging the equation, the time-dependent displacement relation between point D and spring length l_s for the x- and y-axis:

$$\dot{x} = \left[\frac{(l_2 + l_3)(l_1^2 + l_2^2 - r^2)}{2l_1^2 l_2^2}\left\{\frac{4l_1^2 l_2^2 - (l_1^2 + l_2^2 - r^2)^2}{4l_1^2 l_2^2}\right\}^{-1/2} r\right]\dot{r} \tag{7}$$

and briefly:

$$\dot{x} = g_r^x \dot{r} \tag{8}$$

$$\because g_r^x = \frac{(l_2 + l_3)(l_1^2 + l_2^2 - r^2)}{2l_1^2 l_2^2}\left\{\frac{4l_1^2 l_2^2 - (l_1^2 + l_2^2 - r^2)^2}{4l_1^2 l_2^2}\right\}^{-1/2} r \tag{9}$$

$$\dot{y} = \left[\frac{(l_2 + l_3)}{l_1 l_2} r\right]\dot{r}. \tag{10}$$

Likewise:

$$\dot{y} = g_r^y \dot{r} \tag{11}$$

$$\because g_r^y = \frac{(l_2 + l_3)}{l_1 l_2} r. \tag{12}$$

From this equation, the effective stiffness can be derived. At point D, the effective force F_{eff} is expressed as the following equation based on the virtual work theory:

$$F_{\mathrm{eff}} = F_s + g_r^x F_x + g_r^y F_y. \tag{13}$$

If the initial length of the spring is considered as r_0, the elastic force is:

$$F_s = -k_s(r - r_0). \tag{14}$$

Here, the effective stiffness of the applied spring is defined as follows, and it can be expressed as (16) based on (13) and (15):

$$k_{\mathrm{s}} = -\frac{\partial F_{\mathrm{s}}}{\partial r} = -\frac{\partial F_{\mathrm{s}}}{\partial r} - g_r^x \frac{\partial F_x}{\partial r} - g_r^y \frac{\partial F_y}{\partial r} - h_r^x F_x - h_r^y F_y \\ \left(\because h_r^x = \frac{g_r^x}{\partial r}, h_r^y = \frac{g_r^y}{\partial r}\right). \tag{15}$$

If we assume that the external force exerted to the spring mechanism is from the loaded weight on the backpack and there is no ducking motion while the wearer operates this system, the exerted force for each axis is considered as:

$$F_x = 0, \qquad F_y = \text{constant}. \tag{16}$$

The effective force of (13) is expressed as (17) using (12), (14) and (16):

$$F_{\mathrm{eff}} = -k_{\mathrm{s}}(r - r_0) + \frac{(l_2 + l_3)}{l_1 l_2} r F_y. \tag{17}$$

To perform stable walking while loading the allowable weight, the additional condition of $F_{\mathrm{eff}} = 0$ should be added:

$$k_{\mathrm{s}} r_0 = \left(k_{\mathrm{s}} - \frac{(l_2 + l_3)}{l_1 l_2} F_y\right) r. \tag{18}$$

Therefore, the equilibrium position of the spring mechanism while loading the weight is calculated as:

$$r = \frac{k_{\mathrm{s}} r_0}{k_{\mathrm{s}} - ((l_2 + l_3)/(l_1 l_2)) F_y} \left(k_{\mathrm{s}} \neq \frac{(l_2 + l_3)}{l_1 l_2} F_y\right). \tag{19}$$

The selection of the stiffness of the designed spring mechanism is a trade-off issue between the stability of the loaded weight and the comfort of the wearer's leg swing. Adequate stiffness of the designed system is confirmed using (19) for the maximized allowable weight. Actually, for each wearer, practical modification can be performed through an adjustment of the stiffness of the spring mechanism while performing the wearing test.

4. HMI for System Operation

4.1. Sensor Design for the HMI

In this study, a muscle stiffness sensor (MSS) to acquire the signal for the degree of expansion of the muscle is developed. The measured signal is utilized as the wearer's intent signal to operate the proposed exoskeleton system (Fig. 4). Before these sensor systems synchronize with the designed lower extremity system, a calibration process has to be performed. A personal computer is used as a main controller; additionally, a signal analyzer, an external DAQ system and MSS sensors are included. After the installation of the MSS into the operator's body, the stiffness parameter

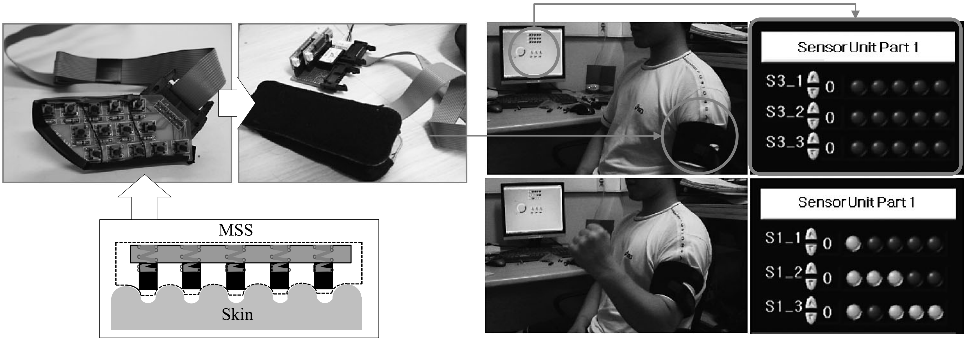


Figure 4. Designed MSS and its performance verification.

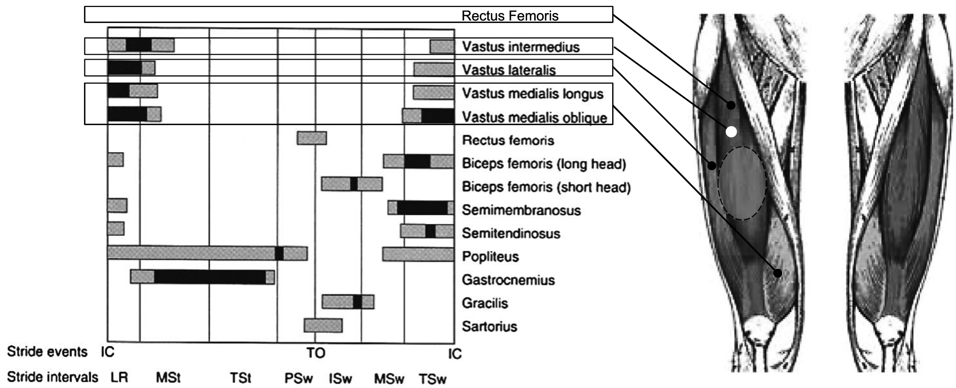


Figure 5. Muscle activation of the lower extremities and MSS position for the proposed system [16].

under maximum activity and under minimum activity for each muscle was measured. Using these data at the maximum and minimum conditions, the stiffness parameter was normalized as:

$$\begin{aligned} S_{\mathrm{NR}} &= \frac{S_{\mathrm{R}} - \min(S_{\mathrm{R}})}{\max(S_{\mathrm{R}}) - \min(S_{\mathrm{R}})} \\ S_{\mathrm{NG}} &= \frac{S_{\mathrm{G}} - \min(S_{\mathrm{G}})}{\max(S_{\mathrm{G}}) - \min(S_{\mathrm{G}})}, \end{aligned} \tag{20}$$

where S_{NR} is the normalized muscle stiffness of the rectus femoris, S_{NG} is the normalized muscle stiffness of the gastrocnemius, S_{R} is the measured muscle stiffness of the rectus femoris and S_{G} is the measured muscle stiffness of the gastrocnemius.

The developed MSS is composed of an array of 15 switches and a flexible electric board. In particular, a space margin for human skin contact and placement of the sensor are considered as the dashed circle of Fig. 5. During the stance phase, the quadriceps muscle group is relied on to control its tendency towards knee flexion collapse with weight acceptance and single-limb support. This muscle group is activated during terminal swinging and then acts eccentrically during weight acceptance, as the knee rotates from the fully extended position at the initial contact to its peak support phase flexion of approximately 20° during the loading response. Thereafter, the quadriceps act concentrically to extend the knee through an early mid-stance, as the extremity is raised vertically over the supporting limb and the anterior orientation of the ground reaction force vector precludes the need for further muscular control of the knee flexion. Most hamstring muscles are activated in the late mid-swing or the terminal swing. Their function with respect to the knee is probably to control the angular acceleration of the knee extension. The short head of the biceps femoris is activated earlier and probably assists in flexing the knee for foot clearance. The algorithm was gradually adjusted to the wearers, such as by calibrating the sensor system and regulating the velocity with fine-tuning before starting the machinery, to develop a handy prototype of the system that is easy to wear. The results of the verification of the effects of assistance through repeated experiments with the lower exoskeleton system and an EMG signal sensing device will be introduced in Section 5.

4.2. Signal Reprocessing using the Backpropagation Algorithm

Several tests were performed for the designed MSS and the feasibility of the sensor is verified in Fig. 6. This result shows that the performance of the designed sensor is similar to the conventional EMG sensor, such as the hysteresis-like feature [17]. According to several performance tests, the motion performed by a human is measured directly while the human wears this sensor system on the considered position.

Additionally, we found other possibilities where this system can be utilized at other motion measurements, such as elbow extension and flexion. However, if this sensor system is applied directly into the exoskeleton system, the performance of the accurate coactivation for leg motion is not sufficient because of its intrinsic hysteresis-like feature. Therefore, to use this system for the designed exoskeleton through the considered strategy, improvement of the measured signal pattern

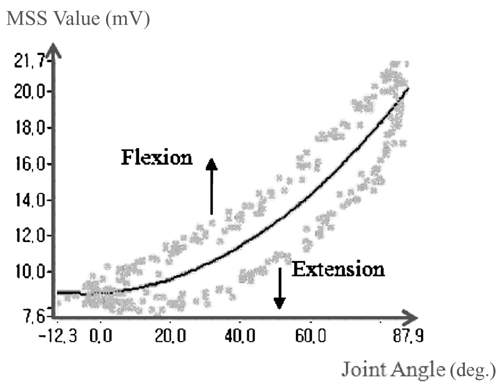


Figure 6. Measured signals of the developed MSS.

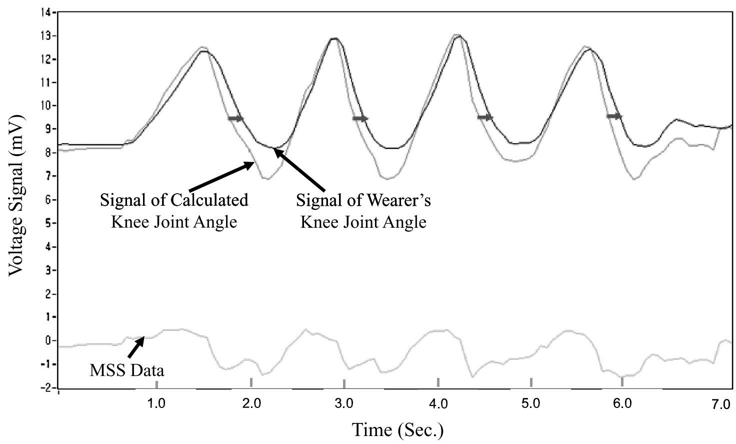


Figure 7. Computed joint angle of a human using measured raw data of the MSS.

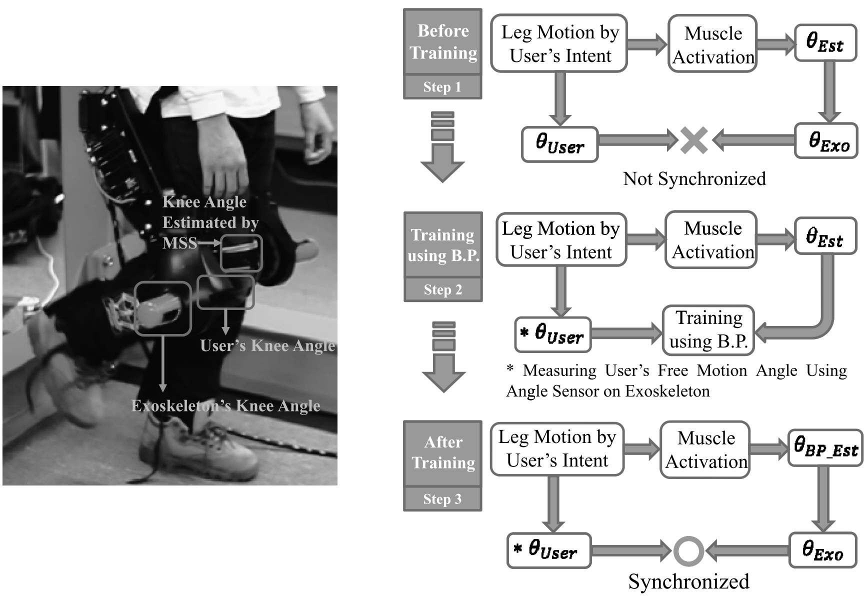


Figure 8. Strategy of synchronization improvement of the HMI using the backpropagation algorithm.

is required because a exoskeleton system should follow the wearer's leg motion correctly (Fig. 7).

Figure 8 shows the basic strategy for motion synchronization between a human and an exoskeleton. As shown in Fig. 7, the estimated exoskeleton motion command from the MSS is not matched with the actual human's knee joint motion. Obviously, the MSS has the advantage of being an easy wearable and versatile system that can be fitted to any user, but its original command signal is not enough to use as a motion command signal for the exoskeleton system. Actually, many researchers have tried to solve this difficulty of a smooth synchronization between a human and an exoskeleton, but the majority of them approached it as a control problem. That is absolutely correct, but every user has a different preference, body size and walking habit, and, most important, calculation load for the real-time control system, which operates many joints simultaneously, is extremely heavy. Therefore, a useful solution that is based on a conventional control strategy is not yet presented.

The above-mentioned problems can be solved by the sensing signal estimation technique. This concept is based on the discrete feedforward system that is controlled adequately by the human nervous system. If an exoskeleton or robot can follow the human's intent correctly for every walking step, and carry its own weight and external weight, a wearer can carry those weights easily with a small ratio of his/her own effort. To implement the proposed sensor system into developed exoskeleton system, the original MSS signal should be modified and used to estimate the real human leg motion correctly; hence, the estimation method of the measured signal for the wearer's own knee motion is proposed using a backpropagation technique of the neural network. Backpropagation is a technology for training the neural network.

The backpropagation algorithm is used for feedforward networks without feedback or loop connection. Figure 9 shows the basic backpropagation algorithm

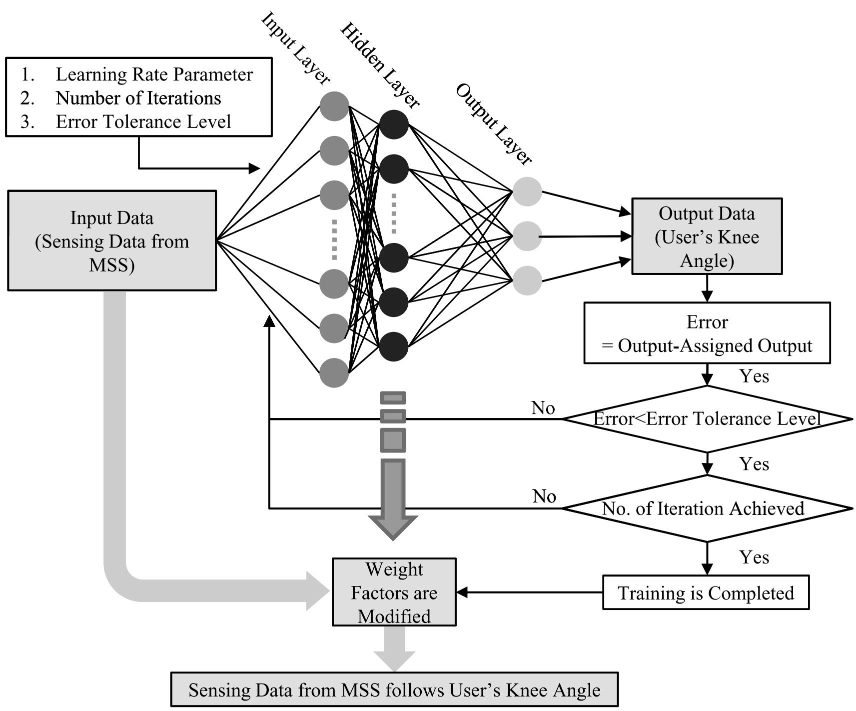


Figure 9. Backpropagation algorithm for sensor signal training.

applied in this study; more detailed mathematical treatments are referred to in Refs [18, 19].

As shown in Figs 8 and 9, the muscle activation signal measured by the MSS is estimated as θ_{Est} and the input to motion command of the exoskeleton as θ_{Exo} directly, while wearer's knee angle $*\theta_{\mathrm{User}}$ is measured by the angle sensor of exoskeleton. Finally, to synchronize this command with the wearer's leg motion, $*\theta_{\mathrm{User}}$ is trained to θ_{Exo} using the backpropagation algorithm. Through this initial training process, the exoskeleton system is ready to be operated by its wearer.

Figure 10 shows the modified MSS signal after training and Fig. 11 shows the results of the walking experiment of the wearer while wearing the exoskeleton that used the trained sensing data of the MSS. As shown in Fig. 11, the exoskeleton followed the targeted wearer's knee angle simultaneously after training.

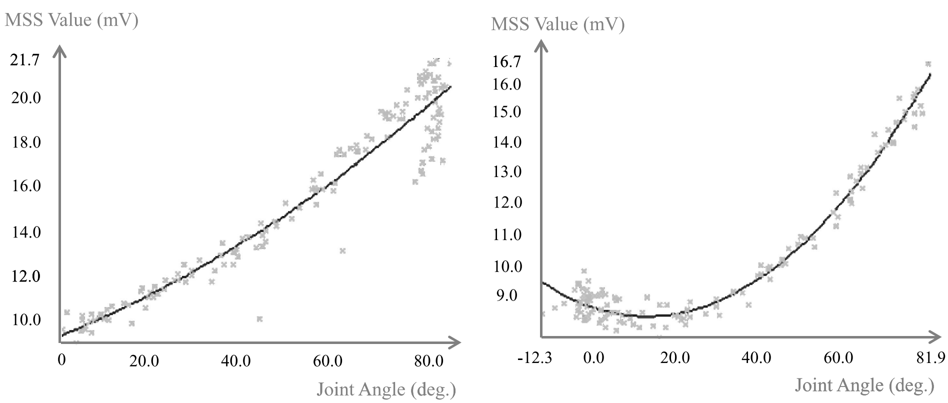


Figure 10. Measured signal of the developed MSS after training of the MSS values by knee flexion (left) and extension (right) motion using the backpropagation algorithm.

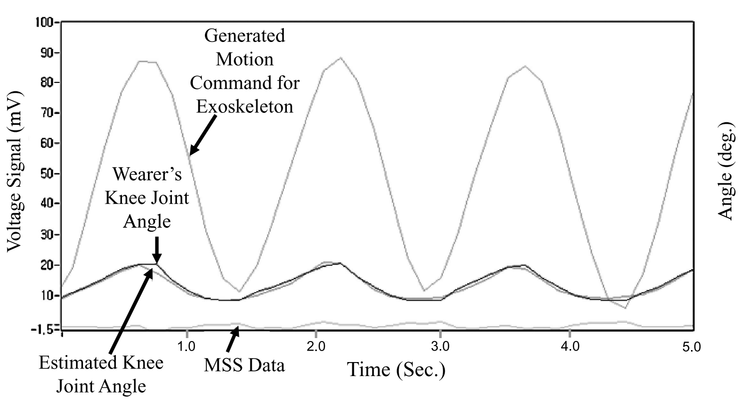


Figure 11. Improved motion command and behaviour using the measured and estimated signals of the MSS.

5. Experimental Results

EMG is typically combined with stride or angular kinematic analysis to provide information on phasic muscle activation patterns. EMG helps explain the motor performance underlying the kinematic and kinetic characteristics of gaits. Actual walking experiments with EMG were performed with the designed exoskeleton system to evaluate the performance of the sensing data estimation technique. The experiments were conducted in the following steps. First, we obtained the EMG signal of subjects while they walked on a step and a step by carrying a 40-kg weight. After that, they wore the developed exoskeleton system. The subject repeated the procedure with the same walking speed (Fig. 12). Finally, we gathered the EMG signal history and verified its feasibility. Figures 13 and 14 show the experimental results. The EMG sampling frequency and gain for signal amplification are 1024 Hz and 1126.7 V/μV, respectively. The EMG electrodes are attached on the upper and lower parts of quadriceps and gastrocnemius muscle groups (Fig. 15).

The EMG signal is gathered during locomotion and compared to the results of the two conditions of not wearing and wearing the exoskeleton. On a level-ground walking condition, the muscle force of the lower limbs is required to be relatively smaller than for a step walking condition because they were mainly activated by gravity within the swing. Then, the percent maximum voluntary isometric contraction (%MVIC) is measured and calculated using a standardized, objective and sensitive tool designed for the measurement of muscle strength. First, the %MVIC

Figure 12. Performance test of the developed system (step walking while carrying a weight of 40 kg).

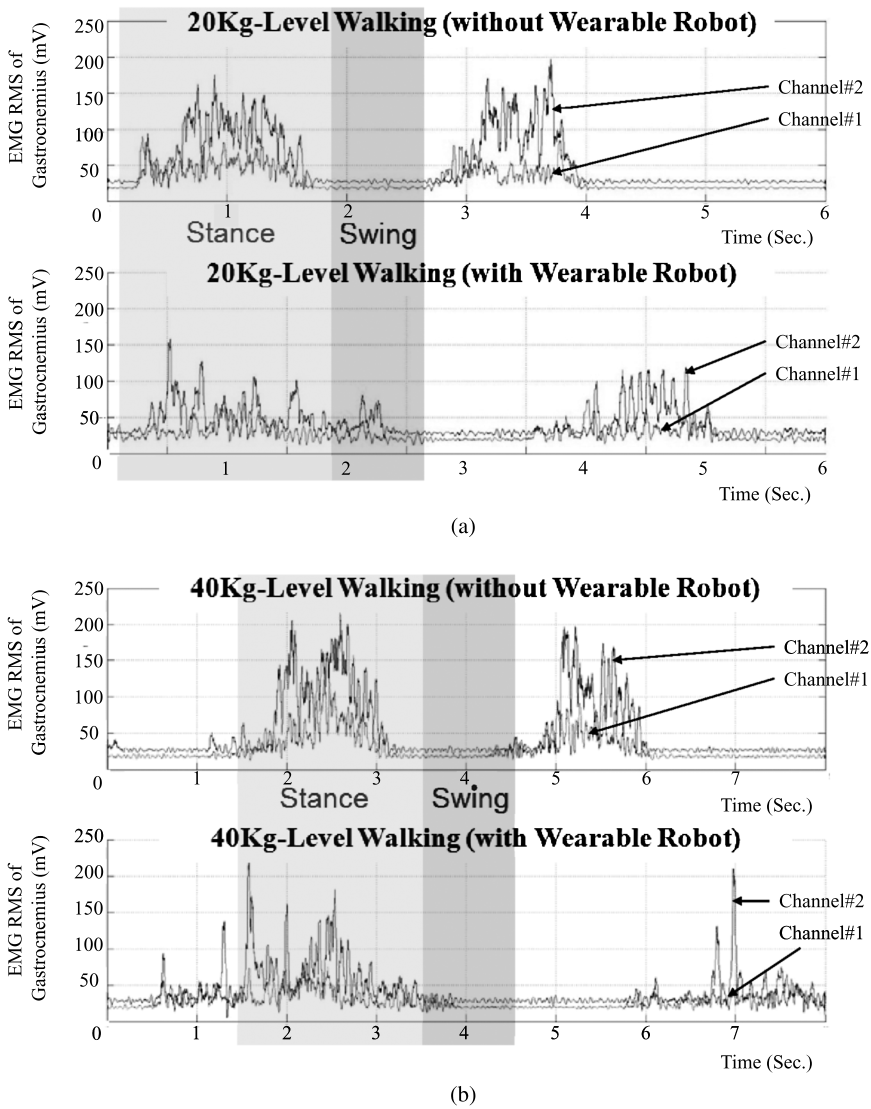


Figure 13. Measured EMG signal for level walking while wearing and not wearing the exoskeleton.

is calculated for the case of walking on level ground. The values decreased as much as 32–49% while wearing the exoskeleton than not wearing it. Table 1 tabulates the measured EMG data and shows the comparison of more muscle force between wearing and not wearing the system while level walking.

As a second experiment, %MVIC is measured in the case of walking on stairs and demonstrated that 11–24% of muscle power is reduced after adapting the designed exoskeleton (Table 2). The degree of the wearer's muscle power assistance has different aspects on level walking and stairway conditions, and better effectiveness is found in the level walking condition than the stairway condition.

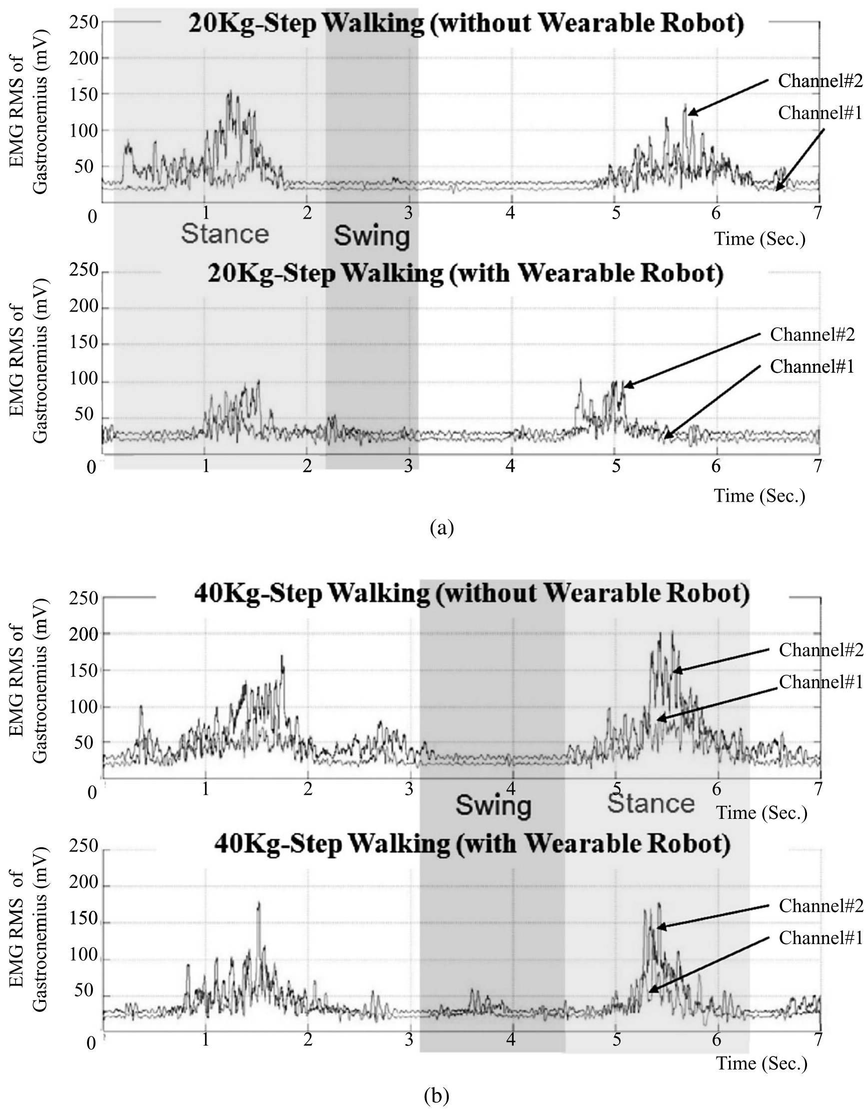


Figure 14. Measured EMG signal for step walking while wearing and not wearing the exoskeleton.

6. Conclusions

We reported on a system designed using a minimized actuator and newly designed passive mechanism. Obviously, the subject who wears the proposed system has to consume his/her own effort while stepping up stairs or ascending a slope because of kicking up the thigh higher than level walking. The hip joint of the proposed system serves as a vertical bearing force against the weight, but swing mechanism is not actuated; hence, the wearer have to swing his/her thigh including the constant force mechanism parts only using muscle force and gravity force. However, the degree

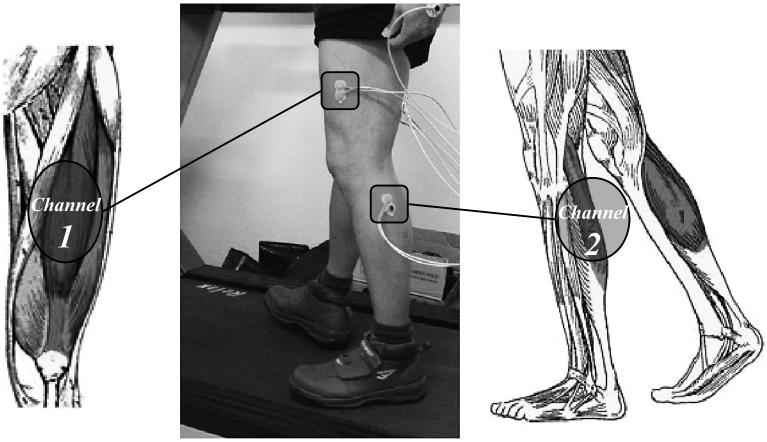


Figure 15. Attachment positions of the EMG electrodes.

Table 1.
Comparison of the required muscle force (%MVIC) between wearing and not wearing the exoskeleton (level walking)

		Without exoskeleton	With exoskeleton	Muscle force decrease
20 kg	Ch. 1	74.5%	30.8%	43.7%
	Ch. 2	84.2%	47.3%	36.9%
40 kg	Ch. 1	95.7%	46.2%	49.5%
	Ch. 2	94.4%	61.9%	32.5%

Table 2.
Comparison of the required muscle force (%MVIC) between wearing and not wearing the exoskeleton (step walking)

		Without exoskeleton	With exoskeleton	Muscle force decrease
20 kg	Ch. 1	54.6%	35.3%	19.2%
	Ch. 2	77.6%	67.1%	10.5%
40 kg	Ch. 1	69.1%	48.0%	21.1%
	Ch. 2	97.5%	87.2%	10.3%

of the effort is extremely reduced considering the carried weight of the total system. Moreover, in spite of being an underactuated system, the proposed exoskeleton system shows acceptable performance, as shown in Tables 1 and 2 and Fig. 16.

In future work, an additional exoskeleton system for the upper extremities will be combined and operated with the proposed lower exoskeleton system simultaneously. The integrated system of Fig. 17 can manipulate the weight at the wearer's

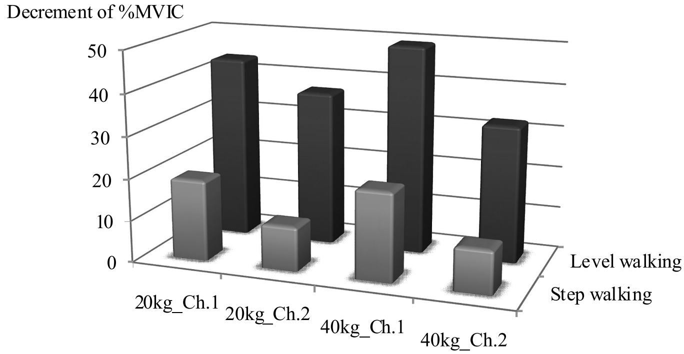


Figure 16. Decrease of %MVIC of the measured muscle while wearing the exoskeleton system.

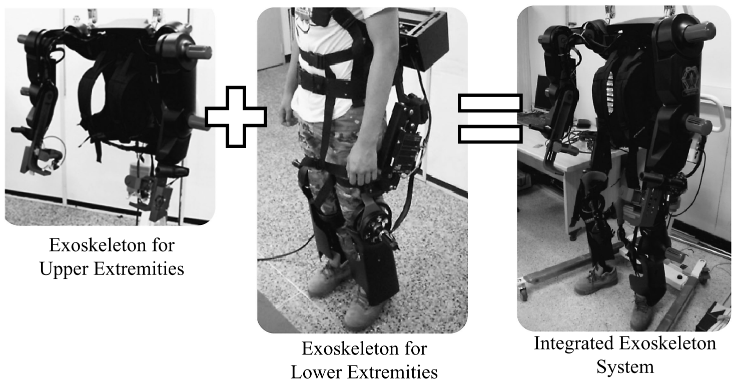


Figure 17. Integrated exoskeleton system for further studies.

intent. The stability problem and maximum load handling performance will be evaluated.

Acknowledgements

This research was partially supported by the robot R&D program of the Ministry of Knowledge and Economy (MKE) of Korea (No. 10035431, Development of wearable robot for industrial labor support) and partially by the MKE, Korea under the Advanced Robot Industrial Promotion Agency (NIPA) (NIPA-2011-C7000-1001-0002).

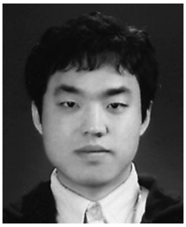
References

1. C. J. Walsh, K. Pasch and H. Herr, An autonomous, underactuated exoskeleton for load-carrying augmentation, in: *Proc. IEEE/RSJ Int. Conf. on Intelligent Robots and Systems*, Beijing, pp. 1410–1415 (2006).
2. C. J. Walsh, K. Endo and H. Herr, A quasi-passive leg exoskeleton for load-carrying augmentation, *Int. J. Humanoid Robotics* **4**, 487–506 (2007).

3. A. Zoss, H. Kazerooni and A. Chu, On the mechanical design of the Berkeley Lower Extremity Exoskeleton (BLEEX), in: *Proc. IEEE/RSJ Int. Conf. on Intelligent Robots and Systems*, Edmonton, AB, pp. 3132–3139 (2005).
4. R. Bogue, Exoskeleton and robotic prosthetics: a review of recent developments, *Industrial Robot* **36**, 421–427 (2009).
5. http://www.cyberdyne.jp.
6. H. Kawamoto, S. Taal, H. Niniss, T. Hayashi, K. Kamibayashi, K. Eguchi and Y. Sankai, Voluntary motion support control of robot suit HAL triggered by bioelectrical signal for hemiplegia, in: *Proc. 32nd Annu. Int. Conf. of the IEEE EMBS*, Buenos Aires, pp. 462–466 (2010).
7. http://corporate.honda.com/innovation/walk-assist.
8. S. Krut, M. Benoit, E. Dombre and F. Pierrot, MoonWalker, a lower limb exoskeleton able to sustain body weight using a passive force balancer, in: *Proc. IEEE Int. Conf. on Robotics and Automation*, Anchorage, AK, pp. 2215–2220 (2010).
9. http://berkeleybionics.com/2010/introducing-elegs.
10. B. Weinberg, J. Nikitczuk, S. Patel, B. Patritti, C. Mavroidis, P. Bonato and P. Canavan, Design, control and human testing of an active knee rehabilitation orthotic device, in: *Proc. IEEE Int. Conf. on Robotics and Automation*, Rome, pp. 4126–4133 (2007).
11. http://yobotics.com/robowalker/robowalker.html.
12. K. H. Low, X. Liu and H. Yu, Development of NTU wearable exoskeleton system for assistive technologies, in: *Proc. IEEE Int. Conf. on Mechatronics and Automation*, Niagara Falls, NY, pp. 1099–1106 (2005).
13. S. A. Paluska and M. D. McKeag, Knee braces: current evidence and clinical recommendations for their use, *Am. Fam. Physician* **61**, 411–418 (2000).
14. M. Q. Feng, Sensor suit for human motion detection, *Report ADA444285*, DTIC (2006).
15. W. S. Kim, S. H. Lee, M. S. Kang, J. S. Han and C. S. Han, Energy-efficient gait pattern generation of the powered robotic exoskeleton using DME, in: *Proc. IEEE/RSJ Int. Conf. on Intelligent Robots and Systems*, Taipei, pp. 2475–2480 (2010).
16. M. Nordin and V. H. Frankel, *Basic Biomechanics of the Musculoskeletal System*, 3rd edn. Lippincott Williams and Wilkins, Baltimore, MD (2001).
17. T. Kimura, K. Yamanaka, D. Nozaki, K. Nakazawa, T. Miyoshi, M. Akai and T. Ohtsuki, Hysteresis in corticospinal excitability during gradual muscle contraction and relaxation in humans, *Exp. Brain Res.* **152**, 123–132 (2003).
18. D. S. Kim, *Neural Networks: Theory and Applications*. Jinhan M&B, Seoul (1993).
19. R. Hecht-Nielsen, Theory of the backpropagation neural network, in: *Proc. Int. Joint Conf. on Neural Networks*, Washington, DC, pp. 593–605 (1989).

About the Authors

Seung-Nam Yu graduated with the BS, MS and PhD degrees in Mechanical Engineering from Hanyang University, South Korea, in 2004, 2006 and 2009, respectively. Since 2010, he has been working as a Senior Researcher of the Korea Atomic Energy Research Institute (KAERI). His research interests are in human–robot interaction and teleoperation techniques related to hazardous environments. He is a Member of the Korean Society of Mechanical Engineers (KSME) and The Institute of Control, Robotics and Systems (ICROS).



Hee-Don Lee received the BS degree in Mechatronics Engineering from the Catholic University of Daegu, South Korea, in 2006, and the ME degree in Mechanical Engineering from Hanyang Unversity, South Korea, in 2008. He is currently a PhD candidate of Mechanical Engineering at Hanyang University. His research interests include physical human–robot interaction, bio-mechanics and robot force control.

Seung-Hoon Lee received the BS degree in Mechanical Engineering from Hanyang University, South Korea, in 2007, and the ME degree in Mechanical Engineering from Hanyang Unversity, in 2009. He is currently a PhD candidate of Mechanical Engineering at Hanyang University. His research interests include human–robot cooperation, robot force control and field robots.

Wan-Soo Kim received the BS degree in Mechanical Engineering from Hanyang University, South Korea, in 2008, and the ME degree in Mechanical Engineering from Hanyang Unversity, in 2010. He is currently a PhD candidate of Mechanical Engineering at Hanyang University. His research interests include human–robot interaction, bio-mechanics and force control.

Jung-Soo Han received the PhD degree in Biomedical Engineering from the University of Iowa, USA, in 1991. From 1991 to 1992, he was an Assistant Professor and a Chief of the Bioengineering Laboratory of Texas Tech University. From 1992 to 1996, he was an Assistant Professor and a Chief of the Bioengineering Laboratory of West Virginia University. Since 1996, he has been working as a Professor in the Department of Mechanical Systems Engineering at Hansung University, South Korea. His research interests include rehabilitation robots and force-assistive exoskeleton systems.

Chang-Soo Han received the PhD degree in Mechanical Engineering from the University of Texas at Austin, USA, in 1989. From 1987 to 1988, he was a Consultant for a Lockheed MAC III design project of Lockheed Austin Division, and from 1996 to 1997, Visiting Professor at the University of California at Berkeley. Currently, he is a Professor of Robotics in the Department of Mechanical Engineering, Hanyang University, South Korea. His research interests include high-precision robotics, biomechanical rehabilitation systems and automation in construction. He is a Member of the Korean Society of Mechanical Engineers (KSME).