Evaluation of active wearable assistive devices with human posture reproduction using a humanoid robot

Article *in* Advanced Robotics · June 2018

DOI: 10.1080/01691864.2018.1490200

ADVANCED ROBOTICS
https://doi.org/10.1080/01691864.2018.1490200

FULL PAPER

Evaluation of active wearable assistive devices with human posture reproduction using a humanoid robot*

Takahiro Ito [a], Ko Ayusawa [b], Eiichi Yoshida [a,b] and Hiroshi Kobayashi [c]

[a]Graduate School of Systems and Information Engineering, University of Tsukuba, Tsukuba, Japan; [b]CNRS-AIST JRL (Joint Robotics Laboratory), UMI3218/RL, Intelligent Research Institute, National Institute of Advanced Industrial Science and Technology (IS-AIST), Tsukuba, Japan; [c]Tokyo University of Science, Tokyo, Japan

ABSTRACT
This study proposes a quantitative evaluation method for assessing active wearable assistive devices that can efficiently support the human body. We utilize a humanoid robot to simulate human users wearing assistive devices owing to various advantages offered by the robot such as quantitative torque measurement from sensors and highly repeatable motion. In this study, we propose a scheme for estimating the supportive torques supplied by a device called stationary torque replacement. To validate the reliability of this evaluation method by using a humanoid robot, we conducted measurements of human muscular activity during assisted motion. Analysis of the measured muscle activity revealed that a humanoid robot closely simulates the actual usage of assistive devices. Finally, we showed the feasibility of the proposed evaluation method through an experiment with the humanoid robot platform HRP-4 and the Muscle Suit active assistive device. With the proposed method, the supportive effects of the assistive device could be measured quantitatively in terms of the static supportive torque acting directly on the body of a simulated human user.

ARTICLE HISTORY
Received 3 July 2017
Revised 7 December 2017 and 25 May 2018 Accepted 28 May 2018

KEYWORDS
Humanoid robot; evaluation of assistive devices; human-like motion

1. Introduction

Humanoid robots are designed to have a structure similar to that of a human so that they can move like humans. These robots are used extensively for mechanical simulation of the human body. A similar application is traditionally found in automotive crash-testing experiments, in which a humanoid dummy is used to obtain quantitative data such as the impact of a collision on the dummy. A few studies have used humanoid robots instead of human subjects to evaluate products intended for human use. Takanishi et al. developed the WABIAN-2 humanoid robot [1] for use as a dynamic simulator to test a walking support device during locomotion [2]. Nelson et al. developed the PETMAN robot [3] designed to reproduce a large range of human motion for testing chemical-protective clothing. Recently, many types of assistive devices such as wearable support devices have been developed, especially in Japan, to cope with the coming demographic shift to an older society. Several such devices have already been commercialized and are used, for example, by patients in rehabilitation facilities or to assist workers engaged in labour-intensive tasks in the transportation field [4–6]. Though these devices are gradually spreading throughout society, a quantitative measurement scheme for their supportive effects remains necessary. Because no standard for assistive devices has been formulated yet, such a quantitative evaluation scheme would clarify the differences in each device's features and allow users to choose devices based on their needs. For this purpose, Miura et al. and Ayusawa et al. introduced an evaluation framework for wearable assistive devices [7, 8]. This evaluation framework relies on the advantages of a humanoid robot for testing. Robots provide quantitative data from internal sensors, reproduce motions with high repeatability throughout the experiment, and do not require institutional review board approval, unlike experiments with human subjects. The framework was realized by utilizing the key technology of motion retargeting [9], which allows a humanoid robot to reproduce human-like motions. The effectiveness of the evaluation framework was tested through some experiments with a passive assistive device called the 'Smart Suit Lite' [7, 8, 10]. In recent years, many active assistive devices [4, 5] with actuators that efficiently support the human body have been developed. When evaluating such active devices, it must be ensured that the humanoid robot avoids conflicts between the actuator torques of the robot joints and those of the assistive

CONTACT Takahiro Ito s1730170@s.tsukuba.ac.jp
*This paper is selected as the "Cutting Edge of Robotics in Japan" by the Editorial Committee of Advanced Robotics.

device. This issue is attributed to the fact that many methods of motion retargeting handle only the kinematic features of human motion. An impedance control scheme for a whole-body humanoid robot [11] and a torque-controlled humanoid robot [12] that can avoid the torque conflict issues has been developed recently. However, reproducing human motions by using a humanoid robot wearing an active assistive device is quite difficult unless data about how the human body generates and controls internal forces during the entire movement is captured. A feasible way of evaluating active assistive devices is estimating the device's supportive effect in quasi-static contact scenarios between the human body and the device [7, 8]. Alternatively, we can estimate torque by performing inverse dynamics computations using human models [13]. The latter method is difficult to apply to the evaluation of a wearable assistive device because it requires detailed models of the assistive device and the contact condition between a human and the device. In this study, we introduce an evaluation framework for active assistive devices by building upon related works. The novelty of the proposed method lies in its capacity for quantitative evaluation by closely reproducing the scenarios in which humans use powerful active assistive devices with feedback data from the robot's sensors.

This research aims to extend the previously reported evaluation framework for application to active assistive devices. We introduced a new evaluation method to estimate the supportive torque of the active assistive devices called 'stationary torque replacement' in reference [14]. This method can be used to measure the static supportive torque supplied by the devices when a humanoid robot reproduces human postures. In the proposed method, the humanoid reproduces the zero-torque state when using an assistive device under the assumption that human muscles are relaxed during motion with the device. In addition, we demonstrate the validity of this evaluation scheme through measurements of human muscle activities. These tests confirm the hypothesis that the use of muscle power is decreased during supported motion.

The remainder of this paper is organized as follows. First, we explain the scheme for the evaluation of the static supportive torque of devices in Section 2. A specific procedure for estimating the supportive torque is addressed after explaining our method for reproducing human postures. Then, in Section 3, human muscular activity is measured for validating humans' actual usage of a device. In Section 4, we present experiments involving an active assistive device called 'Muscle Suit' and estimate the supportive torque it provides by using the HRP-4 humanoid robot. Section 5 concludes paper by discussing the feasibility and effectiveness of the proposed method.

2. Scheme for the evaluation of active assistive devices

In this section, the proposed evaluation scheme to estimate the effect of active assistive devices is introduced. Many types of assistive device have been developed, some of which are already commercially available. In this study, we focus on active assistive devices that support the lower back of humans because these are the most frequently commercialized devices [15–17]. The device used herein is called 'Muscle Suit'. The proposed analytical framework is composed of two parts, as shown in Figure 1. In the first step, human motions while wearing the assistive device are recorded using a motion-capture system. Then, the human motion trajectory is converted into a humanoid trajectory to extract typical postures when using the device. In the third step, the supportive torque supplied while maintaining the extracted postures is estimated by following the stationary torque replacement mathematical process, which will be described below. Before that, the motion-retargeting process is explained.

2.1. Motion-retargeting method for humanoid robot

In the proposed framework, typical postures that are encountered when a human subject uses an assistive device need to be imitated by a humanoid robot. Because a sufficient degree of similarity between robot and human motions is necessary for our evaluation process to be successful, we use the motion-retargeting method proposed in [9]. In this method, a humanoid robot reproduces whole-body human motions as closely as possible. The motions of a human subject are first recorded using a motion-capture system and are converted into motions that can feasibly be reproduced by a humanoid robot. Because the body structures of the robot and humans are somewhat different, execution of the method involves solving a simultaneous optimization problem that comprises three subproblems: (a) inverse kinematics problem to compute the joint angle trajectories of a human model when archiving the measured data, (b) problem of identifying the morphing function between the human and robot models, and (c) motion-planning problem of the robot considering physical constraints such as balance. The method can automatically reproduce the motion of a humanoid robot based on cost and mapping functions. In this study, the humanoid motion is computed using motions captured from human subjects wearing the device. In the computations, the wearable assistive device is assumed to be fixed to the human body and is modelled such that the device mass is concentrated at points located on body coordinates. Because we focus on

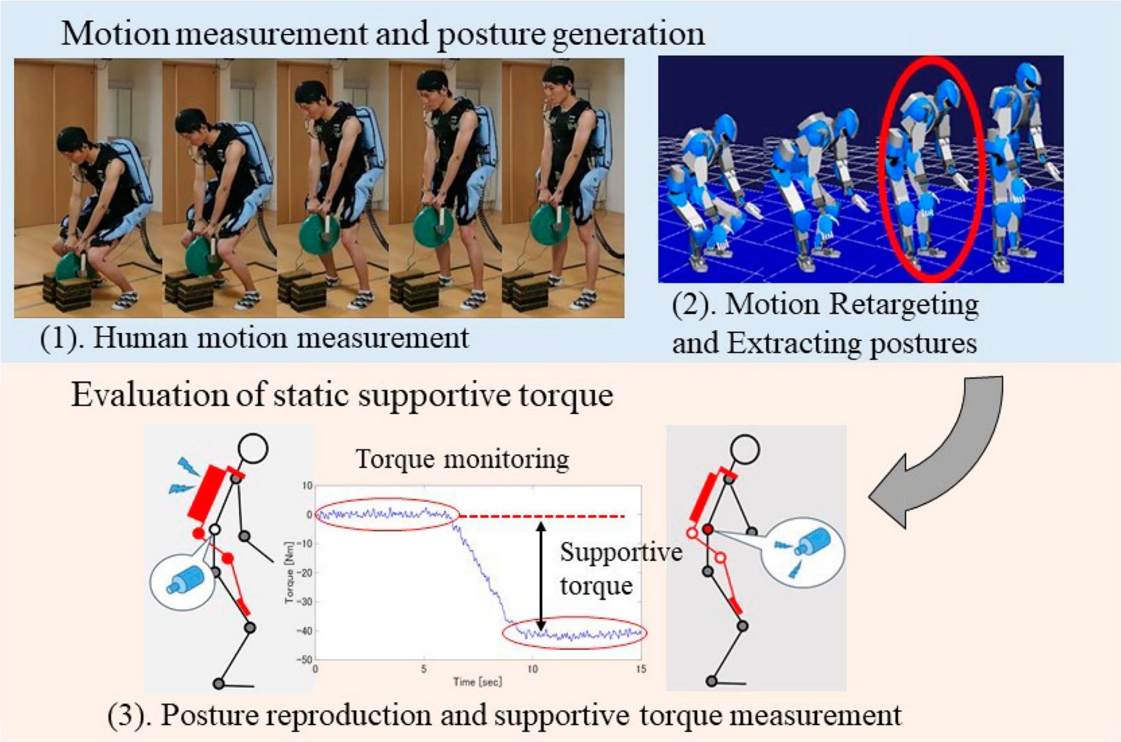


Figure 1. Overview of the proposed evaluation framework. (1) Trajectory of human motion when using the assistive device is captured using a motion-capture system. (2) Typical postures when using the device are extracted from the humanoid motion trajectory generated by the motion retargeting method. (3) The supportive torques of device are estimated by comparing robot joint torques with and without device support while the robot reproduces the extracted postures.

slow motions such as lifting a heavy object with assistance of the device, the typical postures are extracted to avoid torque conflict problems, as mentioned earlier. The postures are finally used to estimate the static supportive torques supplied by the assistive device, which will be explained in the next section.

2.2. Proposed method for estimating static supportive torque

The static supportive torque is estimated by activating the actuators of either the device or the humanoid robot in turn as the robot maintains a specific posture. The equation of static equilibrium (Figure 2) of the robot when wearing the assistive device can be expressed as

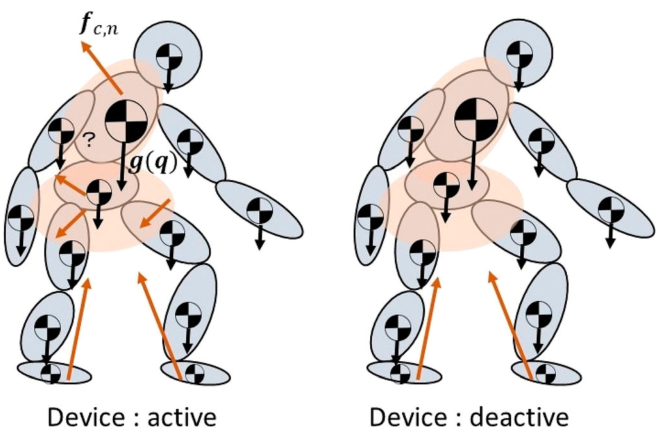


Figure 2. Internal state of robot when supportive force is activated or not activated while wearing device.

$$g(\boldsymbol{\theta}) = \boldsymbol{\tau}_{\mathrm{Joint}} + \boldsymbol{\tau}_c, \tag{1}$$

where

$g(\boldsymbol{\theta})$ is torque due to gravity and weights (which is a function of the generalized coordinate $\boldsymbol{\theta}$),
$\boldsymbol{\tau}_{\mathrm{Joint}}$ is the joint torque generated by the motors and
$\boldsymbol{\tau}_c$ is the torque due to external forces.

In Equation (1), the right-hand side represents the stationary torque constructed of internal and external system to formulate the static equilibrium conditions.

The external torque $\boldsymbol{\tau}_c$ can be split into torques from contact forces from the environment and the supportive force applied by the assistive device. Because we focus on devices for supporting the upper body as mentioned in Section 2, no contact force is supplied from the environment because we assume that the device is to be worn while standing up. The external torque $\boldsymbol{\tau}_c$ then only includes the force applied by the assistive device $\boldsymbol{f}_{\text{spt}}$.

$$\boldsymbol{\tau}_c = \boldsymbol{\tau}_{\text{spt}}(= \boldsymbol{J}_{\text{spt}}^{\text{T}} \boldsymbol{f}_{\text{spt}}), \tag{2}$$

where τ_{spt} is the supportive torque supplied by the device and $\boldsymbol{J}_{\text{spt}}$ are the Jacobian matrices of each contact point. Because $g(\theta)$ is constant, the right-hand side of Equation (1) has redundancy. We can take advantage of this fact to estimate the supportive torque τ_{spt} by measuring the robot joint torque τ_{Joint} equation (1) from the current sensors in each joint, as follows:

(i) The humanoid reproduces the given posture while wearing the assistive device. Let i be defined as the index of the joint that is mainly supported by the device. The servo controller of the joint i is deactivated so that the torque at joint i is zero.

(ii) The assistive device is activated and starts supporting the user. The supportive force from the device is gradually increased until the target angle at joint i is reached. In this state, the humanoid robot reproduces the entire joint angle while being completely supported by the assistive device at joint i. When the joint coordinates of the robot are $\boldsymbol{\theta}^{(1)}$, the current state of the robot can be expressed using Equations (1) and (2):

$$\boldsymbol{g}(\boldsymbol{\theta}^{(1)}) = \boldsymbol{\tau}_{\text{Joint}}^{(1)} + \boldsymbol{\tau}_{\text{spt}}^{(1)}. \tag{3}$$

(iii) The joint torque is then activated while maintaining the state of stationary torque in Equation (3) with proportional differential control. When the supportive torque gradually decreases, the joint torque increases. By applying $\boldsymbol{f}_{\text{spt}} = \boldsymbol{0}$, when the coordinates of the robot are $\boldsymbol{\theta}^{(2)}$ in Equations (1) and (2), we have the following:

$$\boldsymbol{g}(\boldsymbol{\theta}^{(2)}) = \boldsymbol{\tau}_{\text{Joint}}^{(2)}. \tag{4}$$

(iv) The supportive torque can be estimated as the change in joint torques between steps (ii) and (iii). Assuming that the difference in joint positions is negligible, which means the stationary torque holds its value through the above steps, from Equations (3) and (4) with $\boldsymbol{\theta}^{(1)} \approx \boldsymbol{\theta}^{(2)}$, we finally obtain the supportive torque from the following relationship:

$$\boldsymbol{\tau}_{\text{spt}}^{(1)} = \boldsymbol{\tau}_{\text{Joint}}^{(2)} - \boldsymbol{\tau}_{\text{Joint}}^{(1)}. \tag{5}$$

At joint i, the supportive torque $\tau_{\text{spt}}^{(1)}$ is not determined uniquely owing to the redundancy in Equation (3). Because we assume that the torque at the supported joint is equal to zero in actual human usage, we define $\tau_{\text{Joint}}^{(1)}$ as being equal to be zero in case of the robot. Based on this assumption, the supportive torque $\tau_{\text{spt}}^{(1)}$ can be computed as $\tau_{\text{spt}}^{(1)} = \tau_{\text{Joint}}^{(2)}$. The zero-torque state assumption in the human case is detailed in Section 3.

In Equation (3), at step (ii), the stationary torque is determined only by the supportive torque. By contrast, the value is replaced with the joint torque in Equation (4) at step (iii). Because the stationary torque is constant through all steps, Equation (5) holds at step (iv). Consequently the supportive torque $\tau_{\text{spt},i}$ is equivalent to the joint torque $\tau_{\text{Joint},i}$ measured by the robot sensor.

3. Validating torque evaluation by measuring human actual usage

Section 2 introduced the supportive torque estimation scheme. Now, we summarize the motion-measurement process used to gather data about human movement in the evaluation process. First, human motions are captured to generate humanoid motions by using the motion-retargeting method presented in Section 2.1. As mentioned in the previous section, our framework evaluates the supportive effect when the robot's joint is not actuated and is fully supported by the device. Because the joint torque at the supported joint is supposed to be equal to zero, for the calculation to work, our evaluation method needs to know the postures resulting in zero torque at the supported joint. In case of the human body, this zero-torque state corresponds to the low muscle activity in the part of the body that is supported. In this section, we verify this correspondence between the zero-torque state of the robot and the relaxed muscle postures in humans performing lifting motions with the support of an assistive device.

3.1. Experimental environment of human motion measurement

We evaluate the 'Muscle Suit' active assistive device herein, which supports the lower back with pneumatic actuators. The device is shaped like a backpack and is attached to the human body with a belt at the waist and soft pads at the thighs. When the pneumatic actuators on the device are contracted by supplying compressed air, the device powerfully lifts the upper body.

The specific structure of Muscle Suit is shown in Figure 3. The device contains two joints around the waist joint to allow the human user to move naturally.

A snapshot of the motion-capture experiment is shown in Figure 4. Motions performed using the assistive

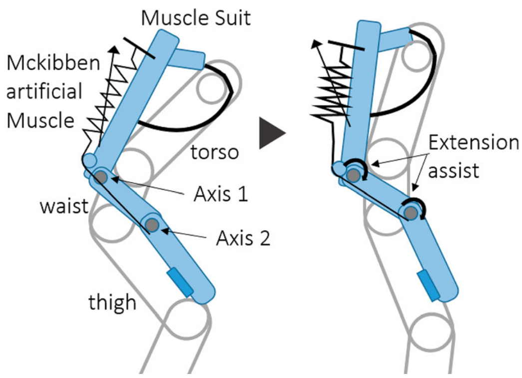


Figure 3. The specific structure of muscle suit.

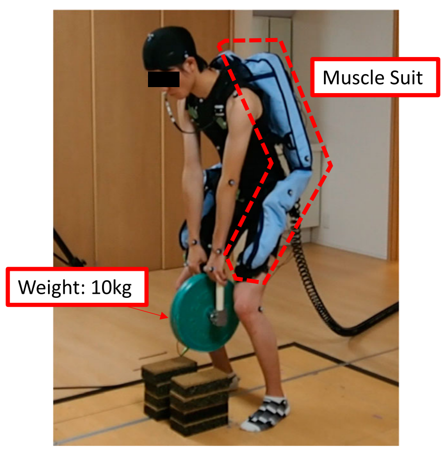


Figure 4. Snapshot of experiment of human motion during equipment measurement; a human lifting a 10-kg weight by using 'Muscle Suit'.

device were recorded using a motion-capture system (Motion Analysis Corp., sampling rate: 200 Hz).

Because Muscle Suit was designed for supporting motions involving lifting objects, we measured the human subject crouching down, holding a 5-kg weight, and then lifting it. We measured the same motion with a 10-kg weight as well. During the motion, we measured surface electromyogram (EMG) signals (DELSYS, sampling rate: 1000 Hz) from the lower back (erector spinae in the lumbar region), as shown in Figure 5. These EMG signals indicate muscular activity.

3.2. Human muscle activity analysis with muscle suit

The EMG readings indicate relative muscle activity. Therefore, we estimated the torque from EMG data with a human model to clarify the supportive effect of the device. Note that torque estimation based on the human model in this research does not rely on the specific model of the device or the contact points between the device and

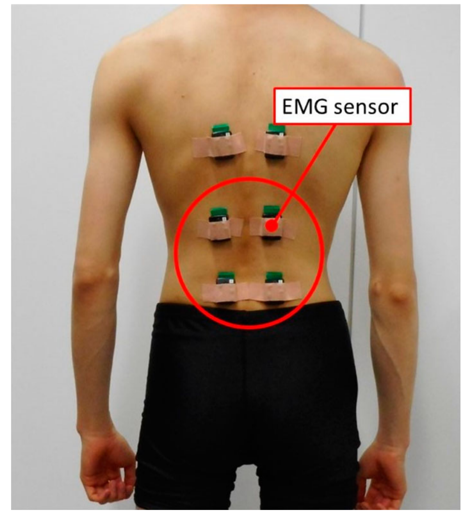


Figure 5. Placement of electromyography (EMG) measuring equipment.

the user. The device mass is only considered as a mass point at the corresponding position.

The human body is often modelled as a multi-body system in a manner similar to the modelling of a humanoid robot [13]. The human joint torque at the lower back can also be formulated as Equation (1). Let us reintroduce the equation as follows:

$$\boldsymbol{g}(\boldsymbol{\theta}) = \boldsymbol{\tau}_{\text{Joint}} + \boldsymbol{\tau}_c. \tag{6}$$

In the human motion analysis, $\boldsymbol{g}(\boldsymbol{\theta})$ can be estimated through inverse dynamics computations of a human skeletal model [13] from the captured data. When the assistive device is attached to a human body, it is difficult to distinguish the supportive torque from the human joint torque owing to redundancy of the contact forces between the body and the device. One approach to solve this issue is estimating joint torque from human EMG data using physiological models. The estimation procedure is as follows:

(1) Human motions with and without the device are recorded, respectively, by using the motion-capture system and the EMG recordings.
(2) $\boldsymbol{g}(\boldsymbol{\theta})$ is calculated by performing inverse dynamics computation [13] in the scenario that the subject does not utilize the device.
(3) EMG signals are converted to integral electromyogram (IEMG) signals that are helpful for inferring muscular activity. τ_{Joint} is computed using the dynamics of the joint driven by several muscles as follows:

$$\tau_{\text{Joint}} = \sum_{i=1}^{n} e_i F_i l_i, \tag{7}$$

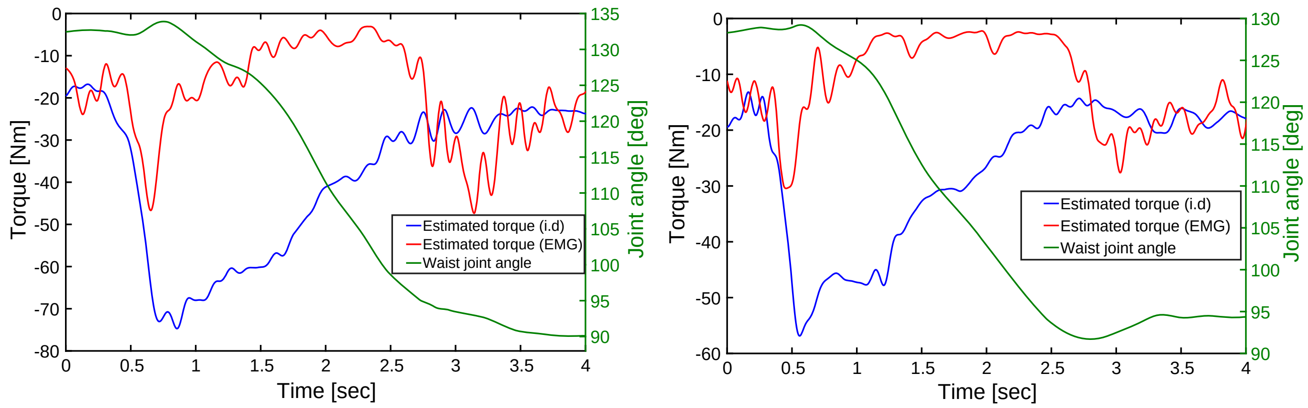


Figure 6. Torque estimated by means of inverse dynamics computation and EMG signals with human motion data (when using assistive device). (a) Estimated torque with 10 kg weight; the blue line shows $g(\boldsymbol{\theta})$ estimated by inverse dynamics computation (i.d.), red line shows the τ_{Joint} estimated using Equation (7) with IEMG signals, and green line shows the waist joint angle. (b) Estimated torque with 5 kg weight; the blue line shows $g(\boldsymbol{\theta})$ estimated by inverse dynamics computation (i.d.), red line shows the τ_{Joint} estimated using Equation (7) with IEMG signals, and green line shows the waist joint angle.

where e_i is a dimensionless IEMG [18] value, which denotes the activity of muscle i (time-series data). e_i is generated from the EMG data once it is rectified and a band-pass filter is applied (lower cutoff frequency: 20 Hz, upper cutoff frequency: 400 Hz, interval of integration: 100 ms); then, it is normalized by the maximum value, resulting in a e_i range of 0 to 1. F_i is the maximum muscle tension, and l_i is the moment arm of the lower back. In this study, the muscle tensions and moment arms were regarded as unknown constants. According to Equations (6) and (7) with $\tau_c = 0$, $F_r l_r$, $F_l l_l$ were identified from EMG signals e_i and $\boldsymbol{g}(\boldsymbol{\theta})$ by using the least squares method in the case of motion data obtained without the device.

$$\begin{bmatrix} g(\boldsymbol{\theta})_{t=0} \\ \vdots \\ g(\boldsymbol{\theta})_{t=\text{end}} \end{bmatrix} = \begin{bmatrix} e_{\text{r},t=0} & e_{\text{l},t=0} \\ \vdots & \vdots \\ e_{\text{r},t=\text{end}} & e_{\text{l},t=\text{end}} \end{bmatrix} \begin{bmatrix} F_{\text{r}} l_{\text{r}} \\ F_{\text{l}} l_{\text{l}} \end{bmatrix}. \quad (8)$$

(4) The joint torque τ_{Joint} when wearing the device is estimated from the identified values $F_{\text{r}} l_{\text{r}}$, $F_{\text{l}} l_{\text{l}}$ and measured EMG signals using Equation (7). $\boldsymbol{g}(\boldsymbol{\theta})$ is also calculated as well for comparison with τ_{Joint}.

We tested this experimental method with a weight-lifting motion by using two different weights (5 and 10 kg). We measured EMG signals at two positions on the lower back (right and left erector spinae muscles). In Figure 6, we show the estimated torque at the human lower back (for weight of 5 and 10kg). The blue line shows the result of $g(\boldsymbol{\theta})$ estimated by inverse dynamics computation, and the red line shows the τ_{Joint} estimated using Equation (7) with IEMG signals during the weight-lifting motion when using the device.

The motion lasted approximately 4 s. The subject started lifting the weight at around 0.5 s and completed the motion at around 3.5 s, as in Figure 6. Although the torque computed from EMG data increased at the beginning of the motion at around 0.5 s and at the end of motion at around 3.0 s, we observe that Muscle Suit drastically reduced the human joint torque between about 1 and 2.5 s. The joint was fully supported by the device, and the joint torque was nearly equal to zero, which validates the stationary torque replacement method. In Figure 6, the ratios of torques from the EMG signals normalized against the maximum torque from inverse dynamics computation are 0.1186 ± 0.0605 (10 kg) and 0.0638 ± 0.0239 (5 kg). Given that negligible torque is observed in the period from the start of the motion to the end of motion, we can apply our method to the postures captured over this duration. Though these normalized values are small, we need to investigate the assumption that we can ignore such small torques when evaluating assistive devices with a humanoid robot. The effect of this assumption on the accuracy of estimating supportive torques will be discussed in Section 4.2.

4. Experiments for estimating supportive torque of assistive device

In this section, our evaluation of an active assistive device with the proposed method is presented. The experiment was conducted with the humanoid robot 'HRP-4' [19] and Muscle Suit, which is mentioned above. The proposed framework requires that the humanoid robot has a geometric structure similar to that of a human. HRP-4 is among the most suitable robots for this experiment because its geometric structure is designed according to anthropometric data of average young Japanese females

Figure 7. Humanoid robot HRP-4 and robot wearing Muscle Suit.

Table 1. Comparison of mass properties between human and HRP-4.

Body link	HRP-4 (%)	Human (%)
HEAD	1.6	6.7
ARM	9.8	4.0
TORSO	20.0	30.4
Upper body	43.3	45.1
WAIST	17.7	14.6
LEG	20.5	20.1
Lower body	58.7	54.8

[20]. An overview of HRP-4 is shown in Figure 7. For discussing the differences in the kinematic properties of human users and humanoid HRP-4, we compare the weight of each body link from the robot model data and the human body segment inertial parameters reported by Dumas et al. [21]. The ratio of the weight of each body link of HRP-4 is calculated and compared with the data of females from the Dumas database, as shown in Table 1. From Table 1, although the ratio of each body link of HRP-4 is not similar to that of humans, the ratio of the entire upper and lower body is similar to that of humans. The robot used herein has 37 degrees of freedom (each arm: 9, each leg: 7 including one at the toe, chest: 3, neck: 2), which means it can reproduce human-like motions. Because our focus is the waist joint, which Muscle Suit mainly supports, the robot provides reliable torque measurements of the kinematic properties of the entire trunk.

Though the standard HRP-4 robot is fitted with a hard cover, we replaced it with a soft fabric to which several types of wearable devices can be attached. In Figure 7, we show HRP-4 wearing Muscle Suit; no additional attachment is required, and the device can be installed on the robot as it can be on humans. The weight to be lifted is attached near the chest joint of the robot.

4.1. Static supportive torque estimation with reproduced postures

The supportive torque was estimated based on measurements of the stationary torque, as detailed in Section 2.2. First, the robot posture was generated from the captured human motion data, which are discussed in Section 3. The stationary torque in Equation 5 can be computed based on the actuator torque readings, which can be obtained from the motor current sensor of the robot. Because Muscle Suit is a powerful assistive device that can fully support the gravity torque of the waist joint, we measured the supportive torque when the joint torque at the waist joint was zero.

Because each joint torque of HRP-4 is measured by the motor current sensor, the torque at the waist joint under equilibrium can be computed as follows:

$$\tau_{\mathrm{waist}} = K_\tau i_{\mathrm{m}} - \tau_{\mathrm{f}}. \tag{9}$$

- K_τ is torque constant;
- i_{m} is motor current of the waist joint;
- τ_{f} is static friction torque of the waist joint.

Although the static friction torque remains undetermined, our method can ignore the friction torque by considering the difference in joint torque $\tau_{\mathrm{spt}} = K_\tau (i_{\mathrm{m}}^{(1)} - i_{\mathrm{m}}^{(2)})$ in Equation 5 under the assumption that the friction condition is unchanged.

The servo system at the waist joint was deactivated to realize the zero-torque state while the device fully supported the joint (Figure 8a). Before reducing the supportive force of the device, the servo control system was activated to maintain the same posture. Finally, the supportive torque of the device was replaced with the actuator torque of the robot after the device was completely deactivated (Figure 8b).

The three postures labelled in Figure 9 were extracted from the captured human lifting motion. Posture 1 is the initial posture when the lifting motion is about to begin, posture 3 represents the end of the lifting motion, and posture 2 is the intermediate position between postures 1 and 3. According to the results in Section 3, we extracted the postures from the range of motion that can be estimated reliably. The range in which human muscular activity was quite small because sufficient support is provided by the device. The results of supportive torque measurements in posture 1 with the 10 kg weight are shown in Figure 10. In the figure, the waist joint torque is approximately 0 Nm from 0 to 6 s. Then, it starts to increase. Meanwhile, the supportive effect decreases from 6 to 10 s. The waist joint torque finally reaches around 40 Nm after 10 s. We applied the same procedure to the other

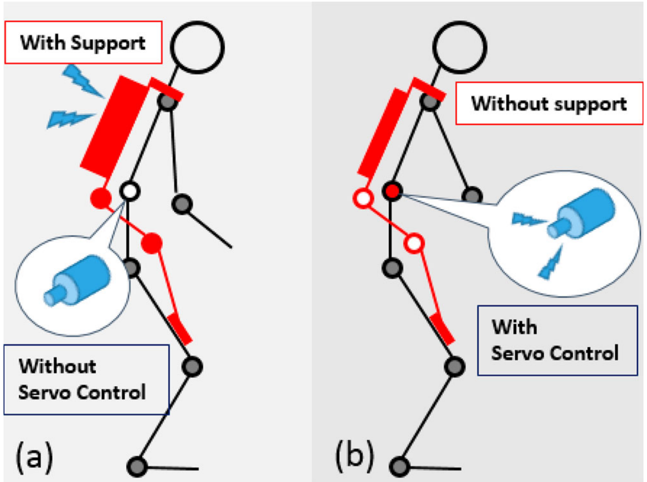


Figure 8. Supportive torque measurement scheme with servo control system.

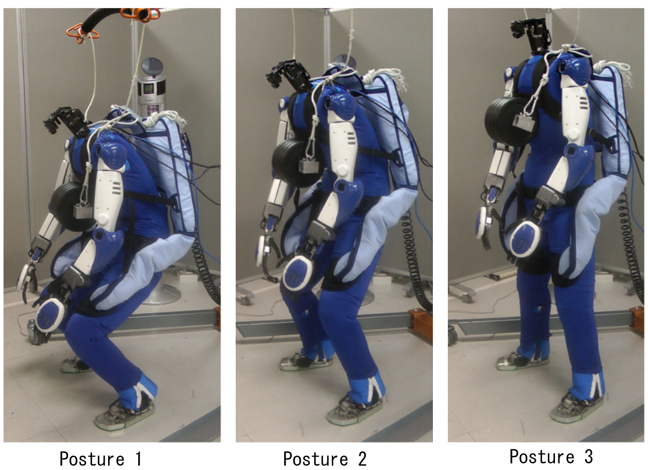


Figure 9. Static posture reproduction with HRP-4 and measurement of supportive torque with 10 kg weight.

two postures and estimated the supportive torques. The experiments with the 5 kg weight and no weight (0 kg) attached on the robot were conducted in the same manner as that in the case of the 10 kg weight. The estimated supportive torques are listed in Table 2. In all experiments, the same measurement trial was repeated thrice, and each torque in Table 2 is the average value of the three trials.

4.2. *Validation of proposed evaluation scheme*

The results in Table 2 demonstrate the feasibility of the proposed method for estimating the static supportive torque supplied by an assistive device.

We first validated the reliability of supportive torque estimation. The value of supportive torque is estimated to be the same as that of gravity torque at the waist joint based on Equation (1). Because gravity torque can be formulated as a nonlinear function of joint angles, supportive torque should be determined relative to the robot posture. Under the assumption of a simple inverted pendulum model, which approximates the dynamics of the upper body, the supportive torque at the waist joint is expected to grow as the angle between the upper body and the vertical axis ϕ increases, as illustrated in Figure 11. The figure also shows each ϕ corresponding to the three postures with a 5-kg weight: 36.5° (posture 1), 30.4° (posture 2) and 22.2° (posture 3). According to the above results and Figure 11, there exists a correlation between ϕ and the estimated torque; from posture 1 to posture 2, the value of $\sin(\phi)$ increases by 1.29 times, and the torque increases by 1.22 times correspondingly, as it does in the transition from posture 2 to posture 3. The largest values of ϕ and supportive torque are observed in posture 1. The angles in the three postures while lifting the 10 kg and those with no weight are shown in Figure 11 as well. The same correlation between the angles and the supportive torques is apparent. The correlation between the supportive torque and the weight condition is observed with the same postures. In posture 1, because the value of $\sin(\phi)$ for each weight is almost the same, the increase in torque from 0 to 5 kg is almost the same as that from 5 to 10 kg. Although this correlation is not observed in posture 2 owing to the large variation of $\sin(\phi)$, it is observed in posture 3. These results indicate that supportive torque could be obtained using proposed method reliably with different loads during the motion.

In Section 3, when a human subject was made to use the device, the joint torque at the lower back was small. By contrast, in our framework, the joint torque of the

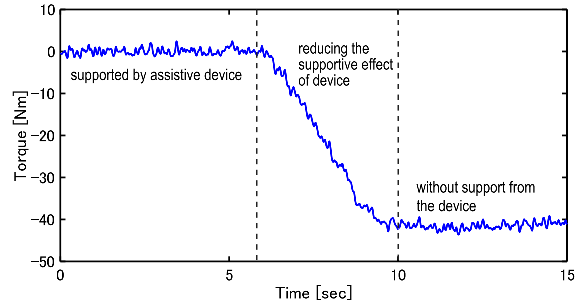
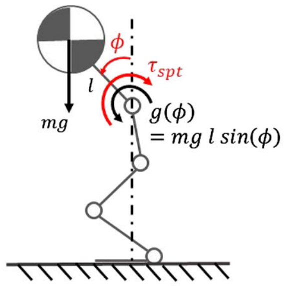


Figure 10. Torque measurement result in posture 1 with 10 kg weight.

Table 2. Results of supportive torque measurements with Muscle Suit.

Weight	Posture 1	Posture 2	Posture 3
0 kg	19.0 ± 1.2	16.2 ± 1.7	12.9 ± 2.5
5 kg	32.9 ± 1.4	27.0 ± 1.3	23.1 ± 0.7
10 kg	41.2 ± 1.1	42.4 ± 2.1	32.6 ± 3.7

Unit: [Nm].



5kg	Posture 1	Posture 2	Posture 3
ϕ [deg] ($\sin(\varphi)$)	36.5 (0.59)	30.4 (0.46)	22.2 (0.38)
		× 1.29	× 1.21
τ_{spt} [Nm]	32.9	27.0	23.1
		× 1.22	× 1.17

0kg	Posture 1	Posture 2	Posture 3
ϕ [deg] ($\sin(\varphi)$)	39.1 (0.63)	35.2 (0.58)	26.4 (0.44)
		× 1.17	× 1.26
τ_{spt} [Nm]	19.0	16.2	12.9
		× 1.09	× 1.29

10kg	Posture 1	Posture 2	Posture 3
ϕ [deg] ($\sin(\varphi)$)	41.6 (0.66)	42.6 (0.68)	20.9 (0.37)
		× 0.97	× 1.78
τ_{spt} [Nm]	41.2	42.4	32.6
		× 0.98	× 1.30

Figure 11. Relation between supportive torque τ_{spt} and waist joint angle with respect to vertical axis from floor in case of adding no weight (0 kg), 5 kg weight and 10 kg weight. The rates of increase are shown between each value with red colour. They show that the increase in torque corresponds to the increase in angle.

robot is forced to be zero while estimating the supporting torque. We now discuss the effect of this assumption on the accuracy of our estimates. Consider the joint torque of the robot corresponding to such a small torque measured from the human subject. There are, of course, differences in body dimensions between the human subject and the humanoid robot. To validate the influence of the fluctuations in human muscle activity during the supported motion, we mapped the human joint torque to that of the humanoid robot with the following procedure. First, the mean value of the torque estimated via EMG during the lifting motion data in Section 3.2 is normalized against the maximum torque estimated by means of inverse dynamics computations. Then, this normalized value is applied to the maximum supportive torque measured in the humanoid robot experiment. The mapped results are as follows: 2.1 $\pm$ 0.7863 Nm (5 kg) and 5.0 $\pm$ 2.6 Nm (10 kg). Compared to the standard deviation of the estimated torques, as summarized in Table 2, the mapped values are of the same order of magnitude. This implies that even if we assume that the joint torque of the robot is zero while estimating the supportive torque, this assumption has only a minor influence on the accuracy of the proposed framework.

5. Conclusion and future work

In this paper, we presented a quantitative evaluation method called 'stationary torque replacement' for testing active wearable assistive devices by using a humanoid robot. The proposed method helps evaluate the effect of such devices in terms of the supportive torque supplied by the device when the robot reproduces a human posture while wearing the device. We showed that the supportive torque provided can be estimated by replacing the supportive torque with the joint torque of the robot in static equilibrium. Assuming human muscles are relaxed when using the device, we demonstrate that the proposed method can reliably estimate the supportive torque when the robot reproduces the same posture in equilibrium. Experimental measurements of human muscle activity during the motions are supported by the assistive device.

The feasibility of the proposed method was validated through experiments with the HRP-4 humanoid robot and the Muscle Suit active assistive device. In the experiment, supportive torques were estimated quantitatively for several postures generated from captured human lifting motion data. The results showed the expected correlation between the supportive torques and the reproduced postures, indicating that the torques can be estimated reliably. The accuracy of the estimation was validated by comparison with motion measurements recorded with a human subject. Although the torques on human joints are not zero when using the device, the error due to the joint torque of the robot is as small as the standard deviation of the torque estimation. This result indicates that the accuracy of the estimation is not significantly influenced by this source of error.

In future, the proposed method for evaluating the supportive effect will be extended to allow torque estimation in dynamical scenarios. For this purpose, the human motion analysis results reported in this study will be useful for constructing a torque-based control system that adapts humanoid robot motion to the external forces generated by active assistive devices. We finally aim to reproduce an entire human movement while using the device with the robot to evaluate the effectiveness of the method in estimating the torques with which an assistive device supports dynamic movements.

Acknowledgements

We express sincere gratitude to Dr. Yumeko Imamura of AIST for her support and advice for the measurement of human motion.

Disclosure statement

No potential conflict of interest was reported by the authors.

Notes on contributors

Takahiro Ito received the B.E. degree from the Ibaraki University, Japan in 2015, and the M.S. degree in system and information engineering from University of Tsukuba, Japan in 2017. He is currently in the second year of his Ph.D. in system and information engineering at University of Tsukuba, and CNRS-AIST Joint Robotics Laboratory, UMI3218/RL. His current research interests include evaluation of active assistive device by using humanoid robot, contact estimation on humanoid robot. He is a member of the IEEE and RSJ.

Ko Ayusawa received the B.S. degree in mechanical engineering, and the M.S. and Ph.D. degrees in mechano-informatics from the University of Tokyo, Japan, in 2006, 2008, and 2011, respectively. He worked with the Department of Mechano-Informatics at the University of Tokyo, as a postdoctoral researcher from 2011 to 2012, and as a project assistant processor in 2013. He is currently a researcher of Intelligent Systems Research Institute, National Institute of Advanced Industrial Science and Technology (AIST), Tsukuba, Japan, and a researcher of CNRS-AIST JRL (Joint Robotics Laboratory), UMI3218/RL. His research interests include identification of human/humanoid dynamics, motion control for humanoid robots, and kinematics and dynamics simulation for human musculoskeletal models.

Eiichi Yoshida received M.E. and Ph.D. degrees on Precision Machinery Engineering from Graduate School of Engineering, the University of Tokyo in 1993 and 1996, respectively. In 1996, he joined former Mechanical Engineering Laboratory, later reorganized as National Institute of Advanced Industrial Science and Technology (AIST), Tsukuba, Japan. He served as Co-Director of AIST/IS-CNRS/ST2I Joint French-Japanese Robotics Laboratory (JRL) at LAAS-CNRS, Toulouse, France, from 2004 to 2008. Since 2009, he has been serving as Co-Director of CNRS-AIST JRL (Joint Robotics Laboratory), UMI3218/RL, and appointed as Director in 2017. He also serves as Deputy Director of Intelligent Systems Research Institute (IS-AIST), AIST, Tsukuba, Japan since 2015. He was awarded Chevalier, l'Ordre National du Mérite from French Government for his long-term contributions to French–Japanese collaboration on robotics in 2016. He is a Senior Member of IEEE, and member of RSJ and JSME. His research interests include robot task and motion planning, human modeling, and humanoid robots.

Hiroshi Kobayashi was born in 1966. He received the B.S., M.S., and Ph.D. degrees in mechanical engineering from Tokyo University of Science, Japan, in 1990, 1992, and 1995, respectively. He received a Japanese society for the promotion of science (JSPS) Research Fellowships for Young Scientists as DC1 from 1992 to 1995 and JSPS Postdoctoral Fellowship for Research Abroad from 1996 to 1998 at AI Lab., University of Zurich. He was a lecturer at Tokyo University of Science in 1998, associate professor in 1999, and has been a professor since 2008. He is very much interested in not only unique research works but also real products for supporting human daily life physically and has 75 patents including 27 sold ones. He is a founder and CTO of venture company INNOPHYS Co., Ltd. established in 2013. He received more than 20 awards such as JSME Best Paper Award in 1996, 2005, 2010, 2014. He is a fellow member of the Robotics Society of Japan.

ORCID

Takahiro Ito http://orcid.org/0000-0003-2886-1067

References

[1] Ogura Y, Aikawa H, Shimomura K, et al. Development of a new humanoid robot WABIAN-2. In: Proceedings of the IEEE International Conference on Robotics and Automation; 2006. p. 76–81.
[2] Omer AMM, Kondo H, Lim HO, et al. Development of walking support system based on dynamic simulation. In: Proceedings of the IEEE International Conference on Robotics and Biomimetics; 2008. p. 137–142.
[3] Nelson G, Saunders A, Neville N, et al. PETMAN: A humanoid robot for testing chemical protective clothing. J Robot Soc Japan. 2012;30:372–377.
[4] Sankai Y. HAL: Hybrid Assistive Limb based on cybernics. In: Kaneko M, Nakamura Y, editors. Robotics research, Springer tracts in advanced robotics. Vol. 66, Berlin: Springer; 2011. p. 25–34.
[5] Kobayashi H, Aida T, Hashimoto T. Muscle suit development and factory application. Int J Autom Technol. 2009;3(6):709–715.
[6] Kong K, Jeon D. Design and control of an exoskeleton for the elderly and patients. IEEE/ASME Trans Mech. 2006;11(4):428–432.
[7] Miura K, Yoshida E, Kobayashi Y, et al. Humanoid robot as an evaluator of assistive devices. In: Proceedings of the IEEE International Conference on Robotics and Automation; 2013. p. 671–677.
[8] Ayusawa K, Yoshida E, Imamura Y, et al. New evaluation framework for human-assistive devices based on humanoid robotics. Adv Robot. 2016;30(8):519–534.
[9] Ayusawa K, Morisawa M, Yoshida E. Motion retargeting for humanoid robots based on identification to preserve and reproduce human motion features. In: Proceedings of the IEEE/RSJ International Conference on Intelligent Robots and Systems; 2015. p. 2774–2779.

[10] Imamura Y, Tanaka T, Suzuki Y, et al. Motion-based-design of elastic material for passive assistive device using musculoskeletal model. J Rob Mech. 2011;23(6): 978–990.

[11] Ott C. Cartesian impedance control of redundant and flexible-joint robots. Springer Tracts Adv Robot. 2008;49.

[12] Englsberger J, Wener A, Ott C, et al. Overview of the torque-controlled humanoid robot TORO. In: Proceedings of the IEEE-RAS International Conference on Humanoid Robots; 2014. p. 916–923.

[13] Nakamura Y, Yamane K, Fujita Y, et al. Somatosensory computation for man–machine interface from motion capture data and musculoskeletal human model. IEEE Trans Rob. 2005;21:58–66.

[14] Ito T, Ayusawa K, Yoshida E, et al. Stationary torque replacement for evaluation of active assistive devices using humanoid. In: Proceedings of the IEEE-RAS International Conference on Humanoid Robots; 2016. p. 739–744.

[15] Muscle Suit from Innophys Co., Ltd. [Internet]. Tokyo [cited 2017 Jun 25]. Available from: https://www.innophys.jp.

[16] HAL for lower back from CYBERDYNE Inc. [Internet]. Tsukuba [cited 2017 Jun 25]. Available from: https://www.cyberdyne.jp/products/Lumbar_LaborSupport.html

[17] AWN-03B from ATOUN Inc. [Internet]. Nara [cited 2017 Jun 25]. Available from: https://www.atoun.co.jp/products/atoun-model-a.

[18] Konrad P. The ABC of EMG: A Practical Introduction to Kinesiological Electromyography. USA: Noraxon. Inc.; 2005.

[19] Kaneko K, Kanehiro F, Morisawa M, et al. Humanoid robot HRP-4 – humanoid robotics platform with lightweight and slim body. In: Proceedings of the IEEE/RSJ International Conference on Intelligent Robots and Systems; 2011. p. 4400–4407.

[20] Kaneko K, Kanehiro F, Morisawa M, et al. Cybernetic human HRP-4C. In: Proceedings of the IEEE-RAS International Conference on Humanoid Robotics; 2009. p. 7–14.

[21] Dumas R, Cheze L, Verriest JP. Adjustments to McConville et al. and Young et al. body segment inertial parameters. Int J Biomech. 2007 Jan;40(3):543–553.