Admittance Control of Powered Exoskeletons
Based on Joint Torque Estimation

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ABSTRACT In recent years, exoskeletons have been widely accepted as rehabilitative and walking assistive devices for either healthy persons or patients with mobility impairment. Depending on mobility of the wearers, the control strategies for exoskeletons may differ significantly. This paper aims at developing an admittance control framework for exoskeletons to assist healthy persons in safer, faster, or more energy-efficient walking. A crucial step to accomplish assistive walking for healthy persons is to detect their intention. Direct sensing of the wearer’s biological signals, such as electromyography (EMG) and electroencephalography (EEG), requires additional sensors, which increases the cost and makes it inconvenient for the wearer to put on and take off the exoskeleton. Instead, we propose to detect the wearer’s intention by estimating the total torques applied from the wearer to the human-exoskeleton system based on the motor current and joint angles. Then the velocity commands to the joint motors are generated according to the estimated torque and a predefined mechanical admittance function of each joint. Consequently, the exoskeleton complies with the wearer’s intention. Rigorous theoretical analysis is performed on robust stability of the closed-loop system. Then experiments are carried out to verify the accuracy and robustness of the proposed torque estimation algorithm. In addition, experimental data show that the wearer’s gait can be shaped in a desired way by choosing an appropriate admittance function in the proposed admittance control loop.

INDEX TERMS Exoskeleton, walking assistance, intention detection, disturbance observer, compliance control, admittance control.

I. INTRODUCTION
Lower-limb exoskeletons are wearable mechatronic devices that provide power to assist either healthy wearers or walking disorder patients in standing, walking, and going up and down stairs [1]–[4]. The assistive power from exoskeletons allows wearers to walk more safely, easily, and swiftly. Therefore, walking assistive exoskeletons can benefit those who have to or feel difficult to walk for a long time, such as soldiers, foot patrol guards, the elderly, etc. To provide assistive forces or torques at the right timing with appropriate magnitudes, it is crucial to coordinate the motion between the wearer and the exoskeleton. In other words, the motion of the exoskeleton should comply with the wearer’s intention.

For lower extremity exoskeletons to comply with the wearers’ intention, existing approaches can be classified into three categories. The methods belonging to the first category detect the wearer’s walking intention by measuring biological signals of the wearer. For example, the exoskeleton named HAL (Hybrid Assistive Leg) [5], [6] and the exoskeleton developed in [7] measured the wearer’s electromyography (EMG) signals for intention detection; Gui et al. [8] proposed an adaptive method to estimate active joint torque by EMG signals. Since EMG signals come before contraction and relaxation of muscles, they can be used to predict the wearer’s intention of motion. However, it takes time to attach and detach EMG sensors to and from the wearer’s muscles. As the result, it causes inconvenience to the wearer in putting on and taking off the exoskeleton.

In addition to EMG, electroencephalography (EEG) signals were also investigated for intention detection. In [9]–[13], brain-computer interfaces (BCIs) for exoskeletons were implemented by measuring and decoding the wearer’s EEG signals. The decoding process includes a model that is trained offline to fit neural signals to actual movements; therefore, it is time consuming to adjust the model for each
individual user [3], and the reliability of the model is not guaranteed.

Instead of measuring biological signals to detect the wearer’s intention, the methods in the second category rely on delicately designed control algorithms to achieve the desired compliance. The control system of the exoskeleton named BLEEX (Berkeley Lower Extremity Exoskeleton) was designed to amplify the wearer’s movement by increasing its sensitivity function [14]–[16]. As a result, the exoskeleton becomes very sensitive to the wearer’s motion, and can be driven by just a tiny movement of the wearer. Apparently, high sensitivity to external inputs reduces the stability margin, indicating that the wearer’s response should be fast enough in order to stabilize the exoskeleton system.

On the other hand, Nagarajan et al. [17] designed a control law based on a linearized model to shape the mechanical admittance of the hip joint such that the wearer’s motion was amplified. The method is effective in providing assistive forces to the wearer through the admittance control law. However, the desired admittance was offline determined and cannot be online adjusted to meet different requirements of individual wearers for various types of gaits like walking, strolling, and striding. On the other hand, adaptive admittance control has been applied to rehabilitative upper-limb exoskeletons [18], [19] and collaborative robots [20]. These methods require pre-defined trajectories for the robot and feedback of interactive forces between human and the robot. Therefore, a force sensor is mounted on the end effector. However, these requirements may not be satisfied for lower-limb exoskeletons since it is difficult to identify a single contact point between the wearer and the lower-limb exoskeleton where a force sensor can be installed to measure the interactive forces. Moreover, pre-defined trajectories are useful for patients to train their neuromuscular systems repetitively, but may be restrictive for healthy wearers to adjust their gaits.

The methods in the third category reduce the stiffness of the actuators to attain compliance. Lv et al. [21] showed their design of a walking assistive exoskeleton with backdrivable motors and torque/force sensors. Zhang et al. [22] accomplished admittance shaping assistive control on a series elastic actuator (SEA)-driven robotic hip exoskeleton. Yu et al. [23] also used SEAs to realize force tracking control on their knee-ankle-foot robot. Although these methods endow exoskeletons with compliance, it is challenging to customize actuators of exoskeletons that simultaneously meet various stringent constraints on volume, weight, power consumption, heat dissipation, available output torque, and low stiffness.

All the aforementioned methods have their own strength and weakness. The best choice of control strategies should depend on the desired functionality and the target users of the exoskeleton. In this paper, the target users are healthy persons who can walk and keep balance by themselves. The exoskeleton designed here is aimed at assisting the target users to walk more safely, swiftly, and/or energy-efficiently. For these target users, we think that the methods based on intention detection are most promising because they provide direct information on the wearer’s intended locomotion. In this paper, instead of measuring biological signals of the wearer, we propose to detect the wearer’s intention by estimating the wearer’s torque applied to the human-exoskeleton system that consists of the exoskeleton and the wearer’s lower limbs. This can be accomplished by measuring joint angles, motor current and the ground reaction forces without any biological sensors. Then, the joint motor velocity commands are determined based on the estimated torque and a prescribed mechanical admittance function of the joint, while the motor’s velocity control loop guarantees that the actual joint velocity follows the command.

The proposed method renders a trajectory-free compliance control law. That is, the motion of the exoskeleton is driven completely by the torque from the wearer instead of following a pre-planned trajectory. Consequently, the exoskeleton is able to comply with the wearer’s intention and the joints possess the desired mechanical admittance. We would like to emphasize that trajectory-free admittance control is desirable for walking assistance of healthy persons because the wearers are able to adjust their gaits freely. Moreover, the joint admittance is on-line tunable by the control law, allowing the exoskeleton to adapt itself to different walking conditions for higher speeds, better safety, and higher energy-efficiency.

Then, we implement the proposed torque estimation and admittance control law on the exoskeleton made by Industrial Technology Research Institute (ITRI), Taiwan and carry out experiments to verify accuracy and robustness of the torque estimation algorithm. We also demonstrate by experiments that the healthy wearer’s gait can be shaped in a desired way by applying an appropriate admittance function.

This paper is an extended research of the authors’ previous work [24]. In this paper, we conduct rigorous theoretical analysis on the robust stability of the closed-loop system, and verify the accuracy and robustness of the torque estimation algorithm by experiments.

The remainder of the paper is organized as follows: Section II presents hardware configuration, dynamic model of the human-exoskeleton system, and identification of model parameters. In Section III, design of the torque estimator and admittance control law is introduced. Next in Section IV stability of the compliance control system is analyzed. Then in Section V experimental results of torque estimation and admittance control are presented. In Section VI, we conclude this paper and mention the future work.

II. HARDWARE CONFIGURATION AND MODELS

A. HARDWARE CONFIGURATION

The exoskeleton we use in this paper was made by Industrial Technology Research Institute (ITRI), Taiwan, and slightly modified by the authors for the research purpose (see the left part of Figure 1). The exoskeleton consists of four motors mounted in the hip and knee joints of both legs (see the middle part of Figure 1). Each motor’s driver is connected to the main controller, which is implemented on a Raspberry
Pi embedded computer, in the backpack through the CAN bus. Four force sensing resistors (FSRs) are attached to each leg at the locations shown in the middle and right part of Figure 1 to measure external forces applied to the thigh and shank. These FSRs are used in the experiments only for providing the ground truth to verify accuracy of torque estimation. We should point out that there is no need to use these FSRs in the proposed torque estimation algorithm. Detailed discussion of these FSRs will be presented in Section V.

B. DYNAMIC MODEL FOR THE HUMAN-EXOSKELETON SYSTEM (HES)

The two legs of the exoskeleton are considered to be independent and identical. For each leg, the ankle is a passive joint. As the exoskeleton is tightly attached to the wearer, the wearer’s lower limbs and the exoskeleton are treated as a whole system, which is called the human-exoskeleton system (HES) in this paper and is modeled as two identical two-joint planar manipulators powered by both joint motors and the wearer. Consequently, HES can be viewed as the biomechatronic legs of the wearer, which is controlled and balanced by the wearer’s primary motor cortex. Single leg of HES is shown in Figure 2, and its governing equation is given in (1).

\[
\mathbf{M}(q)\ddot{q} + \mathbf{C}(q, \dot{q})\dot{q} + \mathbf{G}(q) + \mathbf{B}(\dot{q}) = \tau_1 + \tau_h + J^T(q)\mathbf{F}_{GR}
\]

(1)

where \(\tau_1 = [\tau_{l, H}, \tau_{l, K}]^T\) is the joint torque applied from the motors to HES. \(q = [\dot{q}_{l, H}, \dot{q}_{l, K}]^T\) is the joint angle. The directions are defined in Figure 2. \(\dot{q}\) and \(\ddot{q}\) denote the joint velocity and acceleration, respectively. \(\tau_h = [\tau_{h, H}, \tau_{h, K}]^T\) is the torque exerted by the wearer to HES. \(\mathbf{F}_{GR} = [\mathbf{F}_{GR,x}, \mathbf{F}_{GR,y}]^T\) is the ground reaction force which is introduced to HES whenever the foot contacts the ground, while \(\mathbf{J}\) is the Jacobian matrix. The matrices and vectors \(\mathbf{M}(q), \mathbf{C}(q, \dot{q}), \mathbf{G}(q)\) and \(\mathbf{B}(\dot{q})\) are given as (2), as shown at the bottom of the next page. where \(I\) is the inertia, \(m\) is the mass and \(l\) is the length. \(g\) is the gravity acceleration. \(b\) and \(f\) are the viscous and Coulomb friction coefficients, respectively. Subscripts \(T, S, H\) and \(K\) denote the quantities associated with the thigh, shank, hip and knee, respectively. Subscript \(C\) means that the quantity is associated with the mass center. Note that these model parameters are associated with HES; therefore, their values depend on the particular wearer and are uncertain.

C. DYNAMICS OF EXOSKELETON AND MOTORS

The complete model of HES includes (1) as well as the joint motors and their associated drivers. Since both legs are identical, we take only one leg for example in this subsection. Let \(\mathbf{q}_m = [\dot{q}_{m, H}, \dot{q}_{m, K}]^T\) be the vector of motor angles. Note that \(\mathbf{q}_m = \mathbf{r}\mathbf{q}\), where \(\mathbf{r} = \text{diag}(r_H, r_K)\) is the gear ratio matrix. In this paper, both motors operate in the velocity control mode. In other words, the main controller sends velocity command, denoted by \(\dot{\mathbf{q}}_{mc} = [\dot{\mathbf{q}}_{mc,H}, \dot{\mathbf{q}}_{mc,K}]^T\), to the motor drivers. Then the velocity controllers built inside the drivers generate the motor current \(\mathbf{i}_m = [i_{m,H}, i_{m,K}]^T\) such that the motor velocity \(\dot{\mathbf{q}}_m\) follows the command \(\dot{\mathbf{q}}_{mc}\). The motion of the motor’s rotor is governed by the following equation:

\[
\mathbf{J}_m\ddot{\mathbf{q}}_m + \mathbf{B}_m\dot{\mathbf{q}}_m = \tau_m - r_l^{-1} \tau_1
\]

(3)

where \(\mathbf{J}_m = \text{diag}(J_{m,H}, J_{m,K})\) and \(\mathbf{B}_m = \text{diag}(B_{m,H}, B_{m,K})\) are the moment of inertia and damping coefficient matrices of the rotor. \(\tau_m = [\tau_{m,H}, \tau_{m,K}]^T\) is the vector of total motor torques, which is proportional to the motor current, i.e.

\[
\tau_m = K_l\mathbf{i}_a
\]

(4)

where \(K_l = \text{diag}(K_{l,H}, K_{l,K})\) is the motor constant matrix.

According to the manual of the driver, the velocity loop controller is equivalent to the following form:

\[
\mathbf{i}_a(s) = \mathbf{C}_{D2}(s)(\mathbf{q}_{mc}(s) - \mathbf{C}_{D1}(s)\dot{\mathbf{q}}_m(s))
\]

(5)

Note that we slightly abuse the notations in (5) to simplify the presentation by denoting \(\mathbf{i}_a(s)\) and \(\dot{\mathbf{q}}_m(s)\) as the Laplace transforms of \(\mathbf{i}_a(t)\) and \(\dot{\mathbf{q}}_m(t)\), respectively. Here \(\mathbf{C}_{D1}(s) = \text{diag}(C_{D1,H}(s), C_{D1,K}(s))\), \(\mathbf{i} = 1, 2\) denotes the equivalent controllers in the feedback and feedforward loop, respectively.
Combining (1) and (3)~(5), the complete block diagram HES (for one leg) is shown in part (A) of Figure 3, where we define:

\[
\begin{align*}
\tau_e &= P_a \dot{q} = M(q) \ddot{q} + C(q, \dot{q}) \dot{q} + G(q) + B(q) \quad (6) \\
\tau_s &= P_m \dot{q}_m = J_m \dot{q}_m + B_m \dot{q}_m \quad (7)
\end{align*}
\]

Note that \( P_a \) represents the dynamic model of HES (the left hand side of (1)) and is treated as a nonlinear operator that maps the joint velocity \( q \) to the torque applied to HES \( \tau_e \). Similarly, \( P_m \) is a linear operator that maps the motor velocity \( q_m \) to the torque applied to the rotor \( \tau_s \). It will be seen shortly that treating each block in Figure 3 as an operator. Those blocks in Figure 3 help simplifying the complex structure and derive the conditions on robust stability.

**D. SYSTEM PARAMETERS IDENTIFICATION**

As will be shown in the next section, the proposed torque estimation algorithm is a model-based method. Hence the model parameters of HES must be identified in advance. The values of the parameters associated with the motors and drivers can be found in their own spec and manual; therefore, only the parameters in (2) need identifying.

\[
M(q) = \begin{bmatrix}
I_s + m_s l_{SC}^2 + l_T l_{SC} \cos(q_K) \\
I_s + m_s l_{SC}^2 + l_T l_{SC} \cos(q_K)
\end{bmatrix} \alpha
\]

\[
C(q, \dot{q}) = \begin{bmatrix}
-m_s l_T l_{SC} \sin(q_K) \dot{q}_K \\
-m_s l_T l_{SC} \sin(q_K) \dot{q}_K
\end{bmatrix}
\]

\[
G(q) = \begin{bmatrix}
(m_T l_{TC} + m_s l_T) \sin(q_H) + m_s l_{SC} g \sin(q_H + q_K) \\
m_s l_{SC} g \sin(q_H + q_K)
\end{bmatrix}
\]

\[
B(\dot{q}) = \begin{bmatrix}
b_T \dot{q}_H + f_{TC} \text{sign}(\dot{q}_H) \\
b_s \dot{q}_K + f_{SC} \text{sign}(\dot{q}_K)
\end{bmatrix}
\]

\[
\alpha = I_T + I_s + m_T l_{TC}^2 + m_s (l_T^2 + l_{SC}^2 + 2l_T l_{SC} \cos(q_K))
\]

\[
\begin{align*}
\tau_e &= \tau_h + \tau_{GR} = \varphi(q, \dot{q}, \ddot{q}) \theta \quad (9)
\end{align*}
\]

Substituting (1), (4), (6), and (7) into (3) leads to

\[
\tau_h - \tau_{GR} = \varphi(q, \dot{q}, \ddot{q}) \theta \quad (8)
\]

Since we can obtain the motor current, joint angle, and joint velocity from the driver, and estimate joint acceleration according to the joint velocity, the left-hand side of (8) is known. On the other hand, we suppose that there is no torque nor force from the wearer and the ground, i.e. \( \tau_h = 0 \) and \( \tau_{GR} = 0 \). Then only \( \tau_{h} \) is left on the right hand side of (8). Furthermore, \( \varphi(q, \dot{q}, \ddot{q}) \) can be linearly parameterized based on (2) as follows:

\[
\tau_e = \varphi(q, \dot{q}, \ddot{q}) \theta = \varphi(q, \dot{q}, \ddot{q}) \theta = \tau_{GR} - \tau_h
\]

**TABLE 1. Identified parameter values of right leg.**

| Symbol | Parameters | Units | Value |
|--------|------------|-------|-------|
| \( p_1 \) | \( f_{EC} \) | mNm | 8342.99 (pos. direction) | 8873.57 (neg. direction) |
| \( p_2 \) | \( m_j l_{SC} g \) | mNm | 3785.07 |
| \( p_3 \) | \( I_s + m_s l_{SC}^2 \) | mNm/(1/s²) | 2804.41 |
| \( p_4 \) | \( b_l \) | mNm/(1/s) | 4915.77 |
| \( p_5 \) | \( m_s l_{SC} \) | mNm/(1/s¹) | 162.22 |
| \( p_6 \) | \( f_{EC} \) | mNm | 9321.29 (pos. direction) | 9475.27 (neg. direction) |
| \( p_7 \) | \( (m_l l_{RC} + m_s l_{TR}) g \) | mNm | 1826.22 |
| \( p_8 \) | \( (I_s + m_s l_{SC}^2 + m_s l_{TC}^2) \) | mNm/(1/s²) | 3000 |
| \( p_9 \) | \( b_r \) | mNm/(1/s) | 5000 |

**Remark:** We obtain (9) based on the assumptions that \( \tau_h = 0 \) and \( \tau_{GR} = 0 \), which can be satisfied by dangling the exoskeleton on a rack with no wearer in it; hence there is no contact with the ground. Under this condition, the identified parameters in TABLE 1 do not consider the effects of the wearer; these parameters are denoted as nomininal parameters. The difference between the actual and nominal models will be
explicitly taken into account in the robust stability analysis in Section IV, and the effects on torque estimation will be evaluated experimentally in Section V.

III. ADMITTANCE CONTROL SYSTEM

A. WEARER’S TORQUE ESTIMATION

We apply the idea of disturbance observer (DOB) [25], [26] to estimate the wearer’s torque \( \tau_h \). The block diagram of the proposed torque estimator is shown in part (B) of Figure 3. Namely,

\[
\hat{\tau}_h = Q \left[ rP_m q_m + \varphi(q, \dot{q}, \ddot{q}) \dot{\theta} - rK_i a - J^T(q)F_{GR} \right]
\]

where \( \hat{\tau}_h \) is the estimated wearer’s torque, \( \dot{\theta} \) is the nominal parameter, and \( F_{GR} \) is the ground reaction force measured by the force sensors on the feet. \( Q \) is a low pass filter with relative degree at least one. This filter renders a proper system \( QP_m \) and enhances robustness of the torque estimation by eliminating the high frequency components in \( \tau_h \).

B. ADMITTANCE CONTROL

The mechanical admittance (or simply admittance for short in this paper) of a mechanical device is the relation between its torque input and angular velocity output. The objective of admittance control is to shape the admittance of a device such that the admittance possesses desired characteristics. In this paper, admittance control is used to adjust the relation between the wearer’s torque and the angular velocity of each joint. Modifying the admittance function changes the wearer’s experience in walking. For example, an admittance function with a large gain amplifies the wearer’s movement (i.e., a small torque can induce fast movement of the joint), and consequently the wearer experiences an assistive torque from the exoskeleton. Conversely, an admittance function with a small gain introduces prohibitive torque to the wearer.

Since we have estimated the wearer’s torque, the desired angular velocity of the joint is

\[
\dot{q}_c = P_d \hat{\tau}_h
\]

where \( P_d \) is the desired admittance function to be discussed shortly. Then the desired joint motor velocity is \( \dot{q}_{mc} = r\dot{q}_c \), which is the velocity command to the motor. The actual joint velocity follows the desired one due to the high-performance velocity servo control built in the motor’s driver. The admittance control is depicted in part (C) of Figure 3.

\( P_d \) defines how the primary motor cortex of the wearer interacts with his/her biomechatronic legs, i.e. HES. In this paper, \( P_d \) is chosen as a first order system shown below

\[
P_d = (J_d s + B_d)^{-1}
\]

where \( J_d = \text{diag}(J_{d,H}, J_{d,K}) \) and \( B_d = \text{diag}(B_{d,H}, B_{d,K}) \) are the desired inertial and damping coefficients, respectively. \( P_d \) in (13) is a strictly positive real (SPR) system [27], implying that it is passive and dissipates energy. Passivity ensures stable locomotion as the wearer intends to walk and applies torques to HES. The exoskeleton plays a role of shaping the rate of energy consumption during walking by choosing appropriate admittance parameters \( J_d \) and \( B_d \). Moreover, the low-pass nature of \( P_d \) filters out sudden peaks of the estimated torques and delivers smooth joint velocities, thus enhancing safety and the user’s walking experience.

Note that the proposed torque estimation algorithm and the admittance control law constitute a loop around HES (see Figure 3). Therefore, stability of the closed-loop system in Figure 3 must be taken into account. We analyze the robust stability of the closed-loop system in the next section.

IV. ROBUST STABILITY ANALYSIS

In this section, we derive the conditions that guarantee robust stability of the closed-loop system in Figure 3. Before proceeding to the analysis, we briefly introduce the notion of norms for signals and systems that will be used in the derivation.

Let \( u : \mathbb{R}^+ \rightarrow \mathbb{R}^m \) be an \( m \)-DImENSONAL signal. the \( L_2 \) norm of \( u \) is defined as

\[
\|u\|_{L_2} = \left( \int_0^\infty |u(t)|^2 \, dt \right)^{1/2}
\]

where the norm of the integrand is the Euclidean norm of the \( m \)-dimensional vector \( u(t) \) at any time instance \( t \).

Let \( L_2^n \) be the set of all \( m \)-dimensional signals that have finite \( L_2 \) norm, and \( u_\tau \) be the truncation at time \( \tau \) of \( u \) for \( \tau \geq 0 \), i.e., \( u_\tau(t) = u(t) \) for \( 0 \leq t \leq \tau \) and \( u_\tau(t) = 0 \) for \( t > \tau \). Then the definition of a finite-gain \( L_2 \) stable system is given as follows.

Definition (27, Definition 5.1, p.197):

A mapping \( H : L_2^n \rightarrow L_2^n \) is finite-gain \( L_2 \) stable if there exist nonnegative constants \( \gamma \) and \( \beta \) such that

\[
\|H u_\tau\|_{L_2} \leq \gamma \|u_\tau\|_{L_2} + \beta
\]

for all \( u \in L_2^n \) and \( \tau \in [0, \infty) \).
Remark: The smallest $\gamma$ that satisfies (14) is called the gain of $H$, and is denoted by $||H|| = \gamma$.

A. ROBUST STABILITY OF THE CLOSED-LOOP SYSTEM

Define $\hat{\tau}_e = \hat{P}_a \dot{q} = \varphi(q, \dot{q}, \ddot{q})\hat{\theta}$, where $\hat{\theta}$ is the nominal parameter model in TABLE 1, and $\hat{P}_a$ is a nonlinear mapping from $\dot{q}$ to $\hat{\tau}_e$. Due to the nonlinearity of $P_a$ and $\hat{P}_a$, the robust stability analysis will be conducted in the time domain and therefore all blocks in Figure 3 are regarded as input-output mappings in the time domain. Moreover, the blocks associated with the motors and drivers, including $P_m$, $K_t$, $r$, $C_{D_1}$, and $C_{D_2}$, are linear time-invariant (LTI) and diagonal. Hence serial connections of these blocks are commutable. Although $P_d$ defined in (13) is also LTI and diagonal, it can be set as a nonlinear and non-diagonal system without affecting the closed-loop analysis in this section. Similarly, $Q$ is not restricted to be an LTI and diagonal system for stability analysis.

Because the ground reaction force $F_G$ is directly measured and cancelled out in (11), we neglect $F_G$ in the closed-loop analysis. Furthermore, for easy reference in the subsequent derivation, we define the following input-output mappings in (15) to (21):

$$A_1 = (I + r^{-2}P_m^{-1}P_a + P_m^{-1}K_tC_{D_2}C_{D_1})^{-1}$$
$$A_2 = P_m^{-1}K_tC_{D_2}P_d$$
$$B_1 = r^{-2}P_m^{-1}P_a + P_m^{-1}K_tC_{D_2}C_{D_1}$$
$$B_2 = P_m^{-1}K_tC_{D_2}P_dQ(P_a - \hat{P}_a)$$
$$B_3 = Q(P_a - \hat{P}_a)$$
$$K_1 = P_m^{-1}(r^{-2}+K_tC_{D_2}P_dQ)$$
$$K_2 = P_d^{-1}C_{D_2}^{-1}K_1^{-1}(r^{-2}+K_tC_{D_2}P_dQ)$$

Firstly, we consider the velocity control loop of the motor without the wearer’s torque, i.e. $\tau_h = 0$. We can derive the relationship between the motor velocity command $\dot{q}_{mc}$ and the real motor velocity $\dot{q}_m$ from Figure 3 as (22).

$$\dot{q}_m = r(I + B_1)^{-1}P_m^{-1}K_tC_{D_2}r^{-1}\dot{q}_{mc}$$

We assume that $C_{D_2}$ and $C_{D_1}$ stabilize the velocity loop, and $C_{D_2}$ is stable too. This implies that $(I + B_1)^{-1}$ is finite-gain $L_2$ stable.

Secondly, we derive the relationship between the joint velocity $q$ and the wearer’s torque $\tau_h$ from Figure 3. The derivation is straightforward, and the result is shown as (23).

$$q = (I + B_1 + B_2)^{-1}K_1\tau_h$$

(23) can be expressed as the block diagram in Figure 4-(a). Combining the inner-loop of Figure 4-(a) results in the block diagram in Figure 4-(b), where $A_1 = (I + B_1)^{-1}$ is finite-gain $L_2$ stable.

Suppose that the desired admittance function $P_d$ and the filter $Q$ are finite-gain $L_2$ stable, then $K_1$ in (20) is finite-gain $L_2$ stable. Therefore, stability of the system in Figure 4-(b) is determined by the feedback loop consisting of $A_1$ and $B_2$, which is denoted by $S$. Since $B_2$ can be decomposed as $B_2 = A_2B_3$, we can rearrange $S$ such that $S$ is equivalent to the system in Figure 4-(c) in terms of stability. Let $\Delta = P_a - \hat{P}_a$ be the model uncertainty, then $B_3 = Q\Delta$. We assume that $\Delta$ is finite-gain $L_2$ stable and $||Q\Delta|| \leq \gamma$ for some $\gamma > 0$. Based on the small gain theorem [27], the closed-loop system of Figure 4-(c) is finite-gain $L_2$ stable if (24) is satisfied.

$$||A_1A_2|| < 1/\gamma$$

We summarize the result as the following theorem, which gives the conditions that guarantee stability of the closed-loop system in the presence of model uncertainty $\Delta = P_a - \hat{P}_a$.

**Theorem:** If the exoskeleton system in Figure 3 satisfies the following assumptions,

1. $C_{D_2}$, $P_d$, $Q$ and the velocity control loop of the joint motor are finite-gain $L_2$ stable.
2. The model uncertainty $\Delta$ is finite-gain $L_2$ stable and $||Q\Delta|| < \gamma$ for some $0 < \gamma < \infty$.
3. $||A_1A_2|| < 1/\gamma$.

then the exoskeleton system is finite-gain $L_2$ stable.

B. SYSTEM ANALYSIS FOR PRECISE VELOCITY CONTROL

Due to advances in servo motor control, it is reasonable to assume that $C_{D_1}$ and $C_{D_2}$ achieve precise velocity tracking, i.e. $\dot{q}_m \approx \dot{q}_{mc}$. Hence, from (22), we have

$$r(I + B_1)^{-1}P_m^{-1}K_tC_{D_2}r^{-1} = rA_1P_m^{-1}K_tC_{D_2}r^{-1} \approx I$$

Then

$$A_1A_2 = A_1P_m^{-1}K_tC_{D_2}r^{-1}P_d \approx P_d$$

and (24) becomes (27).

$$||A_1A_2|| \approx ||P_d|| < \frac{1}{\gamma}$$

Note that (27) implies that the allowable gain of the admittance function is inversely proportional to the size of the (filtered) model uncertainty.

Next we rearrange the block diagram of the complete system in Figure 4-(b) as that in Figure 4-(d), where $K_2$ is given in (21). Since achieving high-performance trajectory tracking requires high-gain controllers, we can assume that $||C_{D_2}|| \gg 1$. Then $K_2$ is approximated by $Q$. Furthermore, if the uncertainty is negligible, i.e. $\Delta \approx 0$, then the relationship from $\tau_h$ to $q$ is approximated by (28).

$$q = P_d\dot{q}_{\tau_h}$$
It is clear to see the role of $Q$. $Q$ is used to shape the uncertainty $\Delta$ such that $\|Q\Delta\|$ is as small as possible, allowing a large gain for the admittance function (c.f. (27)). Since the uncertainty usually has a larger size in the high frequency range, $Q$ should be small in the high frequency range. On the other hand, if we want to achieve the desired admittance function $P_d$, we should have $Q \approx I$ according to (28). Since the bandwidth of the admittance function is low (relative to the uncertainty), it is desirable to have $Q \approx I$ in the low frequency range. Both conditions imply that $Q$ should be a low pass filter.

V. EXPERIMENT VERIFICATION

In this section, we conduct experiments to verify the accuracy and robustness of the torque estimation algorithm. Then we demonstrate that the gait of the wearer can be shaped by tuning the admittance parameters.

A. ACCURACY OF TORQUE ESTIMATION

The original design of the exoskeleton we used in this paper has no sensors to measure the ground reaction forces. Therefore, to identify the model parameters by the procedure illustrated in Section II, and to verify the torque estimation algorithm in (11), we dangle the exoskeleton on a rack to get rid of the effects of the ground reaction forces. See Figure 5 for the 3D model of the rack (left part) and the photo of the exoskeleton dangle on the rack (middle part). We are currently designing sensing modules that include force sensors under the soles and data transmission channels to the main controller such that the ground reaction forces can be directly measured and eliminated in the near future.

To verify accuracy of torque estimation, we need to obtain the ground truth, i.e. the actual human torque applied to HES. However, torque sensors do not help to get the ground truth because it is difficult to separate human torques and motor torques from the torque sensor measurements. Instead, we design the following experiments. Firstly, the exoskeleton is dangled on the rack with no wearer in it as shown in the middle part of Figure 5. Then, four force sensing resistors (FSRs) are installed in the opposite positions of the thigh and the shank of each leg as shown in the middle and right parts of Figure 1. Then we apply forces directly on these FSRs by hands to push the thigh or shank from both sides. The applied forces are measured by the FSRs and then transformed into the equivalent joint torques (denoted by measured torques in subsequent sections) by multiplying the corresponding Jacobian matrices. On the other hand, the equivalent torques are estimated through the torque estimation algorithm (11) and then fed back to generate the joint velocity commands through the pre-defined admittance (10) and then fed back to generate the joint velocity commands through the pre-defined admittance $P_d$. As a result, the thigh and shank of the exoskeleton comply with the applied forces and move back and forth.

To evaluate the torque estimation performance, we calculate the fitness of the estimated torques to the measured torques in terms of the normalized mean square error (NMSE) defined in (29):

$$\text{Fitness} = \left(1 - \frac{\|\hat{\tau}_{h,i} - \tau_{h,i}\|}{\|\tau_{h,i} - \text{avg}(\hat{\tau}_{h,i})\|}\right) \times 100\%, \quad i = H, K$$

(29)

where $\tau_{h,H}, \tau_{h,K} \in \mathbb{R}^N$ are vectors consisting of $N$ samples of measured torques from the hip and knee. Similarly, $\hat{\tau}_{h,H}, \hat{\tau}_{h,K} \in \mathbb{R}^N$ denote the vectors of $N$-sample estimated hit and knee joint torques, and $\text{avg}(\hat{\tau}_{h,i}) \in \mathbb{R}$ is the average of $\hat{\tau}_{h,i}$ over time. $1 \in \mathbb{R}^N$ is a vector whose elements are all 1’s. The norm used in (29) is the Euclidean norm for $N$-dimensional vectors. From (29) we see that if $\hat{\tau}_{h,i} = \tau_{h,i}$, then the fitness is 100%.

Figure 6 shows the torque estimation results for the hip joint and knee joint. The blue dashed line and red long dashed line denote the measured and estimated torques, respectively.
The left and right columns illustrate the results when the external forces are applied to the thigh and shank, respectively. The fitness of each case is listed below the corresponding subfigure in Figure 6 except the knee joint torque estimation when the external force is applied to the thigh. This is because the measured knee torque is negligible in such a condition, and therefore it makes no sense to evaluate fitness of torque estimation. The fitness of torque estimation in all cases exceeds 70%, which is sufficiently high to conduct admittance control and shape the wearer’s gait in a desired way. Experimental results of admittance control will be presented in Section V.C.

### B. ROBUSTNESS OF TORQUE ESTIMATION

When a wearer puts on the exoskeleton, the model parameter \( \theta \) in (10) deviates from its nominal value which is identified under the condition of no wearers (c.f. Section II.D). Since the proposed torque estimation algorithm is a model-based approach, it is desirable to evaluate robustness of torque estimation with respect to model uncertainties. Therefore, we repeat the experiment in the previous subsection; however, loads of weights 2kg and 4kg are attached to the end of the thigh and shank.

According to anthropometry [28], the mass of the thigh \((m_T)\) and shank \((m_S)\) for our test subject with height 173 cm and weight 67 kg are 6.7 kg and 3.116 kg, respectively. The lengths of thigh \((l_T)\) and shank \((l_S)\) are 0.424 m and 0.426 m, respectively. In addition, the distances from the hip and knee joints to the center of mass (COM) of the thigh \((l_{TC})\) and shank \((l_{SC})\) are both 0.184 m. The moment of inertia of the thigh \((I_T)\) and shank \((I_S)\) are 0.126 kgm\(^2\) and 0.047 kgm\(^2\), respectively. Based on these data, we can calculate the model parameters in Table 1 attributed to the wearer. For example, the parameters of the gravity term \( G(q) \) (c.f. (2) and (10)) have values \( P_2 = ms_{l_{SC}} g = 0.573 g \text{Nm} \) and \( P_1 = (m_T l_{TC} + m_S l_T) g = 2.554 \text{Nm} \), where \( g \) is the gravity acceleration (9.8 \text{m/s}^2); the parameters of the inertial matrix \( M(q) \) have values \( P_3 = I_S + ms_{l_{SC}} = 0.152 \text{Nms}^2 \) and \( P_8 = I_T + m_T l_{TC}^2 + ms_{l_{TC}}^2 = 0.915 \text{Nms}^2 \).

If we attach a load of weight 2 kg to the end of the thigh and a load of weight 4 kg to the end of the shank, then the corresponding parameters are \( m_T = 2 \text{kg}, m_S = 4 \text{kg}, l_{TC} \approx l_T = 0.424 \text{m}, l_{SC} \approx l_S = 0.426 \text{m}, I_S = ms_{l_{SC}}^2 = 0.726 \text{kgm}^2, \) and \( I_T = m_T l_{TC}^2 = 0.36 \text{kgm}^2 \). The associated model parameters have values \( P_2 = ms_{l_{SC}} g = 1.704 \text{gNm}, P_7 = (m_T l_{TC} + m_S l_T) g = 2.552 \text{Nm}, P_3 = I_S + ms_{l_{SC}} = 2.43 \text{Nms}^2 \) and \( P_8 = I_T + m_T l_{TC}^2 + ms_{l_{TC}}^2 = 1.439 \text{Nms}^2 \), which are greater than or close to the values associated with the test subject. We also double the weight of the load on the thigh to take into account a wider range of parameter variations due to different wearers. Hence the experimental setup can simulate severe conditions of parameter variations and is suitable for testing robustness of the proposed algorithm.

The experimental results are shown in Figure 7 to Figure 10, where the blue dashed lines and red long dashed lines in Figure 7 and Figure 9 represent the measured torque and estimated torque, respectively. In Figure 7, the external forces are applied to the thigh and shank. The external force \( (U_{d,H} = I_{d,K}, B_{d,H} = B_{d,K}) = (10000, 10000) \), Red long dashed line represents the estimated torque and blue dashed line represents the measured torque.
force is applied to the thigh. As explained in the previous subsection, the equivalent knee joint torque induced by this external force is negligible and therefore is not shown here. We can observe that the fitness of the torque estimation is similar to that in the no-load condition.

In Figure 9 the external force is applied to the shank. Both the estimation results for the hip and knee joints are presented. Although the fitness of the knee joint torque estimation decreases, we can see that the waveforms of the measured and estimated torques are in phase and very similar to each other. They are different only in the peak values. If we check the torque estimation errors shown in Figure 8 and 10 for the results of Figure 7 and Figure 9, respectively, we see that the maximum errors are still in a reasonable range, which allows successful implementation of the admittance control law.

Since the joint velocity and acceleration are low in normal walking conditions, the most influential model parameters to $\hat{\tau}_h$ are the gravitational parameters, $P_2$ and $P_7$. However, we notice from (2) that the effects of these parameters vanish when the leg is in an upright configuration, i.e. $q_H = q_K = 0^\circ$, and become most significant when $q_H = 90^\circ$ and $q_K = 0^\circ$. The configuration-dependent nature of the model uncertainty results in fluctuation of fitness in each experiment, because the external forces are applied by hands and it is difficult to maintain the same swing angles of the leg during experiments. However, as long as the estimated torque is in phase with the measured one, we are able to implement effective admittance control and provide assistive torques to the wearer at the right timing.

C. ADMITTANCE CONTROL

As we mentioned earlier, the admittance of each joint can be adjusted by the control law. In this subsection, we conduct experiments to show the effects of different admittance on the wearer’s gait. To avoid the interference of ground reaction forces, a test subject wearing the exoskeleton is supported by the rack without his feet on the ground (see the right part of Figure 5). The subject is asked to freely swing his legs in his most comfortable way while the control system in Figure 3 is activated. The desired admittance $P_d$ has the form of (13) whose parameters $J_d = \text{diag}(J_{d,H}, J_{d,K})$ and $B_d = \text{diag}(B_{d,H}, B_{d,K})$ are set as different values. The human torque to HES is estimated by (11), which has verified robustness and accuracy as we demonstrated in the previous subsection.
Two sets of admittance parameters will be fine-tuned in a systematic way such that the effects of GRFs on the estimated torques as indicated by reaction forces (GRFs) directly. Therefore, we can eliminate the parameters for smaller inertia and damping coefficient (i.e. lighter load) result in smaller joint torques, faster joint velocities and shorter gait cycle time. This experiment demonstrates that the proposed method can shape the wearer’s gait by properly tuning the parameters of the desired admittance.

VI. CONCLUSION AND FUTURE WORK

In this paper, we designed and implemented a control scheme which shapes the admittance of the human-exoskeleton system (HES) based on estimated joint torques from a healthy wearer. The degree of compliance was online adjustable by tuning the parameters of the desired admittance function. Robust stability of the closed-loop system, including HES, joint torque estimator, and the admittance controller, was proved rigorously, and experiments were carried out to verify accuracy and robustness of the proposed torque estimation algorithm. Besides, the effects of different admittance parameters on shaping the wearer’s gait were also demonstrated by experiments.

According to the results of this paper, we conclude that the advantages of the proposed method are twofold:

1) It estimates the wearer’s torque as the walking intention. The torque estimation is accomplished without using any biological sensors, and the results are sufficiently accurate and robust. In addition, the estimated torque provides abundant information about the wearer. For example, we can use the wearer’s torque and the joint angular velocity to estimate the power consumption of the wearer during walking.

2) The proposed control scheme is trajectory free. Namely, the motion of HES is completely determined by the estimated wearer’s torque. No pre-planned trajectories for the joints are required. Therefore, the wearer can move freely, and the exoskeleton complies with the wearer. The degree of compliance is online adjustable by tuning a few parameters. Then the gait of the wearer changes accordingly.

In the future, the sensing module we are currently working on will be installed on the exoskeleton to measure the ground reaction forces (GRFs) directly. Therefore, we can eliminate the effects of GRFs on the estimated torques as indicated in (11). Then walking tests will be conducted and the admittance parameters will be fine-tuned in a systematic way such that the wearer can walk naturally and safely with minimal power consumption.

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