

Active Compliance Control Reduces Upper Body Effort in Exoskeleton-Supported Walking

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Abstract—This paper presents a locomotion controller for lower limb exoskeletons so as to enable the combined robot&user system to exhibit compliant walking characteristics when interacting with the environment. This is of critical importance to reduce the excessive ground reaction forces during the walking task execution with the aim of improved environmental interaction capabilities. In robot-aided walking support for paraplegics, the user has to actively use his/her upper limbs via crutches to ensure overall balance. By virtue of this requisite, several issues may particularly arise during touch-down instants, e.g., upper body orientation fluctuates, shoulder joints are subject to excessive loading, and arms may need to exert extra forces to counterbalance these effects. In order to reduce the upper body effort via compliant locomotion, the controller is designed to manage the force/position trade-off by using an admittance controller in each joint. For proof of concept, a series of exoskeleton-aided walking experiments were conducted with the participation of 9 healthy volunteers, 4 of which additionally walked on an irregular surface for further performance evaluation. The results suggest that the proposed locomotion controller is advantageous over conventional high-gain position tracking in decreasing undesired oscillatory torso motion and total arm force, adequately reducing the required upper body effort.

Index Terms—lower body exoskeleton, locomotion control, compliance control.

I. INTRODUCTION

Robotic lower limb exoskeletons are designed to improve the health and life quality for those who suffer impaired lower limb functionality. The advantages of these systems over wheeled assistive devices may be assessed by referencing [1], [2]. Assistive devices with wheels may not provide overall ambulatory support on uneven surfaces, e.g., narrow pathways, stairs. In response to this matter, there is a long-lasting research effort in developing exoskeleton systems [3], [4], which led to several commercially available systems that were proved to be useful in providing robot-aided walking support [5]–[7].

Considering the robot-aided walking task for paraplegia, the above-mentioned systems employ high gain controllers that

achieve precise joint tracking. While useful in its functionality domain, this approach maximizes joint stiffness, and thus, the combined human&robot system experiences excessive ground reaction force peaks that lead to undesired upper body oscillations and increased upper body effort [8]–[11]. On this matter, we argue that the precise joint tracking should not be the most important criterion for the task; compliant contact with the ground should also be addressed in the locomotion control logic [12].

The importance of compliance for exoskeleton systems was underlined in [13], where the researchers proposed the use of pneumatic muscles. Using similar actuators, Ugurlu et al. developed a lower-body exoskeleton with self-balancing capability [14]. Cestari et al. introduced a compliant actuator unit for lower body exoskeletons [15]. Lv et al. defended that the use of backdrivability leads to enhanced performance in exoskeletons [16]. Veneman et al. introduced LOPES, an impedance-controlled exoskeleton [17]. Neuhaus et al. developed an exoskeleton system that can be compliant via a position/force control strategy [12]; however, details regarding the controller were not disclosed.

Sharing the common ground with [12]–[17] in terms of control philosophy, we propose an actively compliant locomotion controller for our electrically-actuated exoskeleton prototype, called TTI-Exo. Since the system is powered via conventional electrical actuators to avoid complexity and contain costs, torque control is not viable. Therefore, an admittance control method is implemented to introduce active compliance on top of the joint position tracking control. The controller prioritizes joint tracking to achieve fine motion control performance, e.g., during leg swinging. When the robot is in contact with the environment, the admittance controller generates the required joint displacement that corresponds to contact force error and then it is added as the secondary feedback to handle force constraints. This results in relatively more comfortable and compliant walking support for the user since the system is able to suppress ground reaction force peaks and perturbations.

In the lower body exoskeleton control literature, the use of compliance was exploited for collaborative human-robot tasks and balancing. However, its potential benefits in reducing the upper body effort for the walking support task have not been demonstrated [18]–[20]. Therefore, the current work contributes toward this direction. We proposed the use of a compliant locomotion controller and tested its effectiveness via real hardware experiments to demonstrate enhanced walking support concerning the upper body effort reduction target for the first time to the authors' knowledge. The current paper

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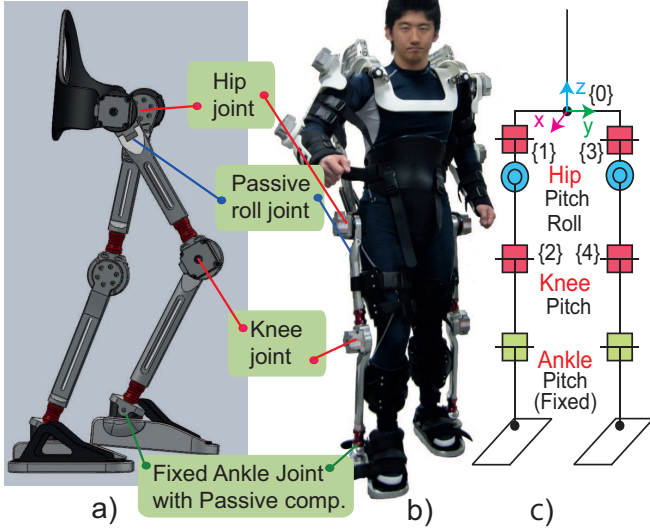


Fig. 1. a) CAD model for the lower body. b) TTI-Exo with an able-bodied user. c) Joint configurations for the lower body; red: Active DoF, blue: passive DoF, green: fixed DoF with sufficiently high stiffness for physical compliance.

is a significantly expanded and improved version of [21] which presented the preliminary results on limited subjects. Improvements on top of [21] are as follows: i) enriched mathematical analysis of the controller with stability proof, ii) newly-conducted experiments on 9 subjects, iii) statistical evaluations of the experiment results, iv) further elaborations of the controller including the friction observer.

This article is organized as follows. The hardware structure of TTI-Exo is briefly presented in section II. The methods used in this work are explained in section III. Experimental results are presented in section IV and discussed in section V. Finally, the paper is concluded in section VI.

II. LOWER EXTREMITY EXOSKELETON: TTI-Exo

Intending to provide robot-aided rehabilitation in a wide range, we developed a whole-body exoskeleton robot, TTI-Exo. The upper body of TTI-Exo was proven to be useful for robotic rehabilitation and power augmentation tasks [22]. The lower body of TTI-Exo is utilized for paraplegia walking support. See Fig. 1 to view the actual system, its CAD model and joint configuration for the lower body.

In order to constitute a simple, durable and applicable system and as well as to contain the overall weight and cost, we followed the conventional design approach [5]–[7]. Each leg is driven with 2 active DoF (degrees of freedom) at the hip and knee joints (flexion/extension), powered via electrical actuators. Each actuator unit is equipped with encoders to measure the joint angle. A passive 1 DoF is added to hip joints through adduction/abduction to allow lateral movements. Due to a reduced number of active joints, the robot cannot maintain the inherent balance; therefore, the user has to engage her/his upper limbs via crutches to ensure quasi-static balance.

Ankle joints are fixed with a flexible but sufficiently stiff material to account for ground impacts through physical com-

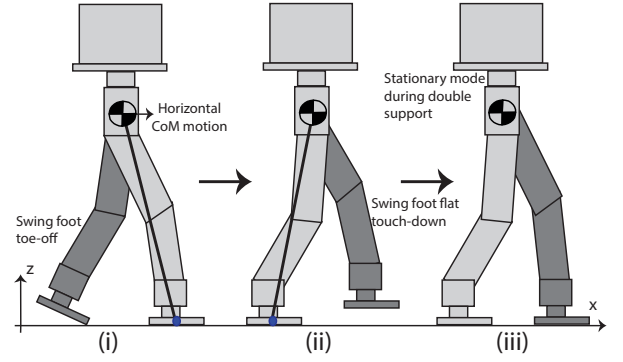


Fig. 2. i) A single support phase starts with a toe-off instant. The support foot center is the pivot. ii) A double support phase is considered to be a delay period; there is no forward motion. This allows users to keep up with the robot motion as she/he needs to engage the upper limbs. iii) Motion generation parameters are adjusted in a way so as to ensure flat foot touch-down to reduce impacts.

pliance. Link lengths are made adjustable to align human and robot joints as much as possible to minimize the differences between human and robot kinematic chains. In particular, the average rotation axis principle is considered for the knee joints [23]. C-shaped braces and straps are utilized to attach the user to TTI-Exo. Potential parasitic force issues due to kinematic chain alignments will be examined in a future work [24].

4 low-cost force sensing units (Interlink FSR 408) were placed under each foot sole to directly measure the vertical contact force (f_z) with the help of a meticulous calibration and modeling procedures [25], [26]. Furthermore, the horizontal center of pressure (p_x) and the horizontal contact force (f_x) can be estimated using force distribution and pendulum model, respectively [27].

$$f_z = \sum_{i=1}^n F_i. \quad (1)$$

$$p_x = \frac{\sum_{i=1}^n (F_i l_{ix})}{f_z}. \quad (2)$$

$$f_x = \frac{x_c - p_x}{z_c} f_z. \quad (3)$$

In (1)–(3), F_i stands for i^{th} force sensor reading. Horizontal and vertical forces are denoted via f_x , and f_z . The distance between the i^{th} force sensor the respective foot sole center is l_{ix} . CoM (Center of Mass) position, (x_c, z_c) can be computed using forward kinematics.

III. METHODS

A. Trajectory Generation

To ensure the user comfort and enhance usability, the trajectory design should consider the following properties: i) the swing foot trajectory, (x_f, z_f) , is carefully synthesized to avoid oblique feet impacts with the ground, ii) the torso orientation, θ , should be kept constant to prevent upper body oscillations, iii) the dominant forward motion parameter x_c is differently designed for single and double support phases, iv)

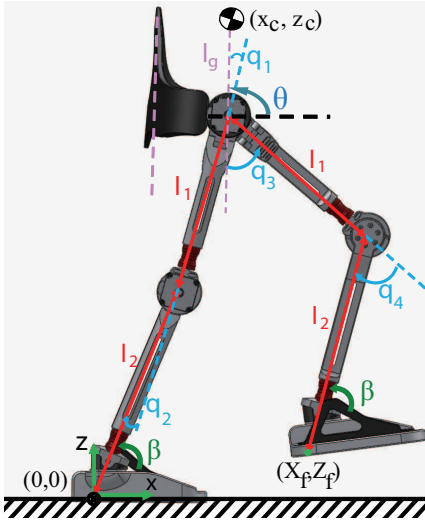


Fig. 3. Kinematic parameters are indicated in the CAD model. The distance between the CoM and pelvis, l_g , is considered to be constant. For each user, link lengths (l_1 , l_2 , l_g) can be adjusted prior to use. In order to extend the stride distance, the ankle angle, β , is fixed to 70° . Active joints are denoted with q_i ($i = 1, 2, 3, 4$) while θ stands for the pitch orientation of the torso.

all trajectories should be smooth and continuous; e.g., there should be no discontinuity in position, velocity, and acceleration profiles to preclude torque jumps and jerky motions.

Considering the fact that the lower body of TTI-Exo has 4 active DoF, we determine that 4 trajectory inputs, (x_f, z_f, x_c, θ) are of critical importance. Boundary conditions are carefully assigned by considering the mean target velocity, foot clearance, and desired stride length. In particular, swing foot trajectories are synthesized to ensure zero velocity and acceleration values at the touch-down moment. Since the robot is underactuated per se, other parameters such as CoM height (z_c) and swing foot orientation could not be imposed as inputs. Yet, trajectory parameters are chosen meticulously with the help of a kinematic simulator to keep CoM height vary within a tight range, and also to make sure that the swing foot touches down with no oblique landing then clears from the ground with toe-off; see Fig. 2.

Our locomotion controller needs to process expected contact forces, i.e., CFR (Contact Force Reference), which are computed in line with the trajectory references. During a single support phase, the vertical CFR is expected to be the total weight as the CoM height varies in a small range and therefore the vertical CoM acceleration is negligible.

$$f_{xref1} = m\ddot{x}_{cref}, \quad (4)$$

$$f_{zref1} = mg, \quad (5)$$

$$f_{xref2} = 0, \quad f_{zref2} = 0 \quad (6)$$

In (4)-(5), m is the total mass (robot and user), g is the gravitational acceleration, f_{xrefi} and f_{zrefi} are the horizontal and vertical CFR for the i^{th} foot. Underscripts 1 and 2 refers to left and right feet; in this specific case, the left foot is the stance foot. As the horizontal CoM trajectory is generated via polynomial interpolation, we can easily obtain

$\ddot{x}_{cref}$. Furthermore, CFR values for a swing foot is set to zero.

During double support phases, the forward acceleration is close to zero due to the nature of static walking; thus, the horizontal CFR is set to zero. In this case, the total weight is desired to be distributed equally to both legs.

$$f_{xref1} = 0, \quad f_{xref2} = 0, \quad (7)$$

$$f_{zref1} = 0.5mg, \quad f_{zref2} = 0.5mg. \quad (8)$$

Having synthesized the 4 trajectory inputs (x_f, z_f, x_c, θ) , joint angle references can be computed using the analytical inverse kinematics algorithms provided in [21]. Refer to Fig. 3 for the kinematic parameters.

B. Compliant Locomotion Control

The block diagram of the proposed locomotion controller is displayed in Fig. 4, with a detailed look at the Active Compliance Controller (ACC). In this figure, τ_{fr} , τ_{cmd} , τ_{cg} , τ_{sv} respectively denote friction compensation torque, command torque, Coriolis and centrifugal and gravity compensation torque, and servo (PD) output torque. Joint angles and contact forces are indicated with q and f . Underscripts *err* and *ref* indicate error and reference values. Measured quantities have no underscript. The secondary feedback from the ACC is q_v .

As explained in section III, the trajectory generator block produces reference joint angles and CFR values. Reference joint angles are inserted to servo controllers to achieve joint tracking. The ACC is the main block to provide compliant locomotion characteristics. Moreover, Coriolis and centrifugal, friction and gravity compensation blocks added to the controller for the enhanced mechanical performance.

1) *Active Compliance Controller*: Active Compliance Control block is mainly responsible for the compliant environmental interaction; see Fig. 4(b). In principle, it computes the joint displacement value q_v which corresponds to contact force error. In other words, the joint angle reference is updated $q_{ref} =: q_{ref} - q_v$, such that the system complies with the contact force constraints to manage force/position control trade-off. To this end, the ACC computes the contact force errors depending on the phase state (double/single).

$$f_{zerr\mu} = f_{zref\mu} - f_{z\mu}; \quad f_{xerr\mu} = f_{xref\mu} - f_{x\mu}. \quad (9)$$

The additional μ letter in (9) tags the legs; right: $\mu = 2$, left: $\mu = 1$. Using contact force errors, the secondary feedback q_v can be determined as below.

$$q_{v\mu} = \begin{bmatrix} \frac{1}{b_{h\mu}s + k_{h\mu}} & 0 \\ 0 & \frac{1}{b_{k\mu}s + k_{k\mu}} \end{bmatrix} \left(J_{c\mu}^T \begin{bmatrix} F_{xerr\mu} \\ F_{zerr\mu} \end{bmatrix} \right) \quad (10)$$

In (10), $(b_{h\mu}, k_{h\mu})$ and $(b_{k\mu}, k_{k\mu})$ parameters respectively denote admittance coefficients that are associated with the hip and knee joints for the μ^{th} leg. $J_{c\mu}$ is the Jacobian that maps the CoM velocity with respect to μ^{th} foot center to joint velocities. Laplace variable is s .

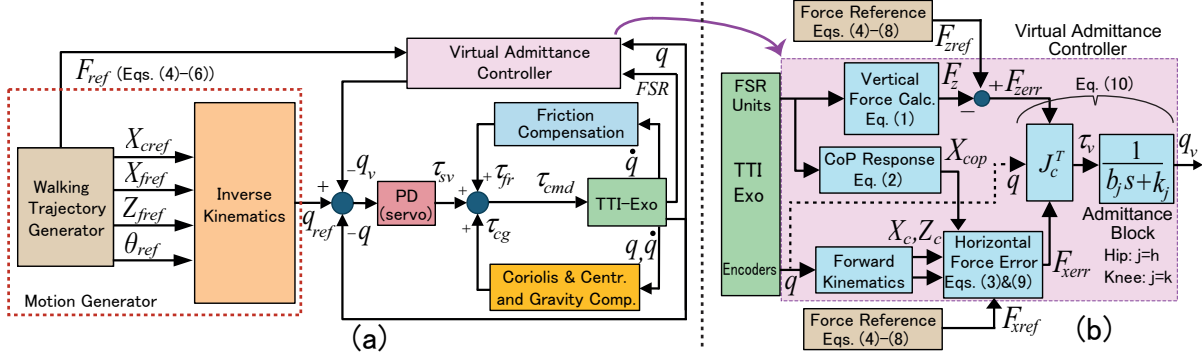


Fig. 4. a) The proposed locomotion controller. b) The internal structure of the Active Compliance Controller (ACC) for a single leg.

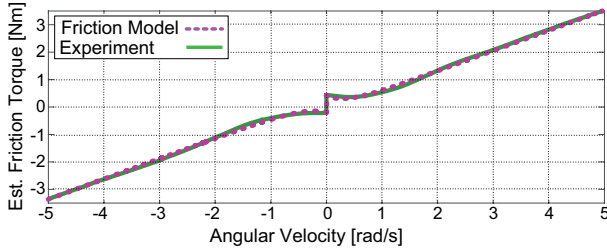


Fig. 5. Friction identification experiments for the left knee joint. Similar results were obtained for other joints and thus not plotted.

The main advantage of using admittance control lies in the fact that force constraints are handled only at required instants, e.g., touch down. If there is no force error to deal with, for instance, when the leg is swinging, the contact force error is zero. Therefore, q_v converges to zero, and the controller automatically prioritizes the joint tracking. If there is a contact force error due to an impact or perturbation, the joint reference is updated in a way so as to handle the force constraint. This trade-off is managed via admittance coefficients; the smaller the admittance coefficients, the more sensitive the ACC to force errors, and vice versa.

2) Compensation Schemes: Friction, Coriolis&Centrifugal and Gravity Torques: In order to ensure the position control for the enlarged position control bandwidth, we added compensation blocks. Coriolis & centrifugal and gravity torques are calculated using a floating base inverse dynamics algorithm, and added as feedforward torques; see τ_{cg} in Fig. 4. This is applied only for the robot; the dynamic and static loading from the human is considered as the external disturbance. Note that the robot moves in a quasi-static fashion; thus, inertia terms are negligible.

The friction compensation method was adopted from our work on the upper body of TTI-Exo; see [22]. The experimentally obtained friction model and the fitted model are depicted in Fig. 5. The friction model was constituted by considering the Coulomb friction, viscous friction and the Stribeck effect as follows [28].

$$\tau_{fr} = \begin{cases} 0.42 - 0.58 \left(1 - e^{-2.88\dot{\theta}}\right) + 0.74\dot{\theta} & \text{if } \dot{\theta} > \epsilon \\ -0.19 + 0.53 \left(1 - e^{-2.26\dot{\theta}}\right) + 0.75\dot{\theta} & \text{if } \dot{\theta} < -\epsilon \end{cases} \quad (11)$$

In (11), an infinitesimal dead region was deliberately constructed via using an empirical ϵ value to prevent overcompensation in practice.

3) Stability: The dynamic model of the robot can be described in (12), where $M(q)$, $C(q, \dot{q})$, $F_r(\dot{q})$, $G(q)$ stand for inertia, coriolis and centrifugal, joint friction and gravity terms, respectively. The disturbance torque of τ_d arises due to physical interaction with the environment.

$$M(q)\ddot{q} + C(q, \dot{q})\dot{q} + F_r(\dot{q}) + G(q) = \tau_{sv} + \tau_{fr} + \tau_{cg} - \tau_d \quad (12)$$

With the addition of feedback linearization torques that account for joint friction, Coriolis and centrifugal and gravity terms, robot dynamics are reduced to the following form, in which the modeling uncertainties in (12) assumed to be negligible.

$$M(q)\ddot{q} = \tau_{sv} - \tau_d \quad (13)$$

As displayed in Fig. 4, the actuator torque command τ_{sv} was designed by considering two loops: i) position control loop with a PD controller, ii) force control loop via an admittance controller.

$$\tau_{sv} = K_p(q_{ref} - q_v - q) + K_d(\dot{q}_{ref} - \dot{q}_v - \dot{q}) \quad (14)$$

In (14), K_p and K_d are diagonal matrices that store PD controller gains. Plugging (14) into (13) results as follows:

$$M(q)\ddot{q} + K_d\dot{q} + K_pq = K_d(\dot{q}_{ref} - \dot{q}_v) + K_p(q_{ref} - q_v) - \tau_d. \quad (15)$$

At this stage, we assume that the diagonal elements of $M(q)$ are more dominant because of the high gear ratio; therefore, its non-diagonal elements are neglected [29]. With

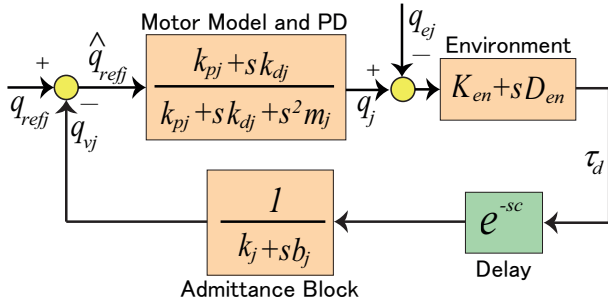


Fig. 6. An abstracted block diagram to provide a simplified view for the compliance control scheme depicted in Fig. 4. Compensated robot dynamics and position feedback (PD) controller are combined within a single block whose reference input can be considered *updated* via the admittance controller; $\hat{q}_{refj} := q_{refj} - q_j$. Measurement delay is represented via e^{-sc} , where c is the delay constant.

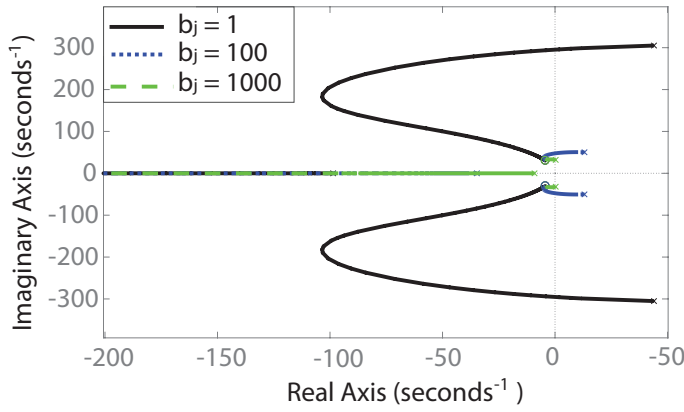


Fig. 7. Root locus plot concerning the changes in the admittance gain k_j .

this assumption in mind, one can reformulate the proposed controller scheme in joint space. Fig. 6 displays an abstracted view of the controller in that sense where compensated robot dynamics and PD control loop are combined via block reduction. Consequently, eq. (15) takes the following form for the j^{th} joint,

$$m_j \ddot{q}_j + k_{dj} \dot{q}_j + k_{pj} q_j = k_{dj} (\dot{q}_{refj} - \dot{q}_{vj}) + k_{pj} (q_{refj} - q_{vj}), \quad (16)$$

$$\tau_{dj} = b_j \dot{q}_{vj} + k_j q_{vj}, \quad (17)$$

in which k_j and b_j are admittance coefficients, m_j is the corresponding diagonal inertia element. Considering Fig. 6, the closed loop transfer function can be expressed as follows,

$$T(s) = \frac{L(s)}{1 + L(s)}, \quad (18)$$

where $L(s)$ is the open-loop transfer function defined between $\hat{q}_{refj}$ and q_{vj} :

$$L(s) = \frac{(k_{dj}s + k_{pj})(D_{en}s + K_{en})}{(m_js^2 + K_d s + K_p)(b_js + k_j)}. \quad (19)$$

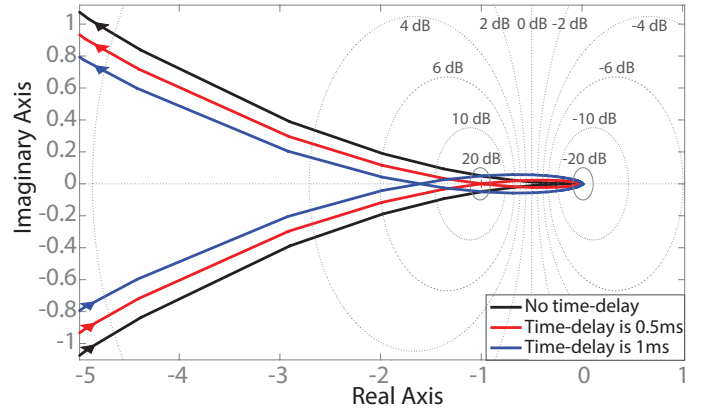


Fig. 8. Nyquist graph for different time delays.

Fig. 7 represents the root-locus of the admittance control system with respect to k_j when b_j has three different values ($b_j = 1$, $b_j = 10$ and $b_j = 1000$) and $e^{-sc} = 1$ (i.e., no time-delay). The stability of the admittance control system can be improved by increasing the stiffness of the system. To stably perform compliant motion when the stiffness of the impedance controller is low, higher damping values should be used. Fig. 8 represents the Nyquist response when the admittance control system possesses different time-delays. As the time-delay increases, the stability of the impedance control significantly deteriorates.

C. Experiment Protocol

1) *Participants*: Nine healthy volunteers (7 males, 2 females) participated in the study. Their average age was 28.6 years (SD = 12.3 years), height 167 cm (SD = 7.8 cm) and body mass 59.5 kg (SD = 10.1 kg). Participants gave written informed consent before starting the experiment. The study was approved by Toyota Technological Institute and was conducted in accordance with the Declaration of Helsinki.

2) *Procedures*: Walking experiments were carried out using a treadmill (Ohtake-Root Kogyo Co. Ltd., ORK-200R-1500VA, Japan). The participants were instructed to keep their legs passive. Before each experiment, the experimental setup was introduced to the participant in question and she/he conducted simple trials to familiarize her/himself to the device.

A single walking experiment is as follows: i) The participant receives walking support when the exoskeleton is position-controlled via the conventional method [6] (1 minute), ii) a rest period of 15 minutes, iii) the participant receives walking support when the exoskeleton is controlled via the proposed admittance control method (1 minute). For each participant, a unique set of walking parameters was strictly kept the same to ensure no change in the trajectory.

In all experiments, single and double support periods were assigned as 1.2 s and 0.14 s, respectively. Since the participants have different lower limb lengths, the stride length was experimentally adjusted in accordance with the participants' choice and varied between 22 to 26 cm. Accordingly, the forward velocity was between 0.6 to 0.71 km/h.

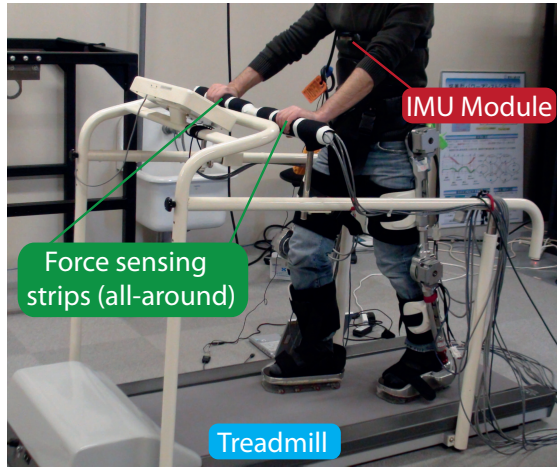


Fig. 9. Experiment setup for the walking experiments. An IMU sensor was firmly placed on the participant's torso to measure pitch rate and horizontal acceleration. Force sensing strips were attached to the treadmill handhold to collect the forces exerted via the participant's arms.

The admittance controller parameters were empirically tuned for each subject by considering the stability and negotiating the trade-off between joint tracking and upper body effort trade-off. For Subject #1, admittance controller gains, k_j and b_j , were set to 452 rad/Nm and 290 rad/Nms for the hip joints and 245 rad/Nm and 148 rad/Nms for the knee joints. For each subject, these gains were tuned separately but found to be within the vicinity of these values.

Two types of experiments were conducted: a) Flat surface walking with 9 subjects. b) Irregular surface walking with 4 subjects. In the second type, wooden blocks with a thickness of 3 cm were used in a way such that the human-exoskeleton combined system had to step on.

3) *Data Collection*: For a quantitative evaluation, we collected pitch rate and horizontal acceleration data via an IMU (LORD MicroStrain, 3DM-GX3-25, North Carolina, US) that was seamlessly attached to the participants' upper torso. The treadmill handhold was equipped with force sensing strips to measure the force that was exerted to the handle via the participant's arms. Fig. 9 displays the experiment setup. Furthermore, the force data from the feet soles and robot joint data from the encoders were readily available to obtain ground reaction force and center of pressure data. The acquired data were low pass filtered (2nd order Butterworth filter with a 20 Hz cut-off frequency) and trends were removed.

4) *Evaluation Criteria*: To quantitatively evaluate the upper torso effort during walking, the following parameters are considered: i) x-axis torso acceleration data, $\ddot{X}_t$, ii) torso pitch rate, $\dot{\Phi}_y$, and iii) the total force exerted by the arms, F_{ta} . Since the overall balance is maintained using the arms, it is expected to observe F_{ta} variation with a comparatively lower magnitude in the case of reduced upper body effort [30], [31]. Likewise, the torso motion should vary less and be well-confined, therefore, reductions in $\ddot{X}_t$ and $\dot{\Phi}_y$ are expected [8]–[11]; see Fig. 10.

5) *Statistical Analyses*: Statistical analyses were performed using SPSS 21 (SPSS Inc., Chicago, USA). For each subject,

TABLE I
EXPERIMENT #1 RESULTS, PERCENTAGE DECREASES IN RMS

	$\ddot{X}_t$	$\dot{\Phi}_y$	F_{ta}	q_{2e}
Subject #1	29.4 %	34.9 %	32.6 %	-43.3 %
Subject #2	33 %	35.4 %	21.2 %	-38.7 %
Subject #3	33.2 %	15.1 %	37.5 %	-48.4 %
Subject #4	-13.8 %	-3.2 %	6.8 %	-33.9 %
Subject #5	37.7 %	50.0 %	45.2 %	-43.2 %
Subject #6	18.6 %	43.2 %	10.5 %	-21.2 %
Subject #7	32.8 %	43.6 %	32.6 %	-39.4 %
Subject #8	53.5 %	34.8 %	11.7 %	-39.9 %
Subject #9	23.2 %	28.5 %	15.5 %	-26.6 %

RMS (Root Mean Square) values of parameters $\ddot{X}_t$, $\dot{\Phi}_y$, and F_{ta} were calculated for the conventional and proposed controllers. The values of the individual subjects were then used for statistical analysis. The effects of the controller on all three parameters were investigated using a paired sample T-test. The level of statistical significance was set to 0.05. We carried out a pilot study ($n = 4$), calculated power-related parameters ($\alpha = .05$, $\text{power} = .80$) and based on that a sample of 9 subjects or more turned out to be sufficient.

IV. RESULTS

A. Experiment #1: Flat Surface Walking

The experimental results concerning Experiment #1 are shown in Table I where we reported percentage reductions in RMS values of $\ddot{X}_t$, $\dot{\Phi}_y$, and F_{ta} . In case of an increase of the parameter value, the reported value is negative. In addition, the increments in RMS value of joint tracking error for the right knee (q_{2e}) are provided as an example. Fig. 11 displays these variations for a single subject for representative purposes where solid green and dotted magenta lines display $\ddot{X}_t$, $\dot{\Phi}_y$, F_{ta} , and q_{2e} variations for the conventional and proposed methods, respectively.

Reductions in RMS values of $\ddot{X}_t$, $\dot{\Phi}_y$, and F_{ta} were observed for 8 subjects when the proposed controller was used. These reductions ranged from 18.6 % to 53.5 % for $\ddot{X}_t$, 15.1 % to 50 % for $\dot{\Phi}_y$, and 11.7 % to 45.2 % for F_{ta} . One subject (Subject #4) did differ from this group. The RMS values of this subject for $\ddot{X}_t$ and $\dot{\Phi}_y$ increased for 13.8 % and 3.2 % respectively. The total force exerted by the arms showed a slight decrease for about 6.8 %. Overall, all three parameters statistically significantly reduced when the proposed controller was used; $t(8) = 4.12$, $p = .003$ for $\ddot{X}_t$, $t(8) = 4.01$, $p = .004$ for $\dot{\Phi}_y$, and $t(8) = 5.12$, $p < .001$ for F_{ta} .

On the contrary to these reductions, joint tracking error increased. In particular, at the moments of touch-down, tracking errors up to ± 2.5 degrees were observed. However, we argue that this outcome did not affect the walking performance significantly as the touch-down and lift-off instants were synchronized when using both conventional and proposed methods; see Fig. 12. For visual inspection, please refer to the multimedia attachment to observe that both walking experiments were synchronized with no apparent phase shift.

In order to provide an alternative view, we also computed the mean $\pm$ SD variations of 8 consecutive walking steps; see Fig. 13. In this figure, green and magenta regions indicate

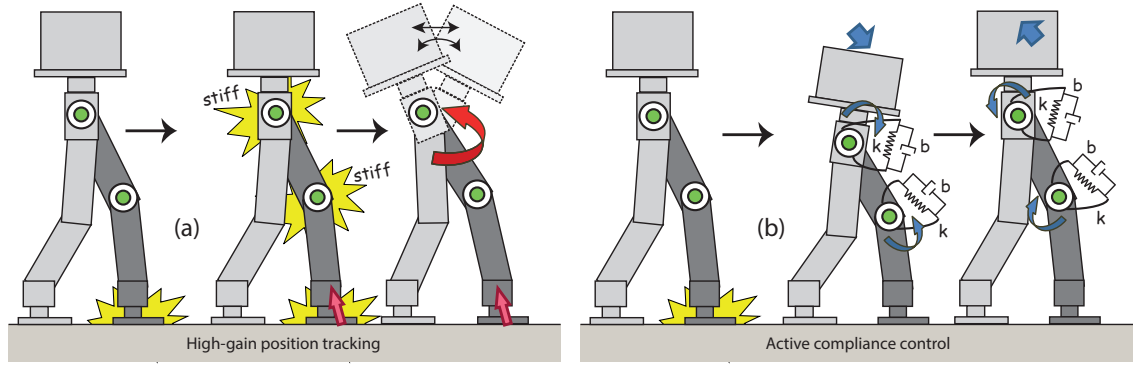


Fig. 10. a) When the robot joints are controlled via high-gain stiff controllers, the exoskeleton is not able to handle contact forces, e.g., ground impacts, in a compliant manner. This strategy has severe effects on the overall postural stability, leading to unavoidable torso oscillations and unexpected jerky movements [8]–[11]. Therefore, the user may have to put extra effort to maintain the balance. b) With the active compliance control strategy, the robot is able to cope with ground contacts and impacts compliantly. As these environmental perturbations are dissipated, the torso oscillations and unexpected forward-backward movements are well contained.

TABLE II
EXPERIMENT #2 RESULTS, PERCENTAGE DECREASES IN RMS

	$\ddot{X}_t$	Φ_y	F_{ta}	q_{2e}
Subject #1	19.8 %	27.8 %	36.0 %	-26.1 %
Subject #2	29.0 %	28.0 %	27.6 %	-39.5 %
Subject #3	26.4 %	28.0 %	33.8 %	-25.0 %
Subject #4	31.1 %	28.5 %	21.3 %	-42.1 %

the mean $\pm$ SD variations for the conventional and proposed methods, respectively. As may be observed, the mean variations of $\ddot{X}_t$ and Φ_y varied similarly in both cases; yet, standard variations covered a larger area in the case of the conventional method. This is an expected result since $\ddot{X}_t$ and Φ_y varied around zero in both cases; however, the amplitudes were larger in the case of the conventional method. When it comes to F_{ta} , the proposed method provided more contained mean variation and standard deviation. The joint tracking, on the other hand, indicated a relatively larger variation as mentioned previously.

B. Experiment #2: Irregular Surface Walking

Due to its challenging nature, Experiment #2 was conducted with the participation of only 4 able-bodied volunteers. The results are tabulated in Table II where we report percentage reductions (increases in the case of negative values) in RMS of $\ddot{X}_t$, Φ_y , F_{ta} , and q_{2e} . Fig. 14 displays these variations for a single subject for representative purposes where solid green and dotted magenta lines display Φ_y and F_{ta} variations for the conventional and proposed methods, respectively.

Examining Table II, percentage reductions in RMS values of $\ddot{X}_t$, Φ_y , and F_{ta} were observed for all 4 subjects when the proposed controller was used. These decreases range from 19.8 % to 31.1 % for $\ddot{X}_t$, 27.8 % to 28.5 % for Φ_y , and 21.3 % to 36.0 % for F_{ta} . Similarly to Experiment #1, joint tracking error increased.

As an alternative view, Fig. 15 displays the mean $\pm$ SD variation regions for 8 consecutive steps. Green and magenta regions indicate the corresponding variations for the conventional and proposed methods, respectively. Mean variations were denoted via dotted lines accordingly. As a result, the

proposed method provided less variation for Φ_y parameter. Furthermore, significant reductions in F_{ta} were also observed.

V. DISCUSSION

Examining the experiment results, the proposed controller is validated to be efficient in reducing the upper body effort, as it introduces active compliance in each joint to enhance environmental interaction capability. The controller simultaneously processes both position and force constraints while automatically managing the position/force tracking trade-off. Unlike classical low-gain controllers, the compliance increases only if necessary, e.g., when the interaction force error arises due to impact with the ground. When the leg is swinging, the interaction force error is zero and the proposed controller produces no output, prioritizing joint tracking.

Through the utilization of active compliance, the attenuation capacity of forward acceleration and torso angle variations were sufficiently improved. These parameters are regarded to be crucial indications of feasible walking performance as discussed in [32], [33]. Furthermore, a direct measurement of the total force exerted by the arms showed significant decreases. Therefore, the upper body effort could be reduced well compared to the conventional stiff control strategy that overly prioritizes joint tracking.

The active compliance also helped to reduce upper body effort when the walking task was performed within the presence of irregularities on the surface. In an exoskeleton-supported task, the environment may not be always well-controlled; therefore, handling unexpected surface irregularities is of importance. The utilization of active compliance potentially improves the robustness of the walking motion.

In stiffly controlled exoskeletons, it is argued that patients learn to smoothly land their feet by exerting extra force via crutches. Although it helps to deal with ground reaction force peaks, this habit could cause severe shoulder pain over time and may not be maintained over longer walking periods. Moreover, the training period might be relatively longer. Through the direct reduction of upper body effort, such potential risks could be well contained and the training periods may be decreased.

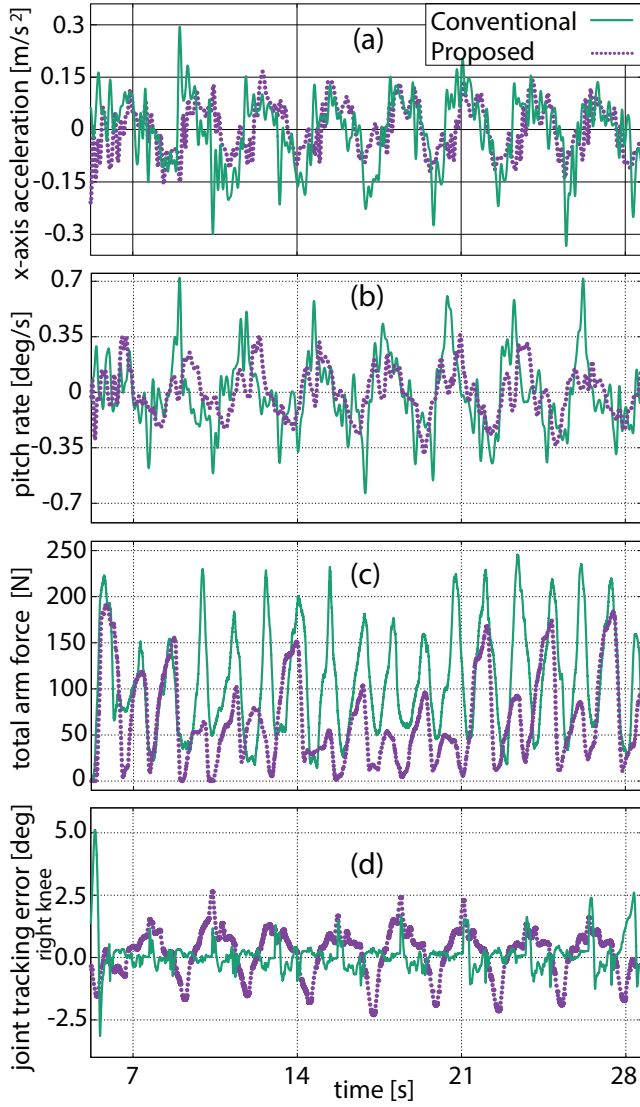


Fig. 11. Experiment #1 results for Subject #1, time-series data. a) forward acceleration, b) pitch rate, c) the total force exerted via the participant's arms, d) joint tracking error, right leg knee joint.

While the proposed controller can introduce certain advantages, the selection of admittance coefficients is not trivial. To this end, we provided a stability analysis to indicate the *boundaries*; however, the optimal selection of these parameters may enable us to fully exploit its favorable features, e.g., enhanced environmental interaction capability. To this end, it is possible to implement a machine learning algorithm to determine the optimal admittance coefficients [34].

The proposed controller utilizes a compensation scheme that accounts for loads due to centrifugal and Coriolis forces, gravity and friction so that the controller bandwidth can be enlarged. Although we experimentally verified friction and gravity compensation as demonstrated in the multimedia attachment, possible parameter uncertainties could affect the system performance in certain working frequencies [35]. Therefore, one can utilize state-of-the-art parameter identification algorithms to avoid possible problems due to modeling uncertainties [36].

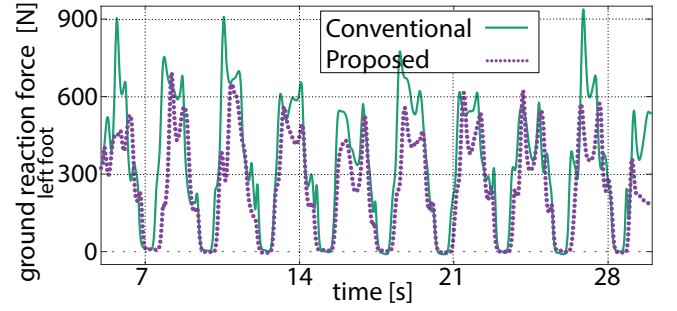


Fig. 12. Experiment #1 ground reaction force results for Subject #1, time-series data for the left foot. Identical results were observed for other subjects.

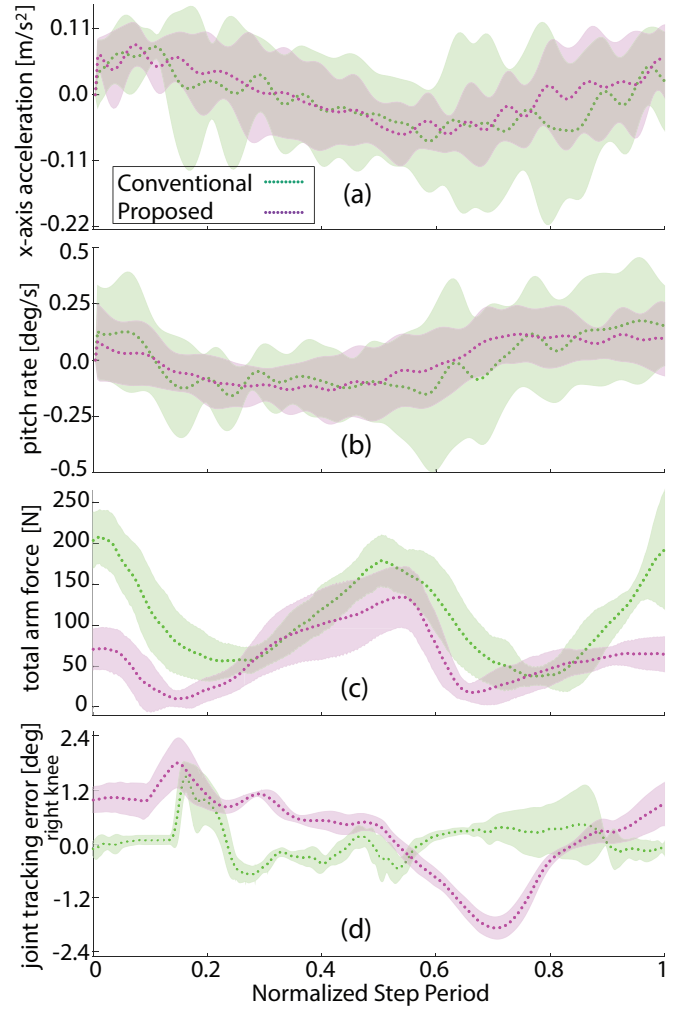


Fig. 13. Experiment #1 results for Subject #1, mean $\pm$ SD variation of 8 steps. a) forward acceleration, b) pitch rate, c) the total force exerted via the participant's arms, d) joint tracking error, right leg knee joint.

As another limitation, the conditions of the experiments could not be randomized, i.e., the subjects always firstly walked while the stiff controller was activated, then they walked while the proposed controller was activated. With this in mind, one may argue that the observed effects could reflect the learning of the subject. Therefore, the current experiment results provide proof of the concept, and further randomized

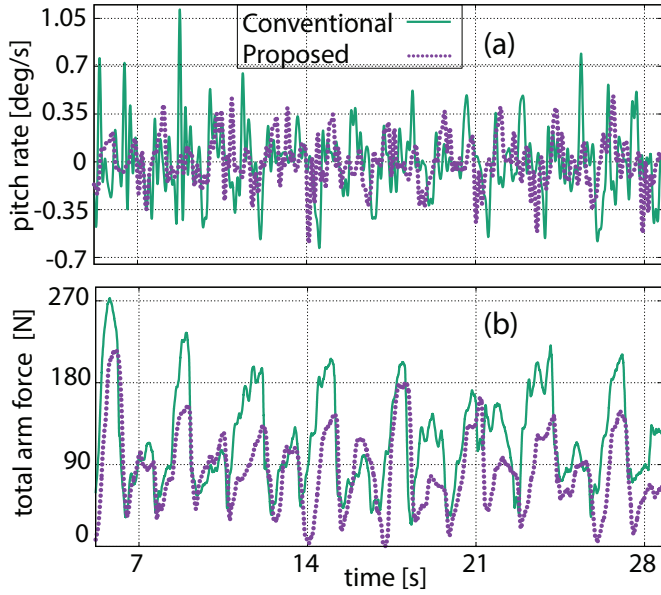


Fig. 14. Experiment #2 results for Subject #3 time-series data. a) pitch rate, b) the total force exerted via the participant's arms.

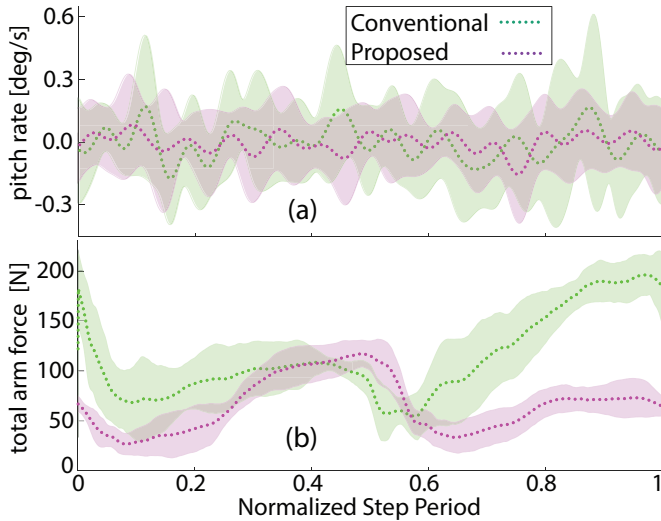


Fig. 15. Experiment #2 results for Subject #3, mean $\pm$ SD variations of 8 steps. a) pitch rate, b) the total force exerted via the participant's arms.

experiments with more subjects will be addressed in our future work.

VI. CONCLUSION

To conclude, we propose a motion controller that is applicable to a great majority of exoskeleton systems that require upper body assistance to maintain the overall balance. Incorporating the active compliance via an admittance control scheme, it is experimentally demonstrated that the external forces that arise during walking can be handled in a physically viable manner even within the presence of surface irregularities.

Having conducted various experiments on able-bodied subjects, we are going to direct our attention to necessary hardware improvements that are required for the clinical trials

on the actual paraplegia patients. We strongly believe that the incorporation of active compliance not only improves the required upper body effort but may also shorten the necessary training period. Therefore, our next work will focus on the exploitation of active compliance from a broader perspective, with the help of clinical trials.

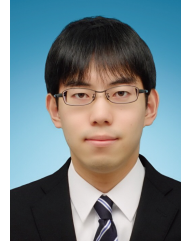
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