Bioinspired Design and Control of BATEX, An Exosuit With Biarticular Compliant Actuators

Ayoob Davoodi, Majid Iranikhah, Arjang Ahmadi, André Seyfarth, and Maziar Ahmad Sharbafi, *Senior Member, IEEE*

***Abstract*—A more human-compatible design of *exosuits* provides new aspects in assisting people. This study presents a novel bioinspired method to design and control an exosuit to support human locomotion. We introduce a methodological design and development of the BATEX as an exosuit with compliant biarticular thigh actuators. By using series elastic actuator (SEA) resembling artificial muscles (AM) instead of direct drive motors in the BATEX, we propose an efficient switching control for the stance and swing phases of walking. Locking the motors in the swing phase converts the SEAs to passive biarticular springs to produce human-like leg swinging. This design is complemented by the force modulated compliance (FMC) controller that uses the ground reaction force to adjust the AM's stiffness in the stance phase. We utilized unassisted walking experiment data to validate the proposed design and control methodology in a simulation study. Optimizing BATEX control parameters with a training set of subjects predicts the energy consumption reduction in the upper leg (knee and hip joints) for more than 27% at five different walking speeds. Finding this measure of assistance level to be more than 21% for the test set of subjects supports the generalization of our approach. Further, individualization of the parameters for each subject shows significant improvement in slow and moderate walking speeds. Pilot assisted-walking experiments with four subjects support the applicability of the introduced method concerning individualization and generalization.**

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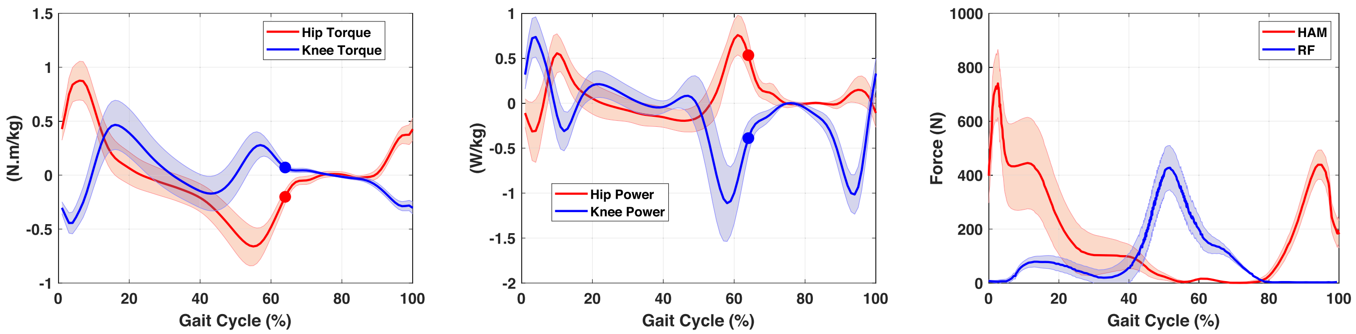
***Index Terms*—Bioinspired hybrid control, biomimetic and bioinspired robotics, exosuit, gait assistance, mechatronic design.**

I. Introduction

SOFT wearable assistive devices, called exosuits, can enhance the physical abilities of both impaired and unimpaired peopleto improve their agility and the adaption to various terrain [1], [2], [3]. The core idea of exosuit design comprises employing muscle-like tension force and removing the rigid skeleton [4], [5]. Careful investigation of the biological musculoskeletal systems reveals other essential features simplifying gait control. The first property is the embedded compliance in the muscle-tendon complex (MTC). Although exosuits are not rigid, they often do not include compliance in their actuation system and mainly rely on soft body tissue [5]. Until now, compliant actuators such as series elastic actuators (SEA), popular in exoskeletons, were not sufficiently investigated in exosuits [5]. The second useful feature is in the biological morphological design of multiarticular muscles, partly implemented in some exosuits [4], [6], [7]. Human-like biarticular actuators affecting two adjacent joints are rarely applied to active exosuits. In [8], biarticular pneumatic muscles were used to support knee and ankle joints like the gastrocnemius muscle of humans. Biarticular thigh springs (like hamstring (HAM) and rectus femoris muscles (RF) of humans) were also tested in passive exosuits [9], [10]. However, biarticular thigh muscles, which are of largest and most effective muscles in humans (especially for balance control) [11], have never been explored in active exosuits. The first contribution of this study is to combine compliance and biarticularity as two bioinspired design features in our *BiArticular Thigh EXosuit (BATEX)*. Because of the negligible overlap between these two muscles' contributions (co-contraction), we expect one motor to be sufficient to derive both actuators, which will introduce another design advantage.

For seamless interaction between humans and assistive devices, we need to predict the user's behavior. In the most bioinspired cases, the robot generates human-like movement patterns [1], [12], which are outcomes of human motor control with no information about how they are generated. Using average values of different human subjects might not even perfectly match each subject's movement patterns. Instead of copying human patterns, understanding the underpinning human locomotion control principles through biomechanical modeling could be a practical solution for developing harmonious *design* and *control*



(a) (b) (c)

Fig. 1. Upper leg (knee and hip) dynamical behavior. The thick line and shaded area demonstrate the mean and standard deviation. (a) Joint torque and (b) power, during walking at 50% of PTS (1.05 m/s), normalized to body mass (adopted from [26]). The circles show the takeoff moment. **(c)** HAM and RF muscle forces during walking, estimated by AnyBody (adopted from [27]).

of exosuits [13], [14]. Neuromuscular control is one of these methods, applied to different assistive devices [12], [15]. These approaches need muscle activity monitoring systems [16], [17] that complicates the sensor network in the assistive devices [18]. The complex neural control makes parameter tuning and individualization (e.g., with human-in-the-loop-optimization [19] extensively intricate. Instead, we will employ modular representation with the locomotor subfunction concept (LSF) [20] and model abstraction through the template and anchor concept [21] to introduce a practical control approach that still keeps the essence of neural control. Within the LSF concept, legged locomotion can be described by three LSFs, namely, *Stance, Swing*, and *Balance* [20]. The BATEX is designed to assist swing and balance subfunctions which could result in less effort and more stability in walking. We use the force-modulated compliant hip (FMCH) model [22] to predict human balance strategy and assist posture control in walking [23]. In this method, the ground reaction force (GRF) will adjust joint (muscle) stiffness. This control concept was successfully implemented on exoskeletons [24] and prostheses [25]. Here, we apply this feedback control scheme for the first time to an assistive device with biarticular thigh actuators, which will be the closest implementation of the FMCH model concerning the interaction between the upper body and leg.

II. BATEX Design Concept

This study is one step toward filling the gap between biology and robotics. We borrowed three concepts from human locomotion to design and control our exosuit: 1) compliant actuators; 2) biarticular arrangement; and 3) an abstract model of reflex-based control using the GRF. This section explains the two first points, and Section IV describes the control concept. We used unimpaired subjects' walking data from [26]. This dataset includes 21 healthy subjects (25.4 ± 2.3 years, 1.73 ± 0.09 m, 70.9 ± 11.7 kg). The subjects walked on a treadmill with integrated 3-D force sensors (Kistler, 1000 Hz) and synchronized high-speed infrared cameras (Qualisys, 240 Hz). Speeds were determined for each subject depending on the preferred transition speed (PTS) from walking to running. The mean PTS for all subjects was 2.07 m/s representing 100% PTS. Subjects walked at 25%, 50%, 75%, 100%, and 125% of PTS.

A. Insight From Human Motor Control

Biomechanical studies demonstrated the low efficiency (the ratio between the produced mechanical energy to the metabolic energy) of the hip and knee joints (about 24%) compared to the ankle joint (61%) [28], [29]. Adding elastic elements could improve the efficiency in the upper leg by storing the negative power and releasing it when needed [29]. Besides, our analyses of human walking at the preferred speed show that hip, knee, and ankle joints generate 23%, 40%, and 37% of the required energy, respectively. The contribution of the ankle is lower in moderate to fast walking [30]. Biological hip and knee joint torque and power patterns (shown in Fig. 1) reveal explicit coordination between the knee and hip joint, which provide more than 60% of the required power in walking [30].

Due to the zigzag arrangement of human leg segments, opposite movement of adjacent joints—observed in different gaits—can be provided by transferring energy from one joint to another [11]. This is supported by reciprocal torque profiles of the hip and knee joints in the swing phase (60% to 100% of the gait cycle). Similar behavior is observed at the beginning and end of the stance phase. Thus, a biarticular thigh mechanism could efficiently support knee and hip joints in these periods.

B. Biarticular Actuation From Human to BATEX

The human leg muskuloskeletal system can be represented by a combination of nine major muscle groups, including six monoarticular and three biarticular muscles [31]. Inspired by the targeted energy management at hip and knee joints, we focus on the functionality of the HAM and RF as biarticular thigh muscles [see Fig. 1(c)]. The HAM muscle has maximum contribution in early stance and late swing, while the significant contribution of the RF muscle is at late stance and early swing (no coactivation). This is consistent with the previous prediction of biarticular muscle contribution from the knee and hip joint torques presented in Section II-A. As a result, the BATEX should

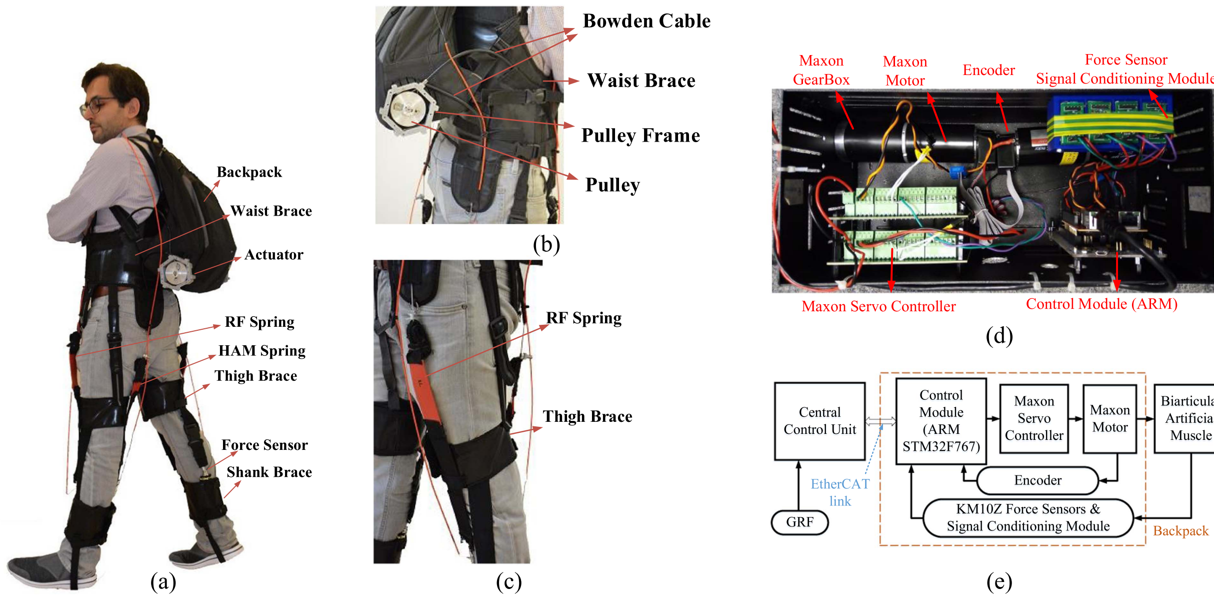


Fig. 2. BATEX exosuit system. (a) Side view of the worn exosuit. The SEA arrangement includes one motor and two serial springs for biarticular thigh actuator for each leg. The subject gave his consent for the publication of the photo. (b) Transmission mechanism using pulley, 3-D printed plastic frame, and the guided Bowden cables. (c) The thigh brace to guide the artificial muscle and keep the lever arm. (d) Mobile actuation system placed in the backpack. (e) Hardware and low level control architecture.

mainly support human walking in the beginning and the end of each of stance and swing phases. Part of this behavior is resulted from motion dynamics and muscle stretch shortening. Therefore, we can see the following.

1) The BATEX exosuit with biarticular thigh actuation can support hip and knee joint in walking.
2) The exo actuator design can be optimized by including an elastic element to manage energy within time domain.
3) One motor is sufficient to generate both HAM and RF muscle-like force in exosuits.
4) We could use a same phase-based controller to control the two artificial biarticular muscles of the exosuit.

III. Design of BATEX Hardware Setup

In this section, we detail the BATEX hardware design.

A. Mechanical Design

BATEX consists of two main parts: 1) the soft wearable suit; and 2) the mobile actuation system [see Fig. 2(a)]. The actuation box is placed in a backpack, including motors, gears, electronic boards, and pulleys. To reduce the weight of the exosuit, we use one motor per leg for both biarticular thigh artificial muscles, as shown in Fig. 2(a). The textile components of BATEX consist of a waist brace, two thigh braces, and two shank braces. The HAM and RF serial compliant arrangements consist of rubber bands, Bowden cable, force sensor, and connecting straps. Bowden cables transmit the force from the pulley to the serial elastic elements [HAM, RF; Fig. 2(b)]. On the top side, the Bowden cable sheath is connected to the pulley cover frame, and the inner cable is attached to the pulley. The Bowden cables are guided through the predefined paths (To tune the hip lever arm) in the waist brace, and the inner cable is connected to the rubber band. The total mass of the exosuit is 5.6 kg, which does not have a significant effect on changing walking behavior due to the distribution of the exo mass mainly close to the body CoM (see a related analysis in [32]). The detailed specification of the BATEX design and actuation box parameter selection are detailed in the Supplementary Material.

The artificial muscles are directed through the thigh brace [Fig. 2(c)] to connect the upper body to the shank. This part can align the elastic element with human limbs and prevent exerting unpleasant forces, e.g., from slipping the RF spring over the knee (patella). In addition to concentrating the SEA force in the sagittal plane, this component keeps the knee lever arms of both artificial muscles around 4 cm, as shown in Fig. 2(a). We prevent thigh brace rotation using a textile strap connecting the thigh and waist braces. This strap is aligned with the femur, as shown in Fig. 2(a) and (b). Besides the lever arm, the maximum deliverable force and the required bandwidth are critical factors in designing the biarticular SEAs. The spring's stiffness is an important parameter that significantly affects these factors. To support human motor control, and inspired by our previous modeling studies [33], we set the HAM and RF artificial muscles to 3.3 kN/m and 1.1 kN/m, respectively. As the artificial muscles produce upward forces, the shank brace's downward movement is not the main issue. However, we limit it by connecting the shank brace to the thigh brace with a strap. To minimize upward (and rotational) movement, we tie up the shank brace with straps passing under the shoe sole.

B. Electronics

The main electronic modules are placed inside the actuation box [Fig. 2(d)]. The actuator control module, an ARM microcontroller (Nucleo STM32F767ZI), collects force sensors' and encoders' data and sends them to the central control unit (CCU). The EtherCAT link communicates between CCU and the control

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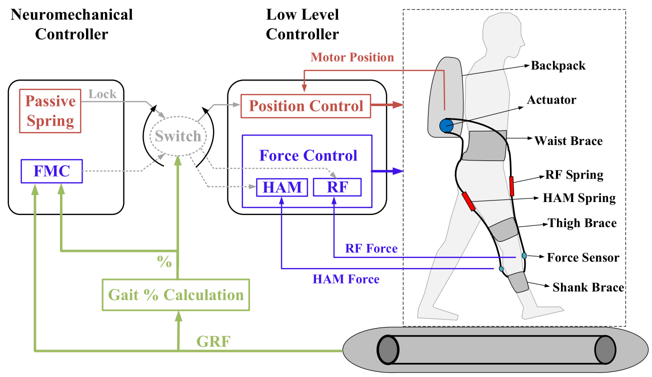


Fig. 3. Control block diagram. Based on gait phase calculation using the GRF, the neuromechanical controller selects between FMC and passive spring. The command will be executed in the low level as force (RF or HAM) or position control.

module [Fig. 2(e)]. For position control, we use the Maxon Motor encoder attached to the back of the motor. The KM10Z force sensor, attached to the shank brace, measures the SEA force for force control. Four force sensor signal-conditioning boards are developed to amplify and filter the force sensor outputs [Fig. 2(d)]. The real-time CCU module is implemented on an off-board ARM microcontroller (STM32F767ZI). The CCU receives mobile actuation system data and the GRF, to calculate desired control commands. The instrumented treadmill (ADAL-WR, HEF Tecmachine, Andrezieux Boutheon, France) measures separated GRFs for different legs. The BATEX sensory information will be sent to a PC to be monitored and stored via LabView software. The delay in GRF measurement to be used for control is 5 ms.

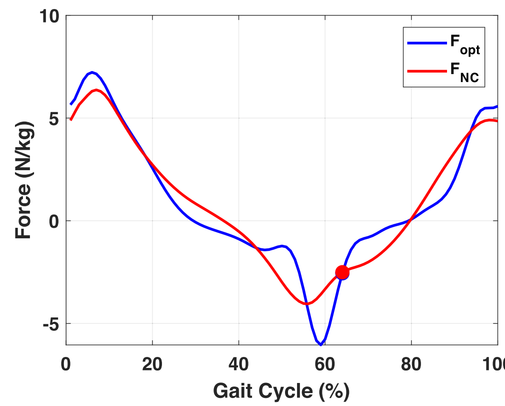
IV. Control

The BATEX bioinspired control approach benefits from the morphological properties of biarticular actuation using GRF feedback, (shown in Fig. 3). In the stance phase, the force modulated compliance (FMC) method [22], [33] is employed using *force control* in the inner loop. In the swing phase, the motor is locked to simplify the BATEX to passive biarticular springs using *position control*.

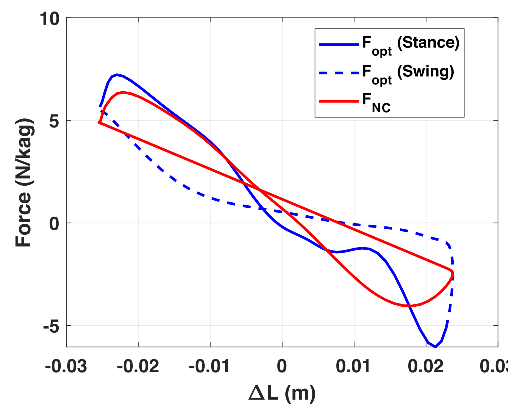
A. Neuromechanical Control

Based on the point-by-point optimization procedure (details, in the Supplementary Material), the optimal muscle force F_{opt} that provides the maximum assistance level with the BATEX was calculated. The blue curve in Fig. 4(a) demonstrates F_{opt} for walking at 50% of PTS during one gait cycle. The positive forces in the first half of the stance phase and the second half of the swing phase suggest the contribution of the HAM actuator. The RF artificial muscle should contribute to the second half of the stance and the first half of the swing phase.

Here, we define *biarticular artificial muscle (BAM)* as one component including both RF and HAM muscles. The force-length relationship is analyzed to find the optimal BAM force (F_{opt}). Based on Fig. 4(b), we could identify an impedance (stiffness) control for the BAM of the BATEX. Let ϕ_h and ϕ_k be the hip and knee angles, respectively. Assuming that the



(a)



(b)

Fig. 4. Optimized biarticular thigh muscle force F_{opt} to generate maximum assistance level with the BATEX and its approximation with the neuromechanical controller F_{NC} for walking at 50% PTS (1.05 m/s). (a) The force pattern during the gait cycle. The correlation between two graphs is 96%. The circle shows the initiation of the swing phase. (b) Force-length relationship to approximate the desired stiffness of the biarticular artificial muscles during the gait cycle. ΔL is the deviation of the artificial muscle from its rest length.

BAM reaches its rest length at midstance, the upright trunk and stretched knee determine the hip and knee rest angles (ϕ_h^0 and ϕ_k^0). Thus, the BAM length changes will be given by

$$\Delta L = r_h(\phi_h - \phi_h^0) - r_k(\phi_k - \phi_k^0). \tag{1}$$

Fig. 4(b) shows that, unlike the stance phase, the previously suggested passive BAM suffices to approximate the optimal force in the swing phase. Changing the force-length slope in the stance phase reveals the need to adapt the stiffness of the spring. The FMC utilizes the GRF (denoted as F_{GRF}) to adapt the muscle stiffness. Instead of muscle force feedback (not measurable) commonly used in neuromuscular models (e.g., [13], [34], [35]), in our control approach, we use the GRF. The outcomes of all muscle forces variation will affect the GRF, which makes this signal informative enough to be used as a control feedback signal. We define neuromechanical controller (F_{NC}) by combining the controllers of stance (with variable stiffness) and swing (with constant stiffness) phases.

$$F_{NC} = c_1 F_{GRF}(L - L_{st}^0) + c_2(L - L_{sw}^0). \tag{2}$$

Here, L, L^0_{st}, and L^0_{sw} are the BAM length, variable spring rest length (for the stance phase), and fixed spring rest length (for the swing phase), respectively, while c_1 and c_2 are constant. With the least mean square error optimization method, all parameters can be calculated to reach maximum similarity between F_{opt} and F_{NC}. Fig. 4 shows that the proposed controller can acceptably predict the optimal force with a correlation of 96%. In the Supplementary Material, we introduce a practical method to implement the neuromechanical control (2) using the existing sensory data. Noticeably, we do not need extra sensors to measure other variables, such as BAM length.

B. Low Level Control

The vertical GRF of each leg is measured individually using an instrumented treadmill for gait phase detection and as a feedback signal in the FMC control. As shown in Fig. 3, the desired low-level controller will apply 1) *force control* of a) HAM artificial muscle from the heel strike to midstance; and 2) the RF artificial muscle from midstance to take off; and 3) position control during the swing phase. In addition to supporting actuation and sensing, the system provides real-time feedback control and data logging at 1 kHz.

The low-level control hardware architecture that has been implemented in an ARM microcontroller [shown in Fig. 2(e)] includes two cascaded controllers. The output for both position and force control will be the desired reference speed for the motor driver. The lowest level is the speed control of the Maxon Servo Controller using an autotuning PID controller.

For the force control in the stance phase, we follow the proposed SEA torque control method suggested by Vallery et al. [36]. For both position and force control, a PID controller is sufficient to complement the Maxon speed controller. The measured BAM forces are sampled with 1 kHz frequency, and then filtered with a 50 Hz IIR filter. A signal conditioning circuit is used to amplify and filter the measured value of HAM and RF artificial muscle [see Fig. 2(d)]. Before doing the human-assisted walking experiments, the required control performance was assured in different tests, described in the Supplementary Material. For the motor position control, an incremental encoder placed at the back of the motor is utilized.

V. Results

This section describes the optimization and experimental results of gait assistance with BATEX. First, the potentials of gait assistance using the BATEX and the possible achievements with our neuromechanical controller are presented. Then, the role of individualization and parameter tuning will be investigated. These outcomes are from analyzing unassisted human walking data at five different speeds [26]. Finally, the results of the assisted walking experiments demonstrate the proposed method's applicability.

Using the artificial thigh muscles, the BATEX with the optimal force can reduce the hip and knee energy of walking at 50% of PTS for more than 50%. Defining *upper leg* energy as the total biological work of knee and hip joints, around 54% reduction is achievable at this speed. We already demonstrated that with our neuromechanical controller, the optimal BATEX force (F_{opt}) could be acceptably regenerated [see Fig. 4(a)]. In the following, we explain the effectiveness of the neuromechanical controller in reducing human energy consumption with the BATEX.

TABLE I
Predicted Reduction of Human Energy and Joint Torque in Training (Tr, 15 Subjects) and Test (Te, 6 Subjects) Datasets With BATEX Using Neuromechanical Controller at Different Speeds (%PTS)

% of PTS	25		50		75		100		125	
Reduced item	Tr	Te	Tr	Te	Tr	Te	Tr	Te	Tr	Te
Hip energy	25	21	41	44	32	20	16	12	13	4
Knee energy	30	28	36	33	34	32	33	32	36	35
Upper leg energy	28	25	38	36	33	28	27	21	27	25
Hip torque	29	31	47	50	43	39	33	25	31	26
Knee torque	6	3	5	3	12	12	18	18	21	24

Note: The control parameters are fixed. The upper leg energy is the summation of hip and knee energy.

A. Walking Assistance With Fixed Control Parameters

We divided the walking dataset (21 subjects) into training (15 subjects) and test datasets (6 subjects) which were selected randomly. First, we find an optimal parameter set of the BATEX controller for the training dataset for each walking speed (percentage of PTS). The method, which is based on the least square error optimization, is described in Section I of the Supplementary Material. Then, the same optimal parameters are applied to the test dataset with the same speed group (percentage of PTS). The results of 50% of PTS are depicted in Fig. 5. The neuromechanical controller can nicely provide the required hip power and torque. As expected, the resulting knee torque in the middle of the stance phase (10% to 50% of the gait cycle) deviates from that of human walking.

We calculate the energy reduction in the hip, knee, and their summation (upper leg), which can be obtained with the BATEX controlled by the neuromechanical controller. The results for all five speeds are summarized in Table I. The upper leg energy reduction is between 27% (for 100% of PTS) and 38% (for 50% of PTS). The energy reduction in the training is slightly higher than that of the test dataset.

B. Individualization

We optimized the control parameter for each subject to investigate the parameter adaptation's effect based on the subject's movement characteristics. Fig. 6 compares the mean energy reduction level having individualized parameters (for all 21 subjects) with the fixed parameters (train and test data). The level of assistance will significantly increase at slow walking (16% at 25% PTS and 9% at 50% PTS).

To better understand the parameter adaptation concerning the motion speed, Fig. 7 shows the mean and variance for the individualized parameters (from (2)) at different speeds. The first term in this equation (including c_1 and L^0_{st}) defines the variable spring, and the second term (including c_2 and L^0_{sw}) relates to the fixed spring. Monotonic changes can be observed in all coefficients except for L^0_{st} from 100% to 125% of %PTS.

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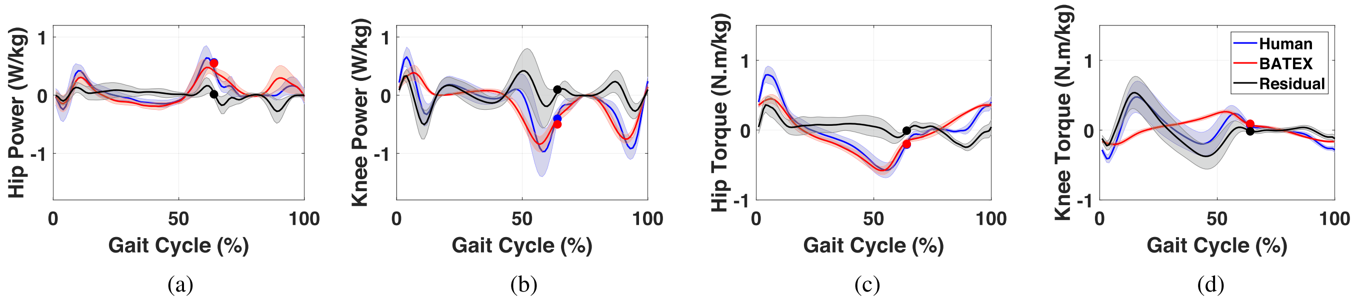

Fig. 5. Optimal neuromechanical control of BATEX to support hip and knee joints in human walking at 50% PTS for test (6 subjects) dataset. The control parameters are fixed for all subjects and optimized based on the train dataset (15 subjects). Human walking data (blue) from [26] is considered as a reference. The prediction of BATEX contribution at each joint and the residual (difference between reference and BATEX) are shown with red and gray colors, respectively. The thick line and the shaded area show the mean and the standard deviation for the target dataset. The circles show the initiation of the swing phase. (a) $R = 92\%$. (b) $R = 88\%$. (c) $R = 93\%$. (d) $R = 33\%$.

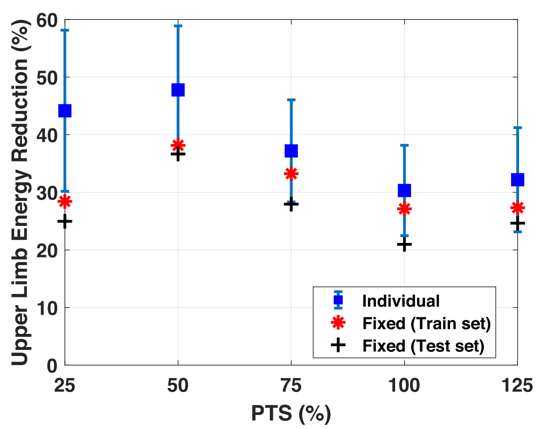

Fig. 6. Comparison between fixed and individualized parameters of BATEX control for walking at different speeds.

C. Assisted Walking Experiment

The developed BATEX and its neuromechanical controller were tested in two experiments to explore individualization and generalization. In total, we conducted experiments with four healthy subjects (2 males, 2 females, 31 ± 3 years, 70 ± 9 kg, and 172 ± 7 cm) with no gait abnormalities. The subjects participated in the study after signing the consent sheet approved by TU Darmstadt ethical committee. All experiments involved a 20-min warm-up walking session to help subjects adapt to the experimental setup, followed by 5-min walking sessions.

In the first experiment, which targeted individualization, we explored a range of stiffness values and control parameters for one male subject walking at his preferred speed (1.1 m/s). Based on the user's oral feedback, we could find a specific parameter set that could provide maximum assistance. The attached video shows how the neuromechanical control of BATEX works. The second experiment addresses the generalization of the proposed methods. We investigated whether a fixed set of parameters could support the other three subjects. So we did not optimize control and spring parameters for individual subjects. User feedback shows that the same set of parameters could assist this new group of subjects. Fig. 8 presents the experimental outcomes for the average values of all four subjects. Fig. 8(a) shows the GRF used for gait phase detection. It is observed that the kinetic behavior of assisted walking resembles that of regular walking. In Fig. 8(b), the measured force profiles of HAM- (red line) and RF-artificial muscles (blue line) are illustrated. The average patterns of four subjects' forces are comparable with human muscle force profiles, shown in Fig. 1(c). The variance of the exo force among subjects is about 10% of the peak force, which supports the applicability of the method for different subjects. We utilize the motor encoder and the force sensors' data to analyze the kinematic changes of the biarticular artificial muscles. Knowing (HAM and RF) spring force and stiffness, the spring deflection (ΔL) is calculated. The force-length behavior of the BAMs in BATEX is drawn in Fig. 8(c). The pattern is comparable with the desired behavior targeted in the neuromechanical control [Fig. 4(b)]. Finally, Fig. 8(d) supports the generation of desired assistive torque patterns predicted by analyzing human walking experimental data at a similar speed, shown in Fig. 5. Using the BAMs' force and length (consequently the velocity), the power of the artificial HAM and RF is calculated, as shown in Fig. 9. Based on the SEA design of the BATEX actuators, the BAM instantaneous power is the summation of the motor power and spring power.

VI. Discussions

Leg morphology and neuromuscular control are essential outcomes of evolution. Biarticular muscles are remarkable morphological features of biological leg design that facilitate locomotion control. Interestingly, the compliant design of biarticular muscles turns them to be used as sensors and actuators at the same time [11]. For example, leg orientation with respect to the upper body can be measured by the deflection of thigh biarticular muscles to be used for control (see Section II of the Supplementary Material). Our findings on using biarticular arrangements and the matching neuromechanical controller for gait assistance are discussed in the following.

A. Design Aspects

To mimic the human locomotor function and corresponding motor control, we introduced the idea of biarticular compliant actuation to support hip and knee joints. First, we investigated

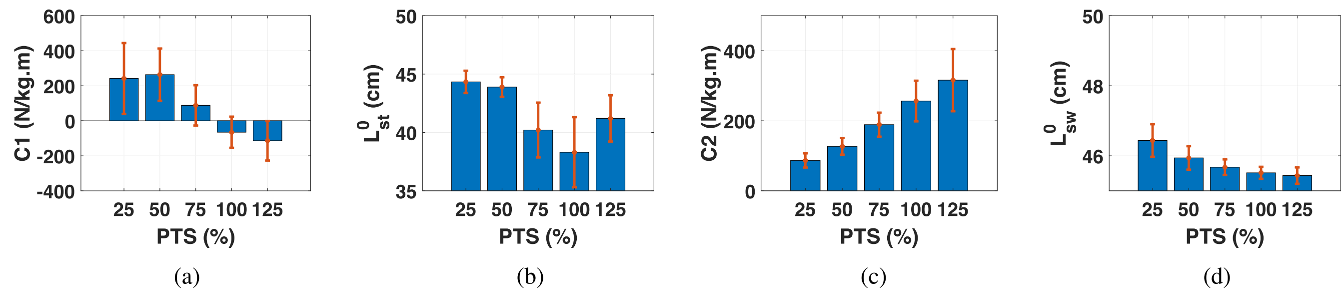

Fig. 7. Progression of the individualized neuromechanical control parameters (2) of BATEX with walking speed. Bar chart (blue bar) and error bar (red line) show the mean and the standard deviation for the control parameters.

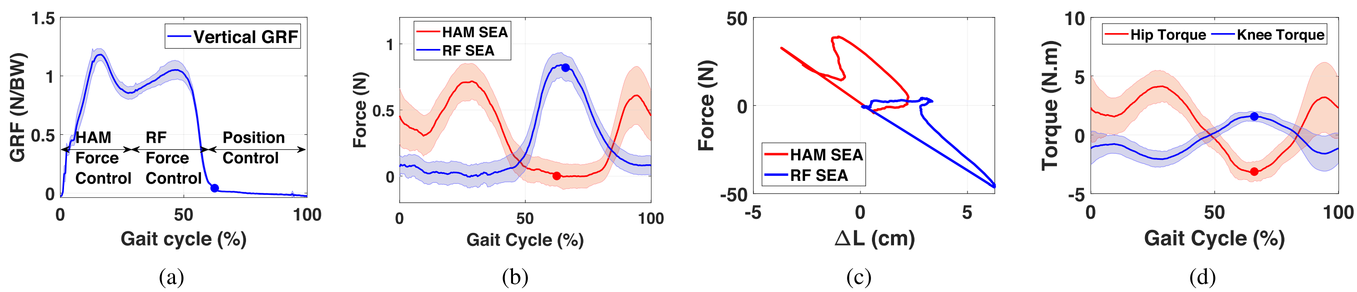

Fig. 8. Experimental result of BATEX for assisted walking at 1.1 m/s (a) Ground reaction force. (b) Measured force profiles of biarticular artificial muscles (BAMs) from assisted walking experiment. (c) Force-length relationship of BAMs. (d) Provided hip and knee torque from BATEX. The circles show the initiation of the swing phase.

biarticular thigh muscles' functionalities in Fig. 1(c) and how they can support the hip and knee joint torque and power management in human walking (see Section I of the Supplementary Material). Our analyses confirm the effectiveness of the BATEX design. The optimizations showed that the BATEX could potentially provide between 40% to 55% reduction in the required biological energy of the upper leg. Assuming an unchanged contribution from the ankle joint, which is about 37% of the total consumed energy on average [30], we envision about 25% to 35% reduction in the whole leg energy consumption, as a rough approximation of the metabolic cost. Our recent experimental results with a passive version of the BATEX exosuit showed 11% metabolic reduction in 11 subjects, which could partially support our predictions. Human reaction to assistance may reduce these predicted assistance levels. Anticipating human reaction to external forces to adapt assistive devices is still an open challenge [12], [37]. Alternatively, further elaborate experimental studies to approach such an assistance level are required. This article focuses on design and control, while the experimental support of biological advantages (with several subjects) remains a future step.

Actuation architecture is another crucial design feature examined in this study. Using one motor to actuate both RF and HAM artificial muscles enables the reciprocal co-contraction of these antagonistic muscles. This way, we can provide significant support for two joints in two directions with only one motor. A cheaper and lighter exosuit with the proposed actuator design is successfully attained in the BATEX.

Unlike the common method of using direct drive motors in exosuits [5], we employed elastic actuation. Based on the human movement analyses, we first demonstrated the advantages of passive springs, especially in the swing phase. These findings are in line with our recently developed passive exosuit [9]. Second, the effective contribution of the springs in the biarticular artificial muscle (BAM) power was supported by experimental results. Fig. 9 shows that the serial spring is the main source of the required BAM power. In 10%–30% of the gait cycle, the contribution of the motor is more than the spring but still comparable. The serial RF spring significantly contributes to the late stance by supporting upper body forward movement (balance function) and body weight compensation (stance subfunction). This contribution provides hip flexion (negative) and knee extension (positive) torque at the late stance, visible in Fig. 8(d). Storing the RF spring energy in the late stance also supports the forward movement of the leg in the following swing phase.

Engineers and biomechanists express that smart mechanical design can simplify control [21], [38]. This mechanical intelligence could be in different levels such as design morphology [39] or actuation system [40], [41]. This kind of body intelligence [38] was also embedded in our exosuit design by carefully determining the mechanical parameters. The hip-to-knee lever arm ratio is one of the morphological design aspects defined based on human locomotion investigations and mathematical modeling analyses. Finding lever arm ratio 2 as the optimal value from swing leg motion analyses later reveals a geometrical relation between the trunk-to-leg angle variation and the biarticular muscle length changes when the shank and thigh segment lengths are comparable [41]. In the human body, a similar lever-arm ratio in HAM muscle (1.33 for RF) [42]

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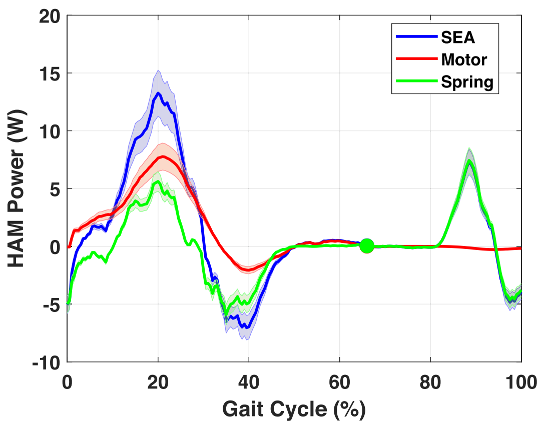
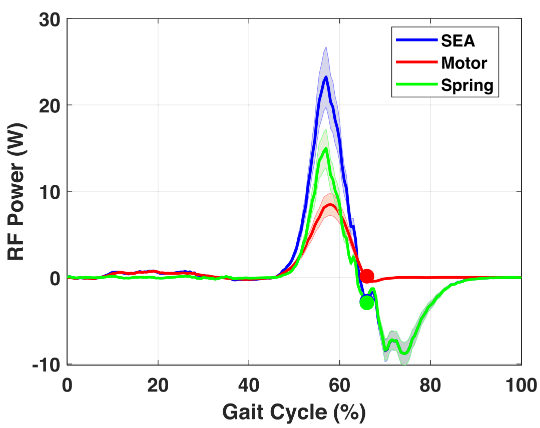


(a)



(b)

Fig. 9. Power profiles of BATEX SEAs during assisted walking at 1.1 m/s. The circles show the initiation of the swing phase.

could also enable this biarticular muscle to measure the angle between the upper body and the leg. Therefore, these muscles can support balancing in the stance, and leg direction control (foot placement) in the swing phase. We could appropriately replicate human hip and knee torque in the swing phase by tuning the hip-to-knee lever arm ratio to 2 in BATEX design. Further, we utilized this relation to implement our force modulated compliance (FMC) control (details in the Supplementary Material). In short, we implemented the body intelligence concept by simplifying control, using a proper selection of the lever arm ratio as a mechanical design parameter.

B. Control Aspects

The optimizations with the conceivable muscle forces could show the maximum assistance level of the BATEX. However, it is only usable with a control method that can predict the desired BAM force during walking. We introduced the neuromechanical controller to estimate the optimal force f_{opt} [see Fig. 4(a)]. There is no limitation in implementing other control methods on BATEX, and other methods might better approximate the optimal force. Nevertheless, by selecting our bioinspired hybrid control, we aim at two goals: 1) introducing a method that matches human motor control; and 2) benefiting from the design features to minimize the motor contribution. We described the second goal in Section VI-A. In the following, we explain more about the first goal.

Employing passive biarticular springs for swing leg control was already identified in our previous studies on human walking [43], robot modeling and experiments [31], simulating assistive devices in neuromuscular models [33], and passive exosuit development [9]. The FMC control was also invented based on human walking analyses [22] and then, was examined in developing balance strategies in 3-D [44] walking models. Later, this method was extended to neuromuscular level [45] and was tested in assisted walking using the rigid exoskeleton LOPES-II [24]. Some advantages of combining these two control techniques (for stance and swing phases) for gait assistance were shown in our previous study using OpenSim simulations [33]. Based on all these findings, we expected that our hybrid control approach would match human walking control. To further investigate these expectations, we performed different analyses in this study regarding generalization, individualization, adaptation to walking speed, and implementation challenges.

1) Generalization: For evaluating the generalization aspect of our neuromechanical controller (NC), we considered a training set to extract the control parameters. We then examined this fixed-parameter set on a test dataset. The hip power and torque in both training and test data confirm that the proposed method matches human hip control, visible from low residuals in Fig. 5. It is also observed that the exosuit cannot sufficiently support the knee joint in the stance phase. Especially around 15% and 50% of the gait, the residual knee power and torque are slightly higher than the unassisted power and torque. The extra effort of the human knee joint will be transferred to the hip joint to support balancing. Nevertheless, the BATEX with neuromechanical control can significantly reduce knee energy and (average) torque at all five walking speeds. In total, the NC with a fixed parameter set can reduce upper leg energy by 27% to 38% for the training dataset and by 21% to 37% for the test dataset (see Table I). As illustrated in Fig. 6, the best generalization of the approach with fixed control parameters is observed in walking at 50% of PTS. The insignificant reduction in assistance level from train to test dataset supports fixed control parameters' applicability. Hence, using the proposed approach, the BATEX could competently assist humans in walking at different speeds, even without parameter adaptation.

2) Individualization: In order to predict how parameter adaptation can improve assistance quality, we optimized the NC parameters for each subject. Fig. 6 shows that individualization can generate remarkable improvement for the two lower speeds in which the difference between test and train data was insignificant. The highest assistance level of the BATEX with the neuromechanical controller is about 50% at 50% of PTS. This means that theoretically, with a learning approach (e.g., human-in-the-loop-optimization [19]), half of the required energy at hip and knee joints can be provided by the exosuit which needs further investigation in assisted walking experiments.

3) Adaptation to Walking Speed: The control parameters variations in Fig. 7 show that the passive spring should be stiffer (increase in c_2) to generate faster walking. However, the normalized stiffness of the variable spring c_1 reduces with speed, and interestingly, its sign changes after 75% PTS to support fast walking. Thus, this active module behaves as an actuator rather than a variable spring. A comparable phenomenon was reported by finding negative leg stiffness in the second half of the stance phase of unassisted fast walking (100% and 125% PTS) in [20]. This observation can be described by the paradoxical muscle movement theory presented for human standing. Loram et al. [46] explained that due to the lack of intrinsic ankle muscle stiffness, the soleus and gastrocnemius muscles move paradoxically, e.g., by shortening when the body sways forward. Similarly, our investigations show that despite the considerable increase in the fixed spring stiffness (c_2) of BATEX with gait speed increment, it is insufficient to support fast walking. Then the negative stiffness of the force-modulated spring (c_1) can compensate for the limitation of the fixed spring stiffness. This finding shows another feature of the proposed bioinspired control matching human motor control.

4) Proof of Concept Experiments: This study does not focus on biomechanical investigations. However, in our assisted walking experiments, we demonstrated that the controlled BATEX could be easily adapted to human walking, as shown in the attached video. In the first experiment with one subject, we tuned the mechanical (spring) and control (FMC) parameters based on user feedback to test individualization with BATEX. Interestingly, the mean control parameters extracted from unassisted walking experiments (Fig. 7) were close to the user-selected RF stiffness in our assisted walking experiment. The preferred values for the other three subjects in the second experiment were different from these values, which demonstrates the possibility for individualization with simple parameter adjustment. However, a fixed set of parameters (not optimized for each subject) could provide assistance feeling for the second group of subjects, which supports the generalization of the proposed design and control approach. The measured data in our experiment (shown in Fig. 8) is consistent with the design goals. We expect that one predefined control (FMC gains) and design (spring stiffness and rest length) parameter set could provide a moderate assistance level for the majority of subjects, and individualized parameter tuning could improve the assistance level. The performance of the developed exosuit needs experiments with additional measurements (kinematic, EMG, and metabolics) and with a higher number of subjects.

5) Implementation Challenges: The main challenges in implementing the proposed control on BATEX emerge in the force modulated compliance approach. The BAM force and length are required to control its compliance. We used the tensile force sensor to measure muscle force. Measuring the muscle length is more complicated. We employed the trunk-to-leg angle (detailed in the Supplementary Material) as a simple way to approximate muscle length. Another option is to use the motor encoder data and spring force (and stiffness) to estimate the BAM length more precisely. This method was utilized to draw the force-length relation in Fig. 8(c). Its online implementation for control needs modification due to its sensitivity to the variations in measured force.

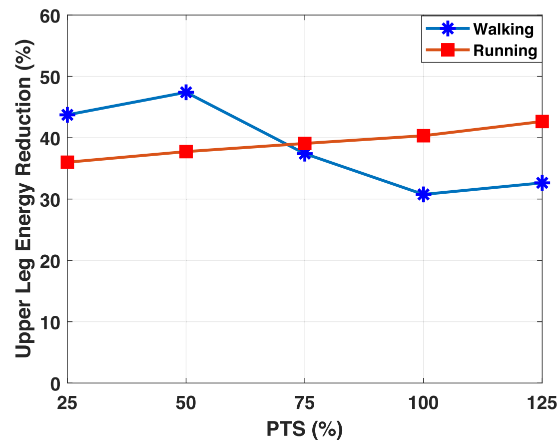


Fig. 10. Predicted energy saving in gait assistance with BATEX at different speeds. Gait speed is represented as percentages of preferred transition speed (PTS) between walking and running.

Another challenging measurement is the ground reaction force. We utilized the force data from the instrumented treadmill, which is not applicable for overground walking. For extending the GRF-based control to ground walking, using insole sensors are the potential solutions to be investigated in the future. Recent advancements in insole sensors, e.g., using magnetic fields [47], [48], or high-sensitivity capacitive sensors [49] which are more robust than the traditional insole force sensor (concerning wear, imprecision, and low sensitivity) are potential solutions for extending the BATEX.

C. Extensions to Assisted Running

Gait assistance is not limited to walking; the same design and control concepts should also work for other gaits, such as running. Our preliminary modeling analyses of assisting running with BATEX revealed substantial advantages. In Fig. 10, we compared the predicted upper leg energy reduction in walking with that of running at different speeds. The experimental running data (from [26]) has a similar experimental protocol and subjects to that of walking data. The first observation is a monotonic increase in energy reduction with respect to running speed (red curve), while the assistance level in walking is reduced after 50% of PTS. The assistance level for running and walking at 75% is comparable, and running at 100% of PTS can be supported by BATEX about 10% more than walking. Thus, the BATEX with NC controller can support running more than walking at speeds that humans prefer running. Hypothetically, this observation might result from the similar essence of BATEX neuromechanical control to human motor control. This matching between human gait selection at different speeds and the assistance level of BATEX supports our design and control methodology.

VII. Conclusion

This study is the first step towards applying a new bioinspired perspective to design and control soft wearable assistive devices.

We demonstrated the potential advantages of the biarticular compliant actuation and a novel neuromechanical control technique to assist human gait. We demonstrated the individualization and generalization levels of the exosuit, first in simulations and then in experiments. The BATEX is the first exosuit developed based on this methodology, with ample possibility for modification e.g., by using a clutch mechanism to disengage the motors at takeoff and engage at touchdown [50] or other combinations of compliance and motors such as parallel elastic actuation [51] (PEA) or serial-parallel elastic actuators [52] (SPEA). Nevertheless, we envision that the BATEX design and control methodology can be introduced as a consolidated technique for assisting humans in a wide range of locomotion tasks.

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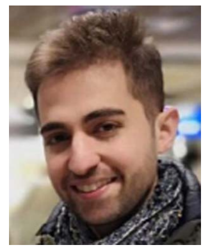

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