From a biological template model to gait assistance with an exosuit

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Abstract

The invention of soft wearable assistive devices, known as exosuits, introduced a new aspect in assisting unimpaired subjects. In this study, we designed and developed an exosuit with compliant biarticular thigh actuators called BATEX. Unlike the conventional method of using rigid actuators in exosuits, the BATEX is made of serial elastic actuators (SEA) resembling artificial muscles. This bioinspired design is complemented by the novel control concept of using the ground reaction force to adjust the artificial muscles’ stiffness in the stance phase. By locking the motors in the swing phase, the SEAs will be simplified to passive biarticular springs, which is sufficient for leg swinging. The key concept in our design and control approach is to synthesize human locomotion to develop an assistive device instead of copying human motor control outputs. Analyzing human walking assistance using experiment-based OpenSim simulations demonstrates the advantages of the proposed design and control of BATEX, such as 9.4% reduction in metabolic cost during normal walking condition. This metabolic reduction increases to 10.4% when the subjects carry a 38 kg load. The adaptability of our proposed model-based control to such an unknown condition outperforms the assistance level of the model-free optimal controller. Moreover, increasing the assistive system’s efficiency by adjusting the actuator compliance with the force feedback supports our previous findings on the LOPES II exoskeleton.

1. Introduction

Lower limb exoskeletons for assisting people with decreased locomotion abilities have attracted the attention of researchers [1] and the healthcare industry [2]. Lower limb exoskeletons are also helpful for other categories of potential users, like firefighters, laborers, and soldiers. Recent studies show societal concerns about aging. For example, muscle force and power decrease by aging and assistive devices are demanded for reducing elderly and patients locomotion costs [3].

Exoskeleton design goals include facilitating human locomotion, reducing the injury risk, enhancing users’ mobility, and improving the ergonomics of the work conditions [6]. One quantitative method for evaluating the functionality of assistive devices is to measure the metabolic cost [7]. Besides any of the design goals, reducing metabolic cost could be another essential target or a side product in the aforementioned gait assistance cases and situations. In recent years, researchers have succeeded in significantly improving the metabolic cost reduction in human gaits using powered exoskeletons [1, 8, 9]. Powered exoskeletons can be categorized as (1) traditional ones with rigid structures connected to the body in parallel to the biological skeletal system (limb) [10], and (2) recently introduced exosuits using soft materials such as textiles [11, 12]. The ability to produce high torques could stabilize gaits in people with severe injuries (e.g. spinal cord) and could also support the applicability of this first group in rehabilitation [13]. In comparison, having a lightweight and low inertia, portability, and bioinspired design are important features of the wearable exosuits. This later property results in more comfort and fewer constraints in movement by resolving misalignment between the subject’s and
robot’s joints and the joints of the assistive device [11]. Therefore, one may argue that soft exosuits are more appropriate than the rigid exoskeletons to be used by unimpaired human subjects for daily activities.

Because of tight interaction between humans and robots, control of the assistive devices becomes very crucial. Control strategies can be divided into model-based and model-free approaches. Model-free control methods include (but not limited to) time-based (predefined trajectory tracking based) approaches [16, 17], predefined gait-pattern based control [18], and EMG-based control [19]. These so-called model-free methods [20] will have a high variance in walking economy among participants for fixed control strategies [9]. This problem can be solved by using a Human-In-the-Loop-Optimization (HILO) and finding optimal assistive patterns for each individual [9]. However, HILO is a time-consuming and difficult process to optimize the accurate parameters for each individual. Furthermore, it is not guaranteed that optimal parameters for one condition (such as a certain speed) will be optimal for all conditions (other speeds). EMG-based control methods also suffer from issues due to non-stationarity and subject-specificity [21]. In some cases, muscles are so close to each other that complicates the measurement of the surface EMG signal [21].

Unlike model-free methods, which use human locomotion control output signals (either joint torque/position or EMG signals) to control the assistive device, the core of the model-based approaches is to understand the underlying control principles of human locomotion. Recently, biologically inspired reflex-based controllers were utilized for gait assistance [59, 66]. Such control approaches could potentially improve robustness and adaptability to new conditions [22]. However, dynamical systems (e.g. neuromuscular models [23]) that can describe human locomotion are complicated. The complexity of bioinspired model-based control of assistive devices can be reduced by model abstraction using the template & anchor concept [24, 25], and by employing bioinspired morphological designs such as biarticular actuators [26, 27]. In short, we propose using human locomotion as a template for the design and control of assistive devices (here, exosuit) instead of copying specific movement patterns (as in model-free methods). The similarity between the mechanical design and control in the human body and the robotic locomotor system can improve human-robot-interaction, and result in efficient and adaptive gait assistance.

From control point of view, biomechanical studies on normal and pathological gait demonstrated that humans use proprioceptive feedback signals (e.g. detecting load as a complex parameter, recorded by very different types of receptors) in locomotion control [31, 32]. In addition to the ability of measuring the load (e.g. body weight) in humans, a high dependency of compensatory leg muscle activation to body weight was demonstrated experimentally [33]. In [34, 35], ground reaction force (GRF) has been used to control a rigid exoskeleton for supporting antigravity muscles on the lower extremities. In [28], the force modulated hip compliance (FMCH) model was introduced as a template for posture control in which the GRF was used to tune the hip joint compliance. This method was later implemented in the LOPES II exoskeleton to assist human walking [29, 30]. The results revealed the advantages of using force feedback in reducing the human metabolic cost of walking.

From mechanical design viewpoint, studies on the biomechanics of human gaits and robotics show that biarticular muscles help generate motions in a more energy-efficient way [26]. In fact, these muscles actuate two joints simultaneously, transferring the energy towards distal joints, and further control the output force direction [26, 27, 36]. Given these features, biarticular assistive devices can be more effective in reducing energy consumption during walking. Several examples for successful applications of the biarticular elements in designing assistive devices can be found in the literature [8, 29, 37]. Different aspects of analyzing biarticular muscles from template modeling and biological locomotor systems to the legged robot and assistive devices are reviewed in [27]. Moreover, adding compliance in a serial elastic actuation mechanism provides further advantages compared to the common method of using direct-drive motors for exosuit [38, 39, 63]. Some of these advantages are (not limited to) recouping energy, increasing robustness (e.g. at impacts), reducing peak power, and required torques of the electric motors. In biarticular compliant actuators, the returned energy of one joint can be stored in the elastic element in part of the gait and returned in another joint in the following gait phase [40]. Due to its zig-zag configuration [41], human leg includes two biarticular thigh muscle groups (rectus femoris and hamstrings), supporting the hip and knee joints, and only one biarticular shank muscle (Gastrocnemius). Previous studies showed that passive biarticular thigh muscles are important for balance control [27, 36, 46] and simple biarticular springs are sufficient to generate human-like swing leg control [36, 47]. In addition, about 63% of the required energy in walking is produced by hip and knee joints [42]. These biomechanical findings motivate further investigations on developing assistive devices using biarticular compliant mechanisms contributing to knee and hip joints.

The main goal of this paper is to develop a novel bioinspired controller for a new exosuit which
provides metabolic advantages, robustness and adaptability to new conditions. In that respect, the main contribution of this study is twofold. Firstly, we introduce a new biarticular exosuit called BATEX (BiArticular Thigh EXosuit) that actuates hip and knee joints with a compliant biarticular actuator. Secondly, we introduce a bioinspired model-based control of the BATEX which is a simplified version of the reflex control [43]. In this GRF-based control, we use an extended version of the FMCH control method to actuate the artificial muscles [28], in the stance phase, and passive biarticular springs for leg swinging. Here, the advantages of using such design and control of an exosuit are shown by analyzing human walking experimental data with an OpenSim model, regarding metabolic consumption and efficiency. We also evaluate the robustness of our control approach through comparing simulations with and without load carriage. The main benefits of the proposed design and control are demonstrated based on these experiment-based modeling outcomes. A comparison between our model-based approach and the optimal model-free control will be presented to evaluate the achievable efficiency and robustness of our method.

2. Materials and methods

This section describes the BATEX design concept and our GRF-based control approach, followed by details about the implementation of our gait assistance method with the BATEX in the OpenSim software. We explain how our model-based controller is designed to approximate the optimal control signals. This is a feedback control method of our exosuit which hypothetically could outperform the robustness and adaptability of the feedforward control (see the supporting results in the next section). The abbreviations and parameter definitions are listed in table 1.

| Abbreviation | Nomenclature                      | Parameter | Definition                        |
|--------------|----------------------------------|-----------|-----------------------------------|
| ABC          | Adaptable biarticular compliance | τ<sub>VPP</sub> | Hip required torque in VPP model |
| BATEX        | Bi-articular thigh exosuit       | f<sub>1</sub> | Leg force                          |
| CMC          | Computed muscle control          | l        | Leg length                         |
| CoM          | Center of mass                    | r<sub>s</sub> | Distance from VPP to hip joint    |
| DD           | Direct drive                      | ψ        | Hip angle                          |
| EMG          | Electromyography                  | r<sub>PPP</sub> | Distance from CoM to VPP         |
| FMC          | Force modulated compliance       | γ        | VPP’s deviation angle from the trunk axis |
| FMCH         | Force modulated compliant hip    | τ<sub>h</sub> | Hip torque in FMCH model         |
| GRF          | Ground reaction force             | c<sub>h</sub> | Normalized hip spring stiffness of FMCH model |
| HAM          | Hamstring                         | f<sub>h</sub> | Force applied by the actuator    |
| RF           | Rectus femoris                    | c        | Normalized stiffness in the FMC   |
| SEA          | Serial elastic actuator           | l<sub>AM</sub> | Length of the artificial muscle |
| SLIP         | Spring loaded inverted pendulum   | k        | Spring stiffness                   |
| TSLIP        | Trunk + SLIP                      | l<sub>0</sub> | Rest length                       |
| VPP          | Virtual pivot point               | t<sub>lock</sub> | Motor’s lock time                |
|              |                                   | T<sub>s</sub> | Operation time interval           |

Table 1. List of symbols, terms and definitions (x denotes one of the FMC, ABC or P for passive).
2.2. Control approach

To describe our state-based feedback control approach, we first define the basics of gait phasing in walking. A full gait cycle begins at the touch down of one foot and continues until the touch down of the same foot in preparation for the next step. The gait cycle has two main phases: the stance and swing, which alternates for each lower limb. The stance phase is characterized by the time the leg moves freely in the air to complete the ground and, the swing phase covers the entire time that the leg moves freely in the air to complete the step. In our control strategy, we separate the stance time that the leg moves freely in the air to complete the ground, and the swing phase covers the entire time that the leg is in contact with the ground, which alternates for each lower limb. The stance phase is defined by the time that one foot is in contact with the ground and the virtual pivot point (VPP) concept was already applied to predict hip torque control BA TEX in the stance phase. To extend this hybrid approach, we first define the basics of gait phasing in walking. A full gait cycle begins at the touch down of one foot and continues until the touch down of the same foot in preparation for the next step. The gait cycle has two main phases: the stance and swing, which alternates for each lower limb. The stance phase is characterized by the time the leg moves freely in the air to complete the ground and, the swing phase covers the entire time that the leg moves freely in the air to complete the step. In our control strategy, we separate the stance and swing phase control using a switching system (see section 2.2.3 for details.). State-based controllers are common methods for control of assistive devices (e.g. [15, 39]). Our control approach in the stance phase stems from the FMCH, a template model-based posture control that predicts human upper body balance in locomotion [28]. As a preliminary step, first, we describe the virtual pivot point (VPP) concept [50] as a backbone of the FMCH model [28]. Then, we explain how to use the FMCH template model to control BATEX in the stance phase. To extend this approach for the segmented leg, in [51], we introduced a simulation model of an exosuit design with a biarticular hamstring (HAM, hip extensor, knee flexor)-like actuators with adjustable stiffness. In this study, we further extend it by (1) adding the second biarticular actuator for simultaneous hip flexion and knee extension (similar to RF muscle), (2) adding a compatible swing leg control, and (3) introducing an optimization-based design method using OpenSim software.

After presenting the stance phase control, the swing phase control will be described. The control scheme for the swing phase is also based on passive compliance, introduced in [47]. Since both of these control strategies use biarticular thigh muscles, we can easily switch between them when the gait phase changes (from stance to swing and vice versa). This hybrid control approach will be called adaptable biarticular compliance (ABC), hereafter.

2.2.1. Stance phase control

To describe our ABC method, we first explain the basic concept and developed control method for the balance locomotor subfunction, which is based on the VPP concept developed by Maus et al [50].

The VPP model: the VPP, which is observed in human and animal gaits, is a point on the upper body above the center of mass (CoM) where the GRFs are intersecting during the stance phase [50]. This concept was already applied to predict hip torque required for posture control using template modeling for different gaits [52, 53]. Template models are simple conceptual models using basic mechanical elements (e.g. mass, spring), which can explain locomotion’s important features [24]. One of the common templates for modeling walking and running is the spring-loaded inverted pendulum (SLIP) model [54]. This model consists of a point mass describing CoM, attached on top of a massless spring, which represents the stance leg. For describing posture control, a rigid trunk can be added to the SLIP model resulting in the TSLIP (Trunk + SLIP) model [52, 53]. Using this model, the required hip torque \( \tau_{\text{VPP}} \) (see figure 1(a)) to redirect the GRF to a determined VPP can be calculated by

\[
\tau_{\text{VPP}} = \frac{f_l l h \sin(\psi) + r_{\text{VPP}} \sin(\psi - \gamma)}{l + h \cos(\psi + \gamma)}
\]

in which, \( f_l \), \( l \), \( r_{\text{VPP}} \) and \( h \) are the leg force, leg length, hip angle, the distance from CoM to VPP and from CoM to hip joint, respectively. The VPP angle is defined by \( \gamma \), the angle between body axis and the vector from CoM to VPP, as shown in figure 1(a). For more details about derivation of VPP formulation please see [50].

The FMCH model: in [28], a new template model was developed to generate VPP using a variable stiffness which is called force modulated compliant hip (FMCH) model. In this model, leg force feedback was used to adjust hip spring in the TSLIP model. It was shown that for the range of joint angles observed in human walking, the FMCH could precisely approximate the predicted hip torques from the VPP model (equation (1)). Furthermore, this model could predict the human hip torque \( \tau_h \) in walking at different speeds [55] with the following simplified equation.

\[
\tau_h = \alpha f_l (\psi - \psi_0).
\]

Here, \( \alpha \) and \( \psi_0 \) are the normalized stiffness (to body weight and leg length) and the rest angle of the adjustable hip spring, respectively (figure 1(b)).

The FMC model, an extension for segmented leg: due to the segmentation of the leg in humans, for using the GRF-based control (e.g. to control the exosuit) the FMCH model was extended to a biarticular level [51]. Since we use the GRF for controlling biarticular muscles and not the hip spring, we will use FMC standing for force modulated compliance and skip the ‘H(hip)’ in FMCH. For this, a virtual leg was defined by a line from the hip to the ankle, and the virtual hip torque was defined between the virtual leg and the upper body. To control this virtual hip torque, both hip and knee joints should be controlled in coordination. In [36], it has been shown that with appropriate design of the thigh biarticular actuators in a bioinspired bipedal robot (BioBiped3), the GRF direction can be controlled with minimum interference with the GRF magnitude. Since the VPP concept is based on GRF direction control, the thigh biarticular muscles (simplified as adaptable springs in figure 1(c)) can be used to mimic human-like balance control in the segmented leg. Thus, the force of biarticular (artificial) muscle, given by this so called FMC
controller, will be

\[ f_{\text{FMC}} = cf_l \max(l_{\text{AM}} - l_{\text{F0}}, 0) \] (3)

in which \( f_{\text{FMC}} \) is the force applied by the biarticular actuators (adaptable springs for RF or HAM AMs), \( f_l \) is the axial leg force, and \( c, l_{\text{AM}} \) and \( l_{\text{F0}} \) are the normalized stiffness, length and rest length of the AM, respectively. For the two biarticular thigh muscles, the tunable control parameters \( c \) and \( l_{\text{F0}} \) will be determined by the optimization method, described in the following section. The max function represents the (muscle-like) unidirectional force generation of the actuators. Based on the same argumentation of [36] the hip to knee lever arm ratio was set to 2 which generates the same effect as the FMCH with the springy leg [51] and equalizes models of figures 1(b) and (c).

2.2.2. Swing phase control
With a simulation study, Dean and Kuo demonstrated that the coupling of the hip and knee joints with biarticular springs could yield efficient swing motion with an appropriate ground clearance similar to humans [48]. They also showed that the knee-to-hip moment arm ratio has a significant effect on the stability of gaits at different speeds. More recently, Sharbafi and his colleagues could predict kinematic and kinetic behavior of swing leg and muscle forces in human walking using a template model consisting of a double pendulum with combinations of biarticular (thigh) springs [47]. These studies’ outcomes are in line with former biomechanical studies, which emphasize the important role of the thigh biarticular muscles in the swing phase of human walking [57]. It was shown that the RF and HAM muscles contribute to the first and second halves of the swing phase, respectively [57]. This evidence supports the ability to provide human-like swing leg motion with passive biarticular springs, which also fits to our FMC control in the stance phase (see figure 1(c)). Therefore, we used fixed springs (with constant \( k \)) for assisting hip and knee joints in the swing phase.

\[ f_p = k \max(l_{\text{AM}} - l_{\text{P0}}, 0). \] (4)

In this equation, \( f_p, k, l_{\text{AM}}, l_{\text{P0}} \) are the force applied by AM, stiffness, the length and the rest length of the AM for the passive mode, respectively.

2.2.3. ABC control of BATEX in walking
The ABC control is composed of adjustable biarticular springs with force modulated compliance for a time interval \( (T_f) \) and fixed springs (with fixed stiffness \( k \)) for another time interval \( (T_p) \). The following equation presents the formulation of ABC control for each leg.

\[ f_{\text{ABC}} = \begin{cases} cf_l \max(l_{\text{AM}} - l_{\text{F0}}, 0), & t \in T_f \\ k \max(l_{\text{AM}} - l_{\text{P0}}, 0), & t \in T_p \end{cases} \] (5)

Here, \( l_{\text{AM}} \) is the length of the AM including the spring length and motor displacement. The reset length of AM is set differently for FMC (\( l_{\text{F0}} \)) and passive mode (\( l_{\text{P0}} \)).

For the HAM-artificial muscle, the time interval \( T_p \) starts with the takeoff (beginning of the swing phase, shown by \( t_{\text{TO}} \)) and ends when the force generated by passive spring equals the force calculated by the FMC control. This moment of force equality is denoted by \( t_{\text{FE}} \) and can be found when the following equality condition is held.

\[ l_{\text{AM}} = \frac{cf_l l_{\text{F0}} - k l_{\text{P0}}}{cf_l - k}. \] (6)
Then, $T_f$ starts from $t_{FE}$ to the next takeoff of the same leg. This switching rule is considered to generate continuous change between the two phases of the ABC control.

For the RF-artificial muscle, $T_f$ starts by touch down (beginning of stance phase, denoted by $t_{TD}$) and ends at a certain time $t_{lock}$ by locking the motor before the beginning of the swing phase. Consequently, $T_f$ is defined from this moment to the next touch down. Determination of $t_{lock}$ is described in section 3.1.

Roughly speaking, the contribution of the HAM-artificial muscle starts by passive spring (when the motor is locked) in the late swing and continues by switching to FMC in the stance phase, while the RF-artificial muscle starts with FMC in the stance and continues by switching to passive spring (by locking the motor at $t_{lock}$) before swing phase starts. Our investigations showed that setting $t_{lock}$ to a moment that the peak of FMC appears ($t_{PF}$) will result in a more efficient and effective control. The influences of the spring stiffness and motors’ locking time on the serial spring rest length and the effective force are discussed in detail in description in section 3.1.

Inspired by the positive force feedback in biological muscles [58], we consider about 40 ms delay ($t_d = 0.040s$) in the GRF signals. Interestingly, this biologically motivated delay results in a better matching between the FMC and the optimal force (optimal force defined in the next section). In fact, this delay exists in the exosuit implementation due to GRF measurement and the settling time in the low-level force control. All in all, the active control time $T_f$ for HAM- and RF-artificial muscles is set as follows.

$$T_f = \begin{cases} [t_{FE}, t_{TO} + t_d], & \text{HAM} \\ [t_{TD} + t_d, t_{PF} + t_d], & \text{RF} \end{cases}$$ (7)

The passive time period $T_p$ in which the motor is locked will be the rest of the gait cycle. In equations (5)–(7) the GRF and AM forces are used to detect the switching time and required forces. This means the ABC control is time-independent.

2.3. OpenSim model development for testing BATEX

For evaluating the applicability, performance of the proposed exo design and control, we employed OpenSim software (https://opensim.stanford.edu), an open-source software for developing experiment-based models of musculoskeletal structures and creating dynamic simulations of movement [61]. Due to difficulties in (human) experiment based validation of assistive devices’ design and control, using neuromuscular simulation models is becoming more popular [4, 22, 59]. By reviewing several studies with
this approach, Grabke et al. stated that the musculoskeletal modeling-based approaches could be considered as alternatives to intuition-based assistive device design [60].

We performed three sets of simulations in OpenSim to test the performance, efficiency and adaptability of BATEX and the implemented ABC control method. First, using an open data-set in OpenSim, we added the BATEX to the human walking model and searched for the optimal actuation to minimize the approximated metabolic cost of walking at the subjects’ self-selected speeds. Such a data-driven optimal force pattern is not generated based on a control principle but considered a predefined reference trajectory for designing the controller. In the second simulation, we applied the ABC control method to generate forces close to the optimal ones using a model-based approach. Finally, we compared the optimal solution (from the first simulation) and the ABC controller with the same parameter found in the second simulation on a new data set with extra load (for the same subjects). This last simulation is used to evaluate the model-based and model-free methods’ robustness against system changes (extra load).

In the following, we first describe human walking data used in the OpenSim model and then discuss the optimization process and the ABC implementation on the BATEX.

2.3.1. Modeling assisted walking in OpenSim

The locomotion task in this study is overground walking at self selected speed for two different conditions: normal walking (at 1.46 ± 0.15 m s⁻¹; referred to as the no-load condition) and loaded condition in which the subjects carry 38 kg in a backpack (modeled on the torso) at 1.27 ± 0.09 m s⁻¹. Our data set includes 7 healthy subjects (age 25 ± 5 years, height 1.86 ± 0.04). This data-set is borrowed from [4] which is available at https://simtk.org/home/assistloadwalk. The experimental data includes 3 overground trials for each subject and each condition, in which the kinematic data, GRF and moments were measured. Details about experiments can be found in [4].

We used OpenSim software (version 3.3) for simulations [61]. A three-dimensional musculoskeletal model with 39 degrees of freedom and 80 hill-type muscles has been used for our simulations. Eight degrees of freedom (bilateral ankle eversion, toe flexion, wrist flexion, and wrist deviation), which are not necessary for our analyses, have been locked. The model was based on 21 cadavers and 24 young, healthy humans [62]. Our simulation workflow was based on [4]. At first, the musculoskeletal model was scaled to match the anthropometry of each subject. Then, joint angle trajectories were calculated using OpenSim’s IK (inverse kinematics) tool. After that, OpenSim’s RRA (residual reduction algorithm) tool was used to reduce the residual forces (applied to the pelvis) resulting from a small discrepancy between force plate data, marker data, and the musculoskeletal model. Finally, by using OpenSim’s CMC (computed muscle control) tool, muscle driven simulations of each trial have been generated.

2.3.2. Optimal control of BATEX

We utilized the CMC tool for designing and predicting the effects of ideal assistive devices. The two biarticular (HAM and RF) actuators, shown in figure 1(d), were implemented in OpenSim using path actuator (figure 3(a)). These elements were modeled as additional reserve actuators and defined to be unidirectional (developing just pulling force), and we added it in bilateral configuration (to both legs). The devices are implemented in OpenSim such that the exosuit only applies force in the sagittal plane. This is how the developed BATEX hardware setup works as well (figure 3(c)). For simplicity, the introduced exosuit was massless with ideally perfect force tracking without power or force limitations.

In order to minimize the muscle energy consumption at each time instant, we can use the instantaneous muscle activation signals. As the generated motion (and consequently, muscle length changing rate) is fixed and determined by the human experimental data, muscle activation could be an appropriate signal to approximate and minimize the consumed energy. Therefore, the CMC’s cost function, $J$, at time instance $t$ is defined based on two terms: body effort, and simulation (modeling and measurement) error. Considering $M$ as the set of lower limb muscles, the effort term is calculated by summation of square of individual muscle instantaneous activation $a_i(t)$ for $i \in M$. This OpenSim model considers a set $R$ of so called reserve and residual actuators to generate forces/moments $f_j(t)$ for $j \in R$ which compensate for measurement and modeling errors. In the assisted mode, the CMC’s cost function $J$ includes the force $f_j(t)$ applied by the exosuit actuators in addition to the previous terms. Defining $w_{ij}$ as the constants for weighting the compensating actuator forces, $f_j(t)$ and $E$ as the exosuit actuator set, the updated CMC’s cost function will be

$$J(t) = \sum_{i \in M} a_i^2(t) + \sum_{j \in R} \left( \frac{f_j(t)}{w_{ij}} \right)^2 + \sum_{k \in E} \left( \frac{f_k(t)}{w_{jk}} \right)^2.$$

Here, $w_{ik}$ weight is set to a large value (1000 N) for lowering the penalty of using actuators instead of muscles. We applied the same kinematic and kinetic (GRF) dataset in both assisted (with BATEX) and unassisted models. Thus, the net joint moment in assisted and unassisted modes is conserved, and the provided supportive effort in the exosuit could potentially yield lower muscle instantaneous activation. This could result in metabolic cost reduction. We employed the cost function $J$ to find the optimal force at each time instant. The resulting control patterns will provide maximum reduction in metabolic energy consumption.
cost with the BATEX while maintaining the kinetic and kinematic behavior. This means that theoretically, applying these patterns to the BATEX will help the individual subjects to consume the minimum metabolic energy.

2.3.3. ABC control in OpenSim

Following the predefined optimal patterns, described in the previous section, does not have any fundamental difference to the trajectory-based methods (see the introduction section). Instead, we use this optimal signal as a reference to tune our bioinspired ABC controller. This way, a GRF-based feedback control will be applied to the BATEX, which could potentially outperform the robustness and adaptability (e.g. perturbation recovery or adapting to changed gait conditions) of the feedforward (trajectory-based) control.

In a former section, we explained the ABC control method, which can prescribe the appropriate force in biarticular thigh AMs in both stance and swing phases. The stiffness and rest length of the fixed and adaptable springs in equation (5) are identified by approximating the optimal reference trajectories with the ABC control, using MATLAB curve fitting toolbox. We also found that the peak of the FMC is a perfect moment to be selected as the locking time \( \text{lock} \) for the RF-artificial muscle. This way, the dependency on time is removed by using the GRF, which was already measured and used in the FMC control (see results section for details).

For implementing the ABC controller in OpenSim, first, we calculated force profiles with equation (5) using kinematic and GRF feedback. Then, the assistive pattern was implemented in the CMC tool as a constraint for the BATEX actuators. In this way, the actuator tracks the commanded force profile, calculated by the ABC method in MATLAB. The main constraining assumption of these simulation studies is that subjects walked with the same kinematics and GRFs in both assisted and unassisted conditions. Studies on soft exosuit reported relatively small changes in kinematics and kinetics with assistance [63]. Hence, assuming fixed kinematic and kinetic behavior is acceptable for unimpaired gaits.

2.4. Metabolic cost calculation

The metabolic cost was computed for both unassisted and assisted simulations as a total effort of all muscles using the Umberger2010MuscleMetabolicsProbe in OpenSim 3.3 [4, 64]. This cost does not include a basal metabolic rate. We compared assisted simulation with unassisted in each trial to calculate the metabolic cost reduction.

Because of differences in subjects’ preferred speed and stride length, the data on one of the force plates was lost in few trials (16 no-load trials and six loaded trials). Hence, the force data was not sufficient for modeling a complete gait cycle in all trials. As a result, we analyze the experimental data for two single support and one double support phase. Thus, we average the instantaneous whole-body metabolic rate over half a gait cycle (one step) supposed to the approximate mediolateral symmetry of walking.

In addition to the metabolic cost, we calculate the efficiency index of the BATEX to show the advantage of biarticular structure with compliant actuation and our hybrid control. This index is introduced in [69] as a ratio between the reduced metabolic power in human body and the consumed power in the exosuit. For each trial, we calculate the net metabolic power saving of the assistive device, by subtracting the metabolic power of assisted walking from that of unassisted walking. Then, we divide the average of the net metabolic power saving by the average
Figure 4. RF and HAM-artificial muscle force and length. (a) and (c) The HAM- and RF-artificial muscle optimal force pattern for the left and right leg respectively, and its approximation by the ABC controller (using SEA) in the OpenSim walking model. We showed the left leg for HAM-artificial muscle and right leg for RF-artificial muscle to have a continuous force profile for demonstration purposes. (b) and (d) Displacement of the motor ($l_M$) and the length of the serial spring ($l_S$) and artificial muscle ($l_{AM}$). Here, the $l_{FMC}$ is the prescribed length of the serial spring to apply FMC force, which is not followed after $t_{lock}$. $T_p$ and $T_f$ are respectively, passive mode and FMC operation time intervals, used in ABC control method. The shaded region shows the swing phase of the gait cycle. The gait cycle starts with the touch down of the right leg.

$$\text{Efficiency index} = \frac{\text{Net metabolic power saved by exosuit}}{\text{Average exosuit positive mechanical power}}. \quad (9)$$

To investigate the effect of the spring in the actuator design, we also compared DD actuation and SEA (the comparison results are reported in the next section).

3. Results

This section describes the simulation results of gait assistance with our bioinspired designed and controlled exosuit. To demonstrate how the ABC controller can be tuned to predict control outputs, which are sufficiently close to the optimal actuation in the biarticular AMs, we first present the results of force patterns in these two (model-based and model-free) approaches. We describe the parameterization of the ABC control (e.g. finding $t_{lock}$, stiffness adjustment, and motor movement) to approach the optimal trajectories of the BATEX in the experiment-based OpenSim model. Then, the quality of this control method is evaluated based on the metabolic cost reduction, exosuit power consumption and efficiency index in comparison with the optimal solution. Finally, we compare the quality of the two controllers in the loaded condition to investigate their adaptability to a new condition.

3.1. Optimization based ABC control in OpenSim

The ABC control parameters are obtained as described in section 2.3.3. In this section, we separate the presentation of the results of this procedure for the HAM- and RF-artificial muscles. In figure 4, the optimal force for the RF and HAM-artificial muscles is shown by gray color. In the late swing phase, the optimal HAM-artificial muscles start to contribute (figure 4(a)). A passive biarticular spring can provide this contribution. The optimal stiffness in the simulation is selected according to the optimization force in the $T_p$ time interval. We found that 15–20 kN m$^{-1}$ is an appropriate stiffness range for HAM-artificial muscle approximate the optimal solution for different subjects. In the stance phase, the second peak of the optimal force can be perfectly approximated by the FMC (the blue curve).
result is consistent for all subjects. A combination of passive biarticular spring in the swing phase and the FMC in the ABC framework (the red curve) nicely predicts the optimal pattern. This result is valid for most of the subjects. In the proposed ABC control for HAM-artificial muscles, the motor is locked during the swing phase (interval $T_p$), and the AM behaves like a passive spring in the late swing, as shown in figure 4(b). From the beginning of the stance phase, the FMC predicts increasing force, which is opposite to the generated force by the passive spring. As soon as these two forces are intersecting, the ABC switches to FMC, which means stretching the spring by moving the motor (see figure 4(b)).

For the RF-artificial muscles, we used a SEA for force generation. Like human RF muscles, the optimal force starts at mid-stance (about 35% of the gait cycle, as shown in figure 4(c)). By tuning the appropriate rest length and normalized stiffness, the FMC can appropriately predict the optimal force in the stance phase, including the first peak (see figure 4(c)). The second peak in the optimal pattern is not consistent with the biological evidence of humans RF muscle actuation in walking [57]. This means that the RF-artificial muscles can reduce the activation of other muscles. We tried to find the best fit for the optimal RF-artificial muscle force with minimal effort. The blue curve in figure 4(c) shows that with only FMC, the AM force will be limited to the stance phase. In ABC control architecture, locking the motor (equal to switching off the motor for non-backderivable motors), turns the FMC into a passive spring which can still contribute in the swing phase. This way, part of the optimal force can be provided without energy consumption (investigating the locking time $t_{lock}$ is described in the following).

The movement of the motor and the spring for RF actuator in the prescribed ABC control (red curve in figure 4(c)) is depicted in figure 4(d). The motor only moves from mid-stance to $t_{lock}$ resulting in an increase in the spring stored energy. Returning the energy could support the forward movement of the upper body and swing leg in the late stance and early swing phases, respectively. The blue line in this graph illustrates the spring length if the FMC was implementing completely (without locking in between). The difference between the spring length ($l_s$) and the RF-artificial muscle length ($l_{AM}$) is constant after $t_{lock}$.

In order to better understand the ability of the proposed ABC control within the SEA arrangement, we investigated the effect of the serial spring stiffness and motor’s locking time $t_{lock}$ in figure 5. The $t_{lock}$ influences the contribution of the motor (consumed energy) and the rest length of the spring, which also affects the duration and the magnitude of the RF-artificial muscle contribution in the swing phase. If we lock the motor shortly after the peak of FMC, tracking optimal pattern with the ABC controller is improved (shown in figure 5(a)). However, the motor needs more energy and power as it works longer and needs to stop and return in the opposite direction immediately. Also, as the passive spring maximum lengthening (accordingly its peak force) occurs after the peak of the FMC (see figure 4(d)), with such a timing the generated pattern will be disruptive (decreasing and abruptly increasing). Locking before reaching the peak force resolves this issue. Further, this timing results in decreasing motor energy and power, while
lowering the quality of tracking the optimal pattern by diminishing the first peak.

Our simulations show that locking the motor simultaneous to reaching the FMC peak force is a perfect compromise between human metabolic rate reduction and motor power consumption. Such timing does not require a sudden change in motor movement direction and provides part of the RF-actuator contribution in the swing phase. Further, it supports the time-independent control described in section 2.3.3.

Increasing stiffness of the serial spring enlarges the magnitude of the RF-artificial muscle force (consequently the peak value), as shown in figure 5(b). With larger stiffness, we need less displacement of the motor to create the same force with the FMC, which means less lengthening in the serial spring. This results in a shorter contribution of the passive spring in the swing phase. Therefore, by decreasing stiffness, the stored energy and lengthening of the spring (as well as motor’s displacement) are increased, which results in applying force for a longer period in the swing phase. Figure 5(b) shows that 2500 N m$^{-1}$ is an appropriate spring stiffness, which can approximate the optimal solution the best. In this figure, the moment of reaching the FMC peak is selected as $t_{lock}$.

3.2. Metabolic cost reduction

Similar to the optimal controller, the BATEX with the ABC control can also reduce the required energy for walking at the preferred speed. The metabolic cost reduction and the exosuit energy consumption in the two simulation cases of the optimization-based (model-free) and the ABC control (model-based) methods are shown in figures 6 and 7. In figure 6 the results for the optimal solution and the ABC method are shown by red and green colors, respectively. The transparent colors represent the results of the loaded condition. The comparison between the percentage of the metabolic cost reduction in simulations for different subjects in figure 6 shows that our control method can effectively reduce the metabolic cost of walking. Although the ABC control reduction in the no-load condition is less than the optimal approach (which was expected), it can provide a bioinspired model-based controller instead of a model-free method of giving a time-based trajectory obtained by optimizing for a specific data set. Figure 6 also shows that for the changed condition (carrying an extra load), the ABC method performance works better than prescribing the optimal solution (previously identified for no-load condition) in reducing the metabolic cost for most of the subjects (5 subjects). Unlike the no-load condition, the mean metabolic cost reduction obtained by the ABC method (10.4%) in the loaded condition is not less than what is achieved using the optimal solution (10%). It is remarkable that following the optimal trajectory of the no-load case in the loaded condition yields about 4% drop in the metabolic reduction while the ABC provides 1% higher metabolic reduction compared to its performance in the no-load condition. These results
Figure 7. Metabolic energy reduction, device energy consumption and efficiency index of different controller. (a) Metabolic energy reduction compared to unassisted mode and device energy consumption for the ABC and the optimal controllers for different subjects (showed by different marker). The mean and the standard deviation of three experimental trials for each subject are shown by the bar height and the error-bar, respectively. (b) The efficiency index (given by equation (9)) of the ABC and the optimal controllers for the BATEX in OpenSim. Note that, both optimal DD and optimal SEA are in the actuation level of applying the same optimized force patterns on the exosuit. Circle and square markers show the mean values for each subject and the mean values for each method, respectively. The squares in the shaded area show the mean value for single joint assistive devices implemented in [4].

indicate the benefits of the ABC method regarding adaptability to new conditions and robustness against system changes. The statistical significance was approved using a paired t-test between assisted and unassisted for each subject in both loaded and no-load conditions (p < 0.001).

To provide a basis for comparing the efficiency of different methods, we illustrate the reduced metabolic energy (in Joule, not in percent) in the human subject’s body versus the consumed energy in the exosuit (positive mechanical work) in figure 7(a) for the no-load condition. As the ABC control also benefits from the compliant design of the actuators (SEA), we also used a compliant design for the optimal solution to reduce the consumed energy in the BATEX. Note that both optimal DD and optimal SEA are in the actuation level of applying the same optimized force patterns on the exosuit, which will not affect metabolic cost reduction, shown in figure 6. As demonstrated in figure 7(a), the consumed energy of the device for applying optimal pattern by adding the serial compliance is reduced by 7% in comparison with DD actuation. The comparison between ABC control and the optimal solution with SEA shows that the ABC is, on average, more efficient. To facilitate realizing this comparison, the efficiency index (defined in equation (9)) is depicted in figure 7(b). This graph shows that not only the average efficiency of different subjects but also the efficiency index for most subjects is higher for ABC compared to (SEA-equipped) optimal case. We also demonstrated the efficiency of optimized exoskeletons with single joint assistance from [4] using the same subjects’ data. Figure 7(b) shows that ABC is more efficient than all the other devices that assist only one joint.

4. Discussion

In this study, we introduced BATEX as a new exosuit with innovative contributions in different levels of design and control: (1) underlying control concept: we developed a human-inspired control method instead of the common trajectory-based control of the assistive devices [65]. (2) Actuation level: implementing compliant biarticular actuation with SEA arrangement is proposed in BATEX design. A combination of biarticularity and compliance, as two bioinspired design features of legged systems, are rarely applied to assistive devices [27]. (3) Feedback signal: using the GRF as sensory feedback for control in locomotion, which was inspired by biomechanical studies [33, 44] is applied for the first time in a soft wearable exosuit.

We investigated the proposed BATEX design and the ABC control idea by human experiment-based simulations in OpenSim software. The results confirmed the advantages of BATEX design and also the proposed model-based control regarding metabolic reduction, efficiency, and adaptation to an unknown condition which will be discussed further in the following. We also verified the applicability of the design and control method in interaction with the human by developing the hardware setup and conducting a pilot experiment (not presented here).
In order to facilitate identifying an optimal design and control for a new assistive device with respect to improving metabolic cost reduction in human locomotion, neuromuscular simulation models are advantageous. In simulation-based analyses of gait assistance, we assume that the kinematic and kinetic behavior stays the same after wearing the exoskeleton, as demonstrated in [4]. We accept the discrepancy between experimental and simulations, especially in predicting humans’ reactions to external forces (e.g. from the exosuit). Nevertheless, such neuromuscular experiment-based models (e.g. OpenSim models) are the best-known software tools for proof of concept before experimental investigations. Using kinetic (GRF) and kinematic (AM length) feedback for control could potentially increase the adaptability to different changes in motion dynamics, environment and also to recover from perturbations. Moreover, by using such simulation models, one can examine the effects of assistive devices on forces and metabolic consumption of individual muscles, which is extremely difficult by experiments [4]. In the following, we discuss the simulation results regarding improvement in the assistance level (metabolic cost reduction), efficiency, and robustness of the proposed method in comparison with an optimal trajectory tracking method in light of the three above-mentioned contributions.

(1) Underlying control concept: our FMCH-based control [28, 51] of the BATEX has a significant difference to the state-of-the-art methods of controlling assistive devices. Instead of replicating the human locomotion control system’s outputs, e.g. joint torques, here we try to discover the underlying control principles for synthesizing locomotor functions. Recently, reflex-based control (introduced by Geyer and Herr [43]) was utilized for gait assistance [59, 66]. Neuromuscular models [43] are useful biologically inspired tools for developing model-based control of assistive devices [22, 66] which could potentially address the adaptability of the controller for different conditions. Nevertheless, the main barrier for using such models to control assistive devices is their complexity and a large number of parameters to be tuned [66]. As the neuromuscular control in human locomotion is too complex and is not well understood [23], we employed a simplified model prescribed by the template & anchor concept [24]. Such kind of template-based control is recently employed for bioinspired legged locomotion control [67] and also assistive devices [15]. In that respect, we selected the FMCH model [28], which was developed to explain human balance control based on the VPP concept [50]. By the use of biarticular thigh actuators with appropriate lever arms, an extension of the FMCH model for a segmented leg was presented [51]. This way, we translated a biomechanical template model of human balance control to a practical anchor level as a core concept for design and control of the so-called BATEX. In the here presented approach, we combined this model with the passive biarticular spring model for human-like swing leg control (the DPS model in [47]). Therefore, the first and maybe the most important novelty of this study is introducing a new (template-based) approach for the design and control of exosuits (for both stance and swing phase). If the fundamental control principles of human walking are correctly identified, we expect to have more harmonized interaction between the assistive device and the human body. As a result, we expect to have improved robustness against perturbations and uncertainties, which was shown in the results with the extra load. Interestingly, the ABC controller could outperform the optimal trajectory tracking method in the loaded condition. Moreover, the assistance level of the ABC increases from 9.4% in no-load condition to 10.4% in the loaded condition (more than 10% of the assistance level in no-load condition), while the metabolic cost reduction of the model-free approach drops from 14.2% to 10% (about 30% of the assistance level in no-load condition). This considerable improvement (from −30% to +10%) in adaptability to the changed condition is achieved thanks to the bioinspired model-based control technique. Further investigations on the experimental setup are required to verify this level of adaptability in the real world.

(2) Actuation level: the force (GRF) modulated compliance model, which is based on the VPP concept, was already applied for control of LOPES-II exoskeleton in our previous studies [29, 30]. In addition to the lack of swing phase control in the LOPES experiments, for implementing the FMC (GRF-based control) in such a rigid exoskeleton, the two single-joint actuators were used to emulate the biarticular actuation. Therefore, part of the advantages of the proposed method was not met. Developing an exosuit with compliant biarticular actuators is our solution to regain these benefits. With biarticularity, transferring energy from one joint to another enables the system to reduce energy consumption [27]. Moreover, adding compliance in a serial elastic actuation mechanism provides further advantages such as recoiling energy, increasing robustness (e.g. at impacts), reducing peak power, and requiring torques of the electric motors. OpenSim simulations demonstrated the ability of the proposed mechanism to reduce metabolic costs 14.2% in average with the optimized solutions. In these simulations, we assumed that the kinematic and kinetic behavior of human subjects did not change. Definitely, the adaptation of human subjects to the exosuit’s force could improve the control quality and may result in higher metabolic cost reduction.

To evaluate the efficiency of the BATEX with the ABC controller, we compared the outcomes of our simulations with those found in [4] for single joint assistive devices. Using different approaches on the same OpenSim model and data set provides a fair
comparison that supports the advantages of the proposed design and control method. The total energy of the human body and the assistive device can be decreased to less than the metabolic cost in the unassisted gait because of: (1) storing and returning energy with compliant elements [69, 70] and (2) transferring energy from one joint to another with biarticular mechanisms [49].

Another remarkable finding of this study is the importance of compliant actuation and the matching controller. Comparing the DD and SEA already shows about 10% improvement in efficiency index, which is not surprising. Applying the ABC control and combining the active (FMC) and passive (spring) modes provides a more pronounced improvement in efficiency index (more than 30%) while using the same compliant actuator (SEA). This means benefiting from the compliance in the actuation system with an appropriate (e.g., bioinspired) controller might be more critical than actuator design.

Recently, we developed the BATEX hardware (shown in figure 3(c)) and tested its performance in a pilot experiment using ABC control (Supplementary data [https://stacks.iop.org/BB/16/066024/mmedia]). Our experimental results (not presented) show that the GRF is a helpful feedback signal for control, as shown in the OpenSim simulations. Moreover, the generated force patterns in the biarticular thigh actuators are similar to their biological counterparts (RF and HAM muscles). The preliminary results support the control concept, and the effects on the metabolic cost need to be tested in a comprehensive experiment with sufficient subjects in the future.

(3) Feedback signal: the last novelty of this study is introducing the GRF as a useful feedback signal. Basically, our control approach is similar to impedance control, while the GRF signal adjusts the actuator impedance (stiffness). Biological evidence supports implementation of GRF for locomotion control [32, 44]. Studies on locomotor disorders confirm the importance of GRF in locomotion control, e.g., when there is a reduced load sensitivity and, respectively, decreased leg extensor activation in Parkinsonian gaits [31]. Inspired by these studies, positive force feedback in muscle control [54, 58], and the FMCH template model are all in line with the idea of using GRF as a feedback signal for locomotion control. Although control theory says that positive feedback destabilizes controlled systems, the positive force feedback in human motor control [58] explains how reinforcing the extensor activity during the loading period of the stance phase will contribute to load compensation without leading to instability [44]. It is noteworthy to mention that the constraints (limited muscle force) and hybrid dynamics of the gait (switching between stance and swing phases) are of reasons to explain how positive feedback could stabilize locomotion.

Contrary to phase-detection-based methods (for gait assistance), the GRF-based control indirectly synchronizes the exosuit control with human movement. This in-phase human–robot interaction could provide advantages in energy efficiency. A comparison between the efficiency of the ABC and the optimal solution supports this claim. Although the ABC results in less metabolic cost reduction in the no-load condition, it is more efficient than the optimal solution even if it is equipped with serial compliance. The developed device could generate efficiency above 100%, which was also observed in previous studies [9, 69]. Applying the GRF as an informative sensory signal for control, removes the need to realize gait phasing. In equations (5)–(7) we removed the dependency to time by using the GRF and the AM properties (e.g., length and force). External perturbations will also be reflected in the GRF patterns, and in the case of designing an appropriate controller, this sensory information could potentially robustify the system against perturbations. Increasing robustness was clearly observed in our analyses of loaded conditions.

Although the GRF is a very useful feedback signal, employing it in an exosuit is not straightforward. In our experimental setup, we utilized the GRF signal measured by the instrumented treadmill, which is applicable for training and rehabilitation tasks. For measuring GRF in a portable exosuit for overground walking, the common technology is using the insole shoe sensors, which suffer from wear, imprecision, and low sensitivity [71]. New technologies are introduced to measure the GRF using magnetic fields [72, 73], or high-sensitivity capacitive sensors [71], which are more robust than the traditional insole force sensor and are more appropriate for our application.

5. Conclusion

This paper addresses the mechanical design and control of assistive devices by introducing the BATEX as a novel exosuit. Due to the tight interaction between humans and robots in gait assistance applications, synthesizing biological locomotion can help develop more effective devices. Human musculoskeletal systems and neural control as two biological solutions led us to develop our neuromechanical framework, which is comprised of compliant biarticular actuation and the ABC control as an abstract version of reflex control. Analyzing human walking assistance using an experiment-based OpenSim model demonstrates the advantages of the proposed design and
control of BATEX regarding metabolic cost reduction, adaptability, and efficiency. The here presented design and control methodology for benefiting from morphological computation using bioinspired actuation system (e.g. compliant biarticular actuators) and neuromechanical control (e.g. ABC) can be further analyzed and implemented in other types assistive devices, e.g. exoskeletons and prostheses. The outcomes of this study are presented for healthy subjects while their applications for rehabilitation and assisting impaired subjects could potentially be investigated in the future.

Author’s contributions

Vahid Firouzi is the first author who performed the OpenSim simulations and analyses. He contributed in writing the manuscript and preparing the figures. Ayoob Davoodi was responsible for manufacturing the exosuit and contributed in writing the manuscript. Fariba Bahrami was a co-advisor of the project and contributed by her constructive discussions within the project and giving feedback on writing the paper. Maziar A Sharbafi is the supervisor of this research and the corresponding author who significantly contributed to writing the article. He was responsible for the conception of the research and design of the exosuit.

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Conflict of interests

The authors declare that they have no competing interests.

Data availability statement

The data that support the findings of this study are openly available at the following URL/DOI: https://simtk.org/projects/assistloadwalk.

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