Varying mechanical compliance benefits energy efficiency of a knee joint actuator

Bacek, Tomislav; Moltedo, Marta; Geeroms, Joost; Vanderborght, Bram; Rodriguez Guerrero, Carlos David; Lefeber, Dirk

Published in:
Mechatronics

DOI:
https://doi.org/10.1016/j.mechatronics.2019.102318

Publication date:
2020

Document Version:
Accepted author manuscript

Link to publication

Citation for published version (APA):
Varying Mechanical Compliance Benefits Energy Efficiency of the Knee Joint Actuator

Tomislav Bacek*, Marta Moltedo, Carlos Rodriguez-Guerrero, Joost Geeroms, Bram Vanderborght and Dirk Lefeber

Robotics and Multibody Mechanics Research Group (R&M), Vrije Universiteit Brussel and Flanders Make, Pleinlaan 2, 1050 Brussels, Belgium

Abstract

In the field of wearable robots, actuator efficiency and user safety are frequently addressed by intentionally adding compliance to the actuation unit. However, the implications compliance has on the actuator’s overall performance in different conditions and activities are not fully understood, largely due to single task-focused experimental evaluations of these devices. To overcome this, our paper analyzes the effects that changing mechanical compliance has on the actuator’s overall performance in different ideal conditions in an experimental test setup. The torque performance and electrical energy consumption of an orthotic, adjustable-compliance knee joint actuator are evaluated during emulated walking and sit-to-stand-to-sit movements. Furthermore, the feasibility of joint operation of a dual mechanical compliance configuration during walking is investigated, and its outcomes reported in this work. The results demonstrate that varying mechanical compliance can lead up to 50% energy savings compared to a no-compliance configuration and show that, in general, changing compliance level leads to either energy-optimal or power-optimal actuator performance, but not both.

Keywords: Compliance, Series elasticity, Parallel elasticity, Variable stiffness, Energy efficiency, Torque controller

1. Introduction

In the context of physical human-robot interaction (pHRI), robot efficiency and user safety are often closely coupled as they both depend on the existence of an energy buffer between the user and robot. Such a buffer benefits both by decoupling their inertias and allowing a dampening effect in reacting to perturbations induced by any of the two. A simple yet effective way to create this buffer is to intentionally add compliance in the transmission of a robotic actuator, an idea pioneered by Pratt’s work on Series Elastic Actuation (SEA) [1]. Due to the advantages SEA-based actuators have over their stiff counterparts for use in the field of pHRI, the idea has been widely exploited ever since in different actuator realizations [2, 3, 4, 5, 6] and successfully applied in exoskeletons [7, 8], rehabilitation robots [9, 10], and knee joint orthoses [11, 12, 13]. However, all SEA-based actuators suffer two major drawbacks. On one hand, their energy storage capacity and attainable output dynamics depend on the fixed spring constant that is often the result of a compromise, severely limiting their applicability and versatility [14]. On the other hand, they cannot reduce actuator’s torque requirements [15, 16], requiring big motors to be used in series with springs.

A natural solution to the first major SEA drawback was the introduction of Variable Stiffness Actuation [17], a principle that allows changing the output stiffness to adapt to environmental changes and task requirements. This is important in pHRI not to constrain humans’ compliance-based versatility and adaptability [18, 19], and to intentionally alter both dynamics of a human-robot system and that of a human itself [20, 21]. However, the ability of VSA-based actuation units to materialize characteristics found in biological systems often comes at the cost of higher complexity, weight, and lack of robustness. Consequently, only a few designs have been applied to power wearable robots [22, 23, 24], despite many proposed designs [25, 26, 27, 28, 29]. Nevertheless, the performance of VSA-based actuators is often reported in particular scenarios/conditions, leaving the implications of their compliance-based adaptability in test benches and real application-scenarios largely unexplored and inadequately understood.

A common approach to overcome the second major SEA drawback is to place a compliant element parallel to the actuator’s joint, resulting in a Parallel Elasticity Actuation (PEA) unit. However, PEA’s demonstrated potential to improve actuators energy performance [30, 31, 32] comes with three significant drawbacks: fixed stiffness, the need for a spring (dis)engaging mechanism, and the inability to decouple motor-load inertia when used with an active device. The solutions to the latter two drawbacks have been proposed in both prosthetic and orthotic knee designs.

*Corresponding author
Email address: tbacek@vub.be (Tomislav Bacek*)
URL: +32 2 629 18 26 (Tomislav Bacek*)

Preprint submitted to Mechatronics August 26, 2019
Most PEA-based actuators use a spring (dis)engagement mechanism and a quasi-passive design \cite{33, 34, 35, 36} that exploits mainly elastic, spring-like behavior of the knee joint. Similarly, some prosthetic knee joint designs complement PEA-based units with SEA-based actuators \cite{35, 37}, an approach recently seen in bipedal robots as well \cite{32}. However, to the best of our knowledge, no available prosthetic/orthotic knee joint device combines VSA- and PEA-based units to fully exploit the natural dynamics of the task while delivering biologically-relevant knee joint torque-angle output. Furthermore, no actuator has been designed that overcomes the PEA’s fixed stiffness by using an active compliant unit working jointly with the PEA-based unit.

The main goal of this paper is to extend our understanding of the benefits that using adjustable compliance actuators alone, as well as in concerted operation with passive compliance, has on the overall performance of the orthotic power units. By combining two independent yet complementary actuation units built around series (MAC-CEPA) and parallel (lockable spring) elasticity concepts \cite{38}, and conducting extensive experimental validation of their performance in different application scenarios and under different conditions, this paper makes two main contributions:

- it provides an extensive analysis of the effects adjustable mechanical compliance has on the energy performance and torque output of the VSA-based knee joint actuator during walking at different cadences and STS movement (combined sit-to-stand and stand-to-sit), and
- it explores the feasibility of compensating fixed, sub-optimal PEA-based unit’s stiffness by superimposing its torque output using an actively controlled VSA-based unit installed in series to the actuated joint.

The remainder of the paper is organized as follows. Section II provides background information necessary to understand the outcomes of the experimental validation presented herein. It includes biomechanical data used in tests, mechatronic design and control of the knee joint actuator, benchtop setup, and theory behind the actuator’s simulated performance. The results of an experimental actuator characterization are reported in Section III, together with their simulated counterpart. Section IV and V present discussion and concluding remarks.

2. Materials and methods

2.1. Knee joint biomechanics in Activities of Daily Life (ADLs)

The actuator’s performance is evaluated using the available biomechanical data of human walking (source: Winter dataset \cite{39}) and STS activities (source: data provided by Afschrift \cite{40}). As arguably the most exercised daily activity, walking was used as a biomechanical source in designing the actuator, which has multiple advantages \cite{41} but also decreases actuator’s versatility. The dataset of the STS activities was used since they make one of the basic requirements of everyday mobility \cite{42, 43}, while significantly differing from walking given their lack of energy-buffer phases that could exploit compliance in the system. Both datasets were used in the form of a torque-angle characteristic (Fig. 1), a common approach in the literature given its usefulness in defining biologically-driven actuation design requirements.

![Walking torque-angle characteristics](image1)

![STS torque-angle characteristics](image2)

Figure 1: Biomechanical data used in the design and validation of the actuator. (left) Normal and slow cadence walking data. The original dataset was stretched in time to account for the actuator’s speed limitation - slow cadence from 1.4 sec/gait to 2.8 sec/gait, and normal cadence from 1.14 sec/gait to 2.3 sec/gait. The dashed line represents the early stance phase when the knee joint exhibits a spring-like behavior \cite{44} exploited in the design of the actuator. Black and green straight lines represent the quasi-stiffness of the early stance phase during normal and slow cadence walking, respectively. Quasi-stiffness is a slope of a linear approximation to the joint’s torque angle profile \cite{45} that defines the joint’s resistance to external forces. (right) Biomechanical data of STS activities. Although qualitatively similar, the two activities pose very different requirements on the actuator. The data were obtained using a chair seat at the knee height, and are stretched to 1.75 sec/activity, up from 1.27 sec for standing up and 1.53 sec for sitting down.
2.2. The adjustable dual-compliance knee joint actuator

The actuator used in this study is a torque controllable knee joint actuator whose detailed mechanical design, low-level control, and preliminary evaluation have been previously published in [38]. Below, only the parts of its mechatronic design and low-level torque controller relevant for understanding the results of this work are presented.

2.2.1. Mechatronic design

The actuator consists of both series and parallel elasticity actuation units combined into a single modular compliant actuator. The series elasticity actuation unit is based on the MACCEPA concept [25], realized using a spindle drive directly connected to the Levearm (Fig. 2 (left)). The actuator is developed guided by an anthropomorphic design approach and can deliver torque only in the sagittal plane. To avoid adding complexity, retain compactness, and allow combining series and parallel elasticity units the actuator’s joint is realized as a simple hinge, a viable solution for activities such as walking [34, 46, 47]. In the current realization, the MACCEPA’s spring pre-compression and thereby the actuator’s angular stiffness can only be changed offline, using a simple worm-gear mechanism.

The parallel elasticity actuation unit, called the weight-acceptance (WA) mechanism, is based on the concept of passive energy exchange that uses a locking element to deliver the torque. In the case of the knee joint, the unit exploits changes in gravitational potential energy during the early stance gait phase. The WA mechanism is designed around a non-back drivable spindle drive that can provide the reactionary torque by friction locking the spring. For the reasons of compactness, the mechanism is placed at the backside of the actuator (Fig. 2 (right)).

![Mechanical realization of the knee joint actuator.](image)

Figure 2: Mechanical realization of the knee joint actuator. The range of motion is 5° in the extension and 95° in flexion, limited by mechanical stops. (left) Series elasticity is realized as a spindle-driven MACCEPA due to its simplicity, compactness, and favorable output characteristic. The output torque in Eq. (1) is generated by compressing the MACCEPA spring through a controlled motion of the Levearm. The MACCEPA cannot be separated from the actuator’s structure, unlike the WA mechanism that can be simply inserted. (right) Parallel elasticity is realized as a quasi-passive mechanism consisting of a motor, non-back drivable spindle, and a spring. The output torque in Eq. (2) is generated by friction-locking the spring along the trapezoidal spindle.

The actuator’s output torque $\tau_{act}$ is a sum of the torque $\tau_{macc}$ provided by the MACCEPA and $\tau_{wa}$ provided by the WA mechanism, with the two defined as follows:

$$\tau_{macc} = \kappa_S \cdot B \cdot C \cdot \sin(\alpha) \cdot \left(1 + \frac{P - |C - B|}{A}\right),$$  \hspace{1cm} (1)

$$\tau_{wa} = \begin{cases} \kappa_P \cdot \Delta x \cdot y \cdot \sin(\beta + \delta), & \text{early stance phase} \\ 0, & \text{otherwise} \end{cases}$$  \hspace{1cm} (2)

where $A$, $B$, and $C$ are the MACCEPA’s geometric parameters, $\kappa_S$ and $P$ its spring stiffness and pre-compression, respectively, and $\alpha$ its spring deflection angle. On the other hand, $\kappa_P$ and $\Delta x = L_{ts} - L_{ts,0}$ denote WA mechanism’s spring stiffness and compression, respectively, while angles $\beta$ and $\delta$ and length $y$ are auxiliary parameters coming from the mechanism’s structure.

The actuator’s two units can work jointly in at least two different modes. Assuming that a high-level controller provides a reference torque online and that a gait phase-detection system is available, the MACCEPA can either be in a zero-torque or an assistive mode during the early stance gait phase. In the former case, it is assumed that the WA mechanism alone can match the biological torque-angle output of this phase. In the latter case, the MACCEPA is superimposing the sub-optimal WA mechanism’s torque-angle output, thereby correcting the actuator’s output and steering it towards the desired one. This is possible due to the actuator’s accurate torque estimation.
2.2.2. Control strategy

The MACCEPA model given in Eq. (1) describes its output torque as a function of geometry, spring characteristics, and deflection angle $\alpha$. However, when interacting with humans, it is more interesting to define MACCEPA’s output in terms of angular stiffness. In that case, the actuator’s output can be described as follows:

$$\tau_{act} = \hat{k}_{joint}(\alpha) \cdot f(\alpha),$$

where $\tau_{act} = \tau_{mac}$ is the torque generated at the actuator’s output, $\hat{k}_{joint}(\alpha)$ is the estimated angular joint stiffness, and $f(\alpha)$ a non-linear function of the deflection angle $\alpha$. To use the MACCEPA as a torque source while applying a position control, the above relation between the torque $\tau_{act}$ and deflection angle $\alpha$ needs to be inverted. This can be done in different ways, including numerical calculations and using neural networks. In this work, a closed-form solution was obtained based on experimental data as follows (details in [35]):

$$\alpha_{ref} = \frac{1}{\mu(P)} \log_e \left( \frac{2 \cdot \hat{k}_{joint} \cdot G(P)}{\tau_{act} + \hat{k}_{joint} \cdot G(P)} - 1 \right).$$

For any spring pre-compression $P$ preset offline and torque reference $\tau_{act}$ defined online by a high-level control, the MACCEPA’s middle-level controller estimates angular joint stiffness $\hat{k}_{joint}$ and calculates the reference deflection angle $\alpha_{ref}$ in real time. In other words, the MACCEPA’s output torque is controlled in an open loop by controlling the deflection angle $\alpha$ (through the Levearm control) in a closed loop. Parameters $\mu(p)$ and $G(P)$ are the slope and amplitude of a tanh function whose inversion led to the Eq. 4. The parameters are obtained by fitting a third-order polynomial through $(G_i, P_i)$ and $(\mu_i, P_i)$ pairs, where $i = \{20, 30, 40, 50, 60\}$% spring pre-compression.

The quasi-passive nature of the WA mechanism means that it only needs to move the nut up the spindle to its engaged position, where the nut is friction-locked while delivering torque, and then down the spindle to the nut’s disengaged position (the lowest position on the spindle). As a result, the WA mechanism’s middle-level control is based on a simple position control. The nut can take any position along the spindle due to its infinite engagement angle resolution, while its actual position depends on the user’s walking pattern. A relation between the motor’s position and position of the nut along the spindle is calculated using the geometry of the mechanism. Fig. 3 gives a schematic of the actuator’s low-level controller consisting of the MACCEPA and WA mechanism part.

![Diagram of the cascaded middle-level actuator torque controller](image)

Figure 3: The cascaded middle-level actuator torque controller. The part on the right of the $\tau_{act}$ is the MACCEPA’s controller, and the part on the left the WA mechanism’s controller. MACCEPA: The open-loop torque control is realized by controlling the $\alpha$, whose reference is calculated using the Eq. 3. The position control consists of a feedback branch, a feedforward branch, and a damping injection on the measured motor velocity $\dot{\theta}_m$. The term $G(s)$ in the feedforward branch represents the mapping from the Levearm position $\theta_{LA}$ to motor’s velocity $\dot{\theta}_m$ based on the actuator’s geometry. WA mechanism: The actual motor position $\theta_{wa}^{act}$ is measured and translated into a nut position $Z_{wa}$ along the spindle using geometric relation $G_{wa}(s)$ between the two variables. The nut position error is then fed to a motor velocity $P$ controller. $\hat{\tau}_{wa}$ is the estimated torque calculated using the WA mechanism’s theoretical model and is subtracted from the total joint’s torque when two units operate together to calculate the MACCEPA’s reference torque.

2.2.3. Gain scheduling

Binary nature of the WA mechanism, whose spring (dis)engagement takes place almost instantaneously, introduces discontinuities in the control loop which are reflected in the MACCEPA’s oscillatory behavior at the moment of (dis)engagement. To dampen the oscillations, gain scheduling is applied to the MACCEPA’s controller gains.

---

3 One of the main benefits of using compliant actuators in the field of wearable robots is their intrinsic ability to turn a force/torque control problem into a position control problem.

4 Angular joint stiffness is estimated based on the torque reference and spring pre-compression $P$, as shown in Fig. 4. The polynomial $\hat{k}_{joint} = f(\alpha_{ref}, P)$ is based on the experimental evaluation of the actuator.

5 Parameters $G(P)$ and $\mu(P)$ are defined as follows: $G(P) = -1.35e - 5 \cdot P^3 + 0.0012 \cdot P^2 + 0.0055 \cdot P + 4.67$ and $\mu(P) = 6.42e - 7 \cdot P^3 - 6.99e - 5 \cdot P^2 + 0.0012 \cdot P + 0.33$. 
The scheduling works by predicting the moment of the WA mechanism’s spring engagement angle based on the measured data and decreasing the gains just before the engagement. All the gains are increased back to their nominal values just before the WA mechanism is disengaged again. The scheduling is realized as follows:

\[
\begin{align*}
\hat{K}_P &= \max(K_{P}^{\text{nom}} - 10 \cdot K_{\text{gain}} \cdot (\theta_{BIO}^{\text{mea}} - \theta_{\text{eng}}), K_{P}^{\text{min}}), \\
\hat{K}_F &= \max(K_{F}^{\text{nom}} - 0.0012 \cdot K_{\text{gain}} \cdot (\theta_{BIO}^{\text{mea}} - \theta_{\text{eng}}), K_{F}^{\text{min}}), \\
\hat{K}_D &= \begin{cases} 0.01, & \text{when WA mechanism engaged} \\ 0.1, & \text{otherwise} \end{cases}
\end{align*}
\]  

(5)

where \(\hat{K}_D\) denotes damping gain, while \(\hat{K}_P\) and \(\hat{K}_F\) denote modified feedback and feedforward gains, respectively, defined to allow smooth transitions between their nominal (\(K_{P}^{\text{nom}} = 2500\) and \(K_{F}^{\text{nom}} = 0.85\)) and minimum (\(K_{P}^{\text{min}} = 1500\) and \(K_{F}^{\text{min}} = 0.4\)) values. The parameter \(K_{\text{gain}} = 100\) is experimentally determined to deliver fast and smooth transition without causing any jerks, and \(\theta_{\text{eng}}\) is WA spring’s predicted (dis)engagement angle.

2.3. Experimental test setup

To test the actuator in different biologically-relevant scenarios, a test setup was built that consists of a robust cage, three actuator connections points, flexible couplings with a torque sensor, and a load motor (Fig. 4). Fixing the actuator to three points allows testing its peak performance and structural stability while operating at its software and hardware limits. The load motor enables controlled movement of the actuator’s output link. By controlling this link in position, it is possible to test to what extent can the actuator reproduce the biological joint torque of walking and STS when the joint angle is imposed. However, direct control of the load motor’s output position means that the load dynamics are masked from the actuator. This is arguably a simplification of the real-life scenario, but such an approach reduces the dimensionality of the actuator’s performance analysis and allows focusing solely on the actuator and its independent parameters. The entire setup is controlled in real-time in the Beckhoff TwinCAT master environment, while the model of the entire benchmarking setup was created using Matlab Simulink® environment. More information on setting-up this environment can be found in [49].

Figure 4: Experimental test setup. The actuator’s output torque is measured using a torque sensor fixed in between two flexible couplings to avoid any misalignment-related damage. The load motor (not depicted) is screwed to the top plate of the aluminum cage and has the upper coupling fixed to its output axis. The actuator is fixed to the cage at the distal part of its two links and at the level of its main rotational axis, which is aligned with the axes of the torque sensor, flexible couplings, and the load motor.

2.4. Data processing

All collected signals were measured at a 1 kHz sampling rate and filtered using a zero-phase low-pass Butterworth filter, both first- and second-order, to remove noise artifacts. The cut-off frequency was determined based on the output of the Fast Fourier Transform (FFT) analysis, with a minimum of 6 Hz [39]. The filtered data was segmented into walking, standing up, and sitting down cycles, averaged over 50 cycles, and reported only using mean value since its standard deviation is either very small or adding it to the graphs would make them hard to read.

The evaluation of the actuator’s performance includes analysis of its output impedance, output torque, and electrical motor power and energy requirements. The motor power of each actuation unit is calculated as a product of the measured motor current \(I_m\) and an estimated motor voltage \(U_m\), \(P_m = I_m U_m\), with \(U_m = L \frac{dI_m}{dt} + R I_m + k_v \dot{\theta}_m\) (\(L, R, \) and \(k_v\) are motor’s terminal inductance and resistance, and speed constant, all respectively) [50]. Subsequently, the actuator’s electrical energy consumption \(E_{\text{act}}\) is calculated as the integral of the summed electrical motor powers
$E_m$ of both actuation units, $E_{act} = \int \sum (P_m^{acc} + P_m^{wa})\, dt$. Electrical motor power and energy and not their output mechanical counterparts were used following the suggestions outlined in [51].

2.5. Theoretical analysis

Some of the results presented hereafter give a comparison of experimental and simulation data in terms of electrical motor energy, power, and two types of motor losses, damping and resistive. To allow this analysis, an elaborate model of the actuator’s dynamics was developed using guidelines presented in [54]. Electrical motor power is calculated as $P_m = I_m \cdot U_m$ using the following equations for the motor current $I_m$ and voltage $U_m$:

$$I_m = \frac{1}{k_t} (\tau_m + v_m \cdot \dot{\theta}_m), \quad U_m = L \frac{dI_m}{dt} + R I_m + k_V \dot{\theta}_m,$$

where $\tau_m$ and $v_m$ are motor’s torque and viscous damping coefficient, respectively. The motor torque is a sum of the inertial ($\tau_m^{int} = J_m \cdot \ddot{\theta}_m$) and load components, where the latter is calculated using spindle-based geometry of the actuator and motor torque equations given by the manufacturer, as follows:

$$r_m^{load} = \frac{F_s \cdot \rho}{2 \cdot \pi \cdot i \cdot C}, \quad C = \begin{cases} 1/\eta, & \text{if power is generated,} \\ \eta, & \text{else,} \end{cases}$$

where $F_s$ and $\rho$ denote spindle’s feed-force and pitch, respectively, while $i$ denotes gearhead’s transmission ratio. The gearhead’s actual efficiency $C$ is calculated depending on the power-flow through the system and using the maximum efficiency $\eta$ provided by the manufacturer. The viscous damping coefficient is approximated based on the motor’s no-load current $I_{nl}$ and speed $\omega_{nl}$, and torque constant $k_t$ as $v_m = \frac{k_t}{\omega_{nl}} I_{nl}$. Damping losses depend only on the motor’s speed and are calculated as $P_{elec,damp} = v_m \cdot \dot{\theta}_m^2$, while resistive losses come from the motor current flowing through its windings and are calculated as $P_{elec,res} = R \cdot I_m^2$.

3. Results

During experiments, the actuator was used in either the MACCEPA-only configuration where only a series elasticity unit was employed, or in the MACCEPA-WA configuration where both elasticity units were operational. To see how the MACCEPA unit compares to a stiff actuator, all experiments were also conducted in the stiff configuration where no compliance was present in the system. To avoid possible effects of the Dyneema rope, the stiff configuration was realized by locking the Leverarm to the actuator’s output link effectively removing the spring altogether, rather than by pre-compressing the spring to 100%.

3.1. Impedance experiments

The ability to change output impedance is one of the main features of adjustable compliance actuators, beneficial for both the user and actuator. Fig. 2 shows how changing the actuator’s compliance level affects the resistance, in terms of reactive torque, that the load motor or a user needs to overcome to back drive the actuator. The tests were done in the MACCEPA-only configuration by moving the actuator’s output link in a controlled manner using a custom-designed multi-sine signal. As the figure shows, increasing the level of mechanical compliance reduces the actuator’s resistance to the movement due to the increased energy-storing capacity of the spring, which absorbs output dynamics. This is even more evident when a torque control is switched ON, whereby the most compliant configuration reduces resistive torque RMS five times compared to the stiff configuration.

3.2. Walking experiments

Walking experiments were carried out in both MACCEPA-only and MACCEPA-WA configurations, although the actuator was not energy-optimized for the former. The reason for including this sub-optimal MACCEPA configuration nonetheless was to investigate how significant are the effects of fixed compliance on the energy consumption of the actuator working in a dynamically changing-compliance environment such as the human knee joint.

3.2.1. MACCEPA-only configuration

Fig. 3 shows the actuator’s performance in the MACCEPA-only configuration while delivering biological torque-angle output in the knee joint during slow and normal cadence walking. In both cases, an increase in compliance level leads to the improved energy efficiency of the actuator, albeit only to a certain level (40% spring pre-compression). For lower levels of compliance, the actuator’s energy efficiency is again getting worse due to increased electrical motor losses, as seen in Fig. 3 (top row). High oscillations in electrical motor power at the onset of torque build-up (0-10% of the gait cycle) are not only a consequence of the high level of compliance which increases the Leverarm’s and consequently the motor’s velocity but also of the high feedforward and feedback gains of the MACCEPA torque controller. At the same time, higher compliance levels lead to lower resistive losses in 15-20% of the gait cycle,
Figure 5: Actuator’s output impedance, characterized as a resistive torque measured at the output controlled in position (20-sec multi-sine signal, 7 degrees peak-to-peak amplitude, 5 Hz frequency spectrum). The torque is a consequence of the reflected inertia, damping, and friction in the actuator. Note different y-axis scale. RMS torque values are given for the entire length (20 sec) of the signal.

when the spring’s energy is used to deliver the high torque of this phase. In terms of the output torque, the actuator’s performance was not significantly affected by changes in compliance level. The small differences in the early swing phase during both walking speeds come from the lower torque resolutions when the actuator is close to its zero-torque zone, also the zone of the lower torque accuracy. The only visibly different performance comes from the stiff actuator, which underperforms in both the early stance and late swing phase, albeit delivering the torque of the mid-swing more accurately.

Figure 6: Actuator’s performance in delivering the biological torque-angle reference during slow cadence (80 kg person, 33 Nm peak) and normal cadence (80 kg person, 50 Nm peak) walking. Similar trends hold for other user weights and are thereby not shown here. All data is normalized to body mass. Energy values refer to motor’s average electrical energy per gait cycle (lower is better). Biological joint energy per gait cycle is also given for comparison. Power peaks for all compliance settings at the very end of the cycle are a consequence of the dataset interpolation. Both power graphs are limited in y-axis for the sake of clarity of the presentation. The highest power peaks were measured with 20% pre-compression at both walking speed, reaching 2.06 W/kg (more than 5x the second highest peak) for slow and 2.8 W/kg (about 2.5x the second highest peak) for normal walking. Due to high torque demands during normal cadence walking, the configuration with 20% pre-compression could not deliver the torque, which was reduced from 80 to 70 Nm peak.
Fig. 7 (top row) shows the effects of compliance on electrical motor losses in experimental setup. Resistive losses dominate the first phase of the gait cycle when high torque needs to be delivered while damping losses dominate the second phase of the gait cycle when high speed is needed. Compliance does not have a significant effect on the damping losses during the swing phase, but it does have during the early stance phase at high compliance levels, coming from high Leverarm speeds. On the other hand, compliance can decrease resistive losses, but only if the system is not too compliant. Too high compliance can lead to an increase in resistive losses even bigger than the one seen in the stiff actuator. Fig. 7 (bottom row) gives a comparison of experimentally-obtained and simulation-based resistive and damping losses. The simulation model significantly underestimates resistive losses in certain parts of the gait cycle because the current used in the model needs to be calculated, unlike experimentally-based one which is measured. Consequently, the simulation does not capture the actuator’s dynamics in full. On the other hand, not much difference can be seen in damping losses across the two environments since these losses depend on the motor speed, a parameter that is controller-imposed in both cases.

Altering the actuator’s compliance affects the energy and power that the actuator needs to deliver the same activity, as shown in Fig. 6. This suggests that there is an energy- and/or power-optimal compliance for different users, activities, and particular requirements of both. That this is indeed the case can be seen in Fig. 8, which gives an overview of the experimental electrical peak power and energy requirements during walking at two cadences and four compliance settings. At lower torque requirements, the lowest compliance is the most energy-efficient but also resulting in the highest power peaks. On the other hand, when higher torques are required a more consistent trend of a clear local minimum can be seen, whereby energy-optimal configuration is always a more compliant one, while power-optimal configuration tends to shift more towards stiff configuration.

A comparison between experimentally-obtained and simulation-based motor power peaks and energy consumption is given in Fig. 9. The simulation-based values are calculated using forward simulation and equations given in subsection 2.5, whereby the experimentally-measured torque-angle output is used as a simulation input to avoid the effects of the MACCEPA controller. In terms of energy consumption, a very similar trend showing a decrease in energy efficiency with an increase in the level of pre-compression can be seen in both datasets. The main difference is consistently higher energy consumption in experimental conditions due to already discussed higher experimental losses and the effects of the controller. A similar vertical offset can be seen in the motor’s peak electrical power data as well. Furthermore, experimental data shows higher magnitudes of increase in the peak power at high compliance levels, likely due to the high controller gains that were necessary for a good performance of the actuator.
Figure 8: Actuator’s peak power and energy requirements during walking at two cadences and four compliance settings, normalized to body mass. Three datasets were used for each cadence, differing in user weight used as a scaling factor. In neither of the six cases does the same compliance level lead to both energy-optimal and power-optimal performance.

Figure 9: The actuator’s peak power and energy requirements during walking - simulation vs experiments. Solid lines depict experimental data and dashed lines simulation data. Experimental data was only recorded and hence given for the pre-compression levels 20%, 40%, 60% and stiff, while simulation data is calculated for 20-60% in increments of 10, 80%, and stiff.

3.2.2. MACCEPA-WA configuration

The MACCEPA can work with the WA mechanism in two modes, a zero-torque and superimposing mode. The implications of the former, which is useful when the WA mechanism’s spring is optimally designed, were demonstrated in our previous work [38]. The latter mode is needed when the WA mechanism’s spring is sub-optimal (Fig. 10 (left)), requiring the MACCEPA to correct the mechanism’s output and steer it towards the desired one.

At the same time, the challenge is to make the two devices work smoothly and in synergy with the user. If the two units are used together without any changes to the MACCEPA torque controller, the MACCEPA will react to the WA mechanism’s (dis)engagement with high-frequency oscillations [51] whose magnitude will depend on the MACCEPA controller gains. This is likely to create a feeling of an impact on the user side and increase electrical power peaks seen on the motor side (Fig. 10 (right)), and should thus be avoided.

The actuator’s performance in the superimposing mode during the early stance gait phase is shown in Fig. 11.
Figure 10: The performance of the actuator in the superimposing mode. (left) When the spring of the WA mechanism is sub-optimal for a given task, either it needs to be engaged at the wrong time to deliver the required peak torque, or at the correct time but then delivering incorrect peak torque. In both cases, there is a (significant) mismatch between the reference and the measured output torque during this phase. (right) To avoid high-frequency oscillations when the two units are working together, gain scheduling is applied to the MACCEPA’s controller gains (see Eq. (5)), which reduces motor power peaks to the same level of the optimal WA mechanism.

By having an actively controlled unit correcting the actuator’s torque output coming from a passive mechanism during this phase, it is possible to deliver a torque more relevant to each person’s needs while avoiding high motor power peaks present when only spring in series is used (see Fig. 9). Compared to using the stiff actuator, concerted operation of the MACCEPA and WA mechanism can not only reduce the actuator’s energy consumption by more than 50% but also its power peaks. Furthermore, the MACCEPA and WA mechanism use less energy than the MACCEPA alone, suggesting that sub-optimal WA spring is beneficial for the actuator’s total energy consumption.

At the same time, the MACCEPA can successfully correct the discrepancy in the desired torque output coming from the WA mechanism, reducing the torque error RMS from about 6.1 Nm in the case of the sub-optimal WA configuration in Fig. 10 (left) to about 2.52 in the superimposing mode. The error could likely be further reduced by using the experimentally-based estimation of the WA mechanism’s torque $\hat{T}_{wa}$ (see Fig. 3), as opposed to a simulation-based estimation used herein. Furthermore, the MACCEPA can also effectively hide the WA mechanism’s fixed stiffness, thus overcoming one of its main design drawbacks and extending its use beyond optimal scenario.

Figure 11: The performance of the MACCEPA and WA mechanism working together to deliver the torque of an 80 kg person (50Nm peak) walking at normal cadence. In the power graph, the shaded area shows the weight acceptance phase, when the WA mechanism is delivering the torque. Two energy values in the brackets represent energy consumption of the MACCEPA and WA mechanism, respectively. The MACCEPA’s pre-compression was set to 40%.
3.3. STS experiments

During walking, the human knee joint mainly dissipates energy and during the early stance gait phase, behaves like a passive torsion spring. During STS activities, the knee both generates and dissipates energy, but no such phase exists where the knee could harvest the energy and then immediately return it. As a consequence, all the experiments are carried out in the MACCEPA-only configuration and are mainly seen as a test for the actuator’s torque tracking and power delivery capabilities since the actuator’s energetic performance during STS experiments is not expected to be as significantly affected by the compliance level as during walking.

The actuator’s performance during the sit-to-stand movement is shown in Fig. 12. Being a work-generating activity, standing up requires high power peaks on the actuator’s motor side to be able to provide the fast-growing torque at the onset of the movement. As a consequence, the motor drains a lot of electrical energy from its source, with a level of compliance having little to no effect on energy consumption. The only noticeable change in energy consumption comes when using no compliance in the system. The 40% pre-compression, which seems to be the most efficient, results in an 18% energy decrease, compared to a 46% decrease in walking with the same pre-compression (Fig. 6). Similar can be seen in the case of the motor power peaks, whereby stiff actuator configuration reaches peak 30-50% higher than those reached during compliant actuation, all of which stay at around 1.4 W/kg. The reason for this can be seen by looking at Fig. 12 and Fig. 13 (top row), the latter giving a comparison of the MACCEPA’s measured velocity and current while delivering standing up torque using different compliance levels. To build a torque using the spring, the MACCEPA’s motor needs to move faster at the onset of a torque-build up, which results in higher electrical power values in 25-40% of the standing up activity. The energy that is stored in the spring can then be used to deliver the peak torque allowing the motor to move slower in this phase and drain less current from the source (40-60%). During the remainder of the activity when high speed is needed, the MACCEPA is beneficial as its spring allows slightly lower motor velocity, but this difference is small enough not to be seen in the electrical motor power. All tested actuator configurations successfully delivered the required torque.

Standing up - actuator electrical power consumption and torque-angle output

![Standing up - actuator electrical power consumption and torque-angle output](image)

Figure 12: The performance of the actuator in delivering the torque of a 60 kg (peak 40 Nm) person during standing up movement. The same trend holds for other conditions as well (40 kg, 75 kg), which are thus not shown here. Energy values refer to the motor’s average electrical energy per activity cycle (lower is better). The biological joint energy of the activity is given for comparison.

Unlike work-generating standing up activity, sitting down dissipates the energy throughout the entire range of motion. As such, an efficient actuation system would be capable of returning the energy to the source during this movement, more efficient being the one returning more energy. As Fig. 14 shows, the presented actuator can successfully generate energy and return it towards the source in both compliant and stiff configurations, though slightly more when compliant. Compliance has a positive effect on power peaks as well, albeit not as big compared to standing up due to, in general, small absolute power values across all compliance settings. Similar to the case of standing up, all tested actuator configurations could successfully deliver the required torque.

This is clearly seen in Fig. 15 showing an overview of peak power and energy requirements during STS activity and different actuator compliance settings. In general, compliance has a positive effect on the actuator’s energy consumption, with 20% and 40% pre-compression settings being the most efficient across all but one condition. Differently from walking scenario, power peaks show the same trend, having the smallest values in a more compliant
configuration. This comes from the nature of the datasets used, STS not having an almost instant torque build-up as is the case with early stance phase of a gait cycle.

4. Discussion

In the context of pHRI, such as in prosthetics and orthotics, intentionally adding compliance to actuation unit is beneficial for multiple reasons, including: (a) safety against abrupt impacts, (b) energy efficiency of the actuator and (c) synergy between the user and environment [20, 21]. In this paper, which focused on the aspect of robotic device efficiency and the effects mechanical compliance has on it across different conditions and activities, the results have shown that adding compliance is highly beneficial, but comes at the cost of performance-related compromises.

Adding mechanical compliance to the active mechanical system between its main motor and output link decouples the inertia of the two, increasing safety and mechanical robustness of the system. By changing the level of compliance, the spring’s energy capacity is altered, which consequently affects the reflected inertia and damping in.
the actuator felt by the user (Fig. 5). This reduces the chances of damaging the actuator’s motor and electronics and protects the users from experiencing adverse effects of the potential impacts. Furthermore, lower output impedance allows softer interaction with the user, which is useful in a user-in-charge assistive mode, for example.

However, adding mechanical compliance to a robotic system affects many of its performance parameters, often requiring a compromise to be made in the level of compliance depending on the required outcome. Positive effects that increasing the level of mechanical compliance has on the interaction safety discussed above may come with a significant increase in peak power demands on the motor during walking, as shown in Fig. 6 (left). The increase, measuring up to seven times the peak in the stiff configuration during normal and up to ten times during slow walking, is a consequence of the reference trajectory that requires very fast torque build-up in a very short period and is further exacerbated by the high controller gains. At the same time, this increase in peak power does not have any effects on the output torque delivery, as Fig. 6 (right) demonstrates.

On the other hand, the actuator’s energy efficiency benefits from an increase in compliance level throughout the entire gait cycle, albeit most notably in delivering the high torque of the early stance gait phase, about 15-25% of the gait (Fig. 8 (left) and Fig. 7 (top)). The benefits of the increase in compliance, however, only come to the point of reaching the optimal compliance when the energy savings can go up as much as 50%. Beyond this point, the actuator’s electrical efficiency starts to decrease again, a result certainly not specific to the knee joint. As the figures show, the reason for a huge increase in the peak motor power with the highest compliance level comes from the very onset of the torque build-up, about 5% into the gait cycle, when low levels of the spring pre-compression require the motor to quickly accelerate and decelerate the leverarm, thus creating high resistive and damping losses. Being always positive, these losses shift the actuator’s energy curve up, canceling the spring’s increased energy capacity and reducing the actuator’s capacity to return more energy to the source.

The effects of compliance on the electrical motor losses are not straightforward and may even seem counter-intuitive. Speed-dependent damping losses during the swing phase are not affected by the compliance despite high velocities the motor needs to deliver (both in the model and experiments, Fig. 7). The reason for this is the joint speed itself, which is dominant over the torque in this phase and requires the motor to follow the swing motion at the same speeds irrespective of the compliance level. On the other hand, the compliance does affect damping losses during the early stance phase, when the joint torque is dominant over the speed. In this phase, adding more compliance results in a noticeable increase in the reference Leverarm velocity in a short time, significantly increasing damping losses. In other words, series compliance, whose benefits usually come from reducing the motor speed and thus its power (ankle joint, for example), can also result in an increased motor velocity and thus increase the actuator’s power. Current-dependent resistive losses, on the other hand, are expectedly dominant only in the early stance gait phase, when the actuator needs to deliver high output torque.

Although not primarily designed for assisting STS activities, the results demonstrate that the actuator can also deliver the torque of these activities without ever over-powering it (Fig. 12 and Fig. 14), and regardless of the level of compliance. Due to the lack of a clear energy-buffer phase in STS activities, the level of mechanical compliance does not significantly alter the energy efficiency of the actuator during these activities, unless no compliance is present.

Figure 15: The actuator’s peak power and energy requirements during STS movement and four compliance settings. For both standing up and sitting down three datasets were used, differing in user weight. In all but two cases the compliance setting resulting in energy-optimal and power-optimal performance differ, with higher compliance generally resulting in a more efficient actuator.
in the system. This is particularly true while sitting down since during standing up, under certain conditions, the optimal compliance can reduce energy consumption by about 15%. Slightly higher power peaks in the stiff configuration come from a higher motor current, which is likely a consequence of both mechanical changes to the actuator that increased friction in the system and energy that did get stored in the spring. In other words, adding mechanical compliance to the actuation unit during STS activities is arguably more interesting from a human-robot interaction and a torque-control point of view than from the actuator’s energy-efficiency point of view. In general, compliance does not change the actuator’s efficiency much compared to the one of a stiff configuration, despite returning more energy towards the power source and having lower power peaks than the stiff configuration.

To further improve the performance and efficiency of the actuator, the WA mechanism has been added to the MACCEPA as a secondary actuation unit that exploits human walking dynamics during the early stance phase of gait by harvesting changes in gravitational potential energy. However, the unit’s passivity and fixed stiffness severely limit its use to the ideal scenario it was designed for (Fig. 10(left)). To overcome this, it is possible to have the MACCEPA work together with the WA mechanism in the so-called superimposing mode, which allows masking the mechanism’s main drawback and extending the range of stiffnesses the actuator can achieve without changing the WA spring (Fig. 11). Furthermore, the results have shown that combining the MACCEPA with the sub-optimal WA mechanism’s configuration can improve the actuator’s energy performance for about 15% compared to the MACCEPA and 55% compared to the stiff configuration. This, together with the possibility to use the WA mechanism as a fall-prevention mechanism during other ADLs by providing a high leg stiffness and extension torque to prevent the leg from collapsing, extends its use well beyond its optimal walking scenario.

There are several limitations to the experimental benchmarking carried out within this work. (i) The biomechanical datasets of level walking and STS activities available in the literature do not distinguish between passive (tendons) and active (muscles) joint torque contributors nor take into account changes in biarticular muscle. As such, these datasets do not represent the actual torque-angle reference that a knee joint actuator needs to deliver when interacting with a user. Moreover, the datasets used in the tests are representative of only a small group of people. (ii) The actuators torque reference was always the full joint torque given in the datasets, while the load motor’s reference was a biological joint angle. This assumption, that the user is capable of moving the leg naturally while being unable to generate any torque around the joint is a simplification of the human-robot interaction and does not fully capture the dynamics the actuator would encounter when assisting the user. (iii) It is known that physical human-robot interaction (pHRI) can have negative effects on the actuator’s performance and needs to be taken into account. The similar also holds for the added inertia to the leg coming from the weight of the orthosis. In this work, pHRI was only indirectly taken into account through the use of a torque sensor and flexible coupling between the actuator and torque sensor. Such a connection can be described using a spring-damper model, which is also used to model interfaces. (iv) The actuator’s performance was analyzed using the motor’s electrical power and energy consumption, which are calculated based on the current measurements and a DC motor model. As such, these parameters are at best a good approximation of the actuator’s actual power and energy consumption, but arguably the closest estimation to the actual consumption we have.

5. Conclusions

The possibility to change its mechanical compliance makes the actuator versatile and allows it to be optimized for both minimum power peak and energy consumption. These two criteria, as shown in this work, generally require different actuator settings not only in walking but also in STS activities, which can also benefit from the compliance in the system. The results have also shown that the optimal compliance depends on the specific requirements of the assisted task rather than being actuator-specific, which is the reason why this work did not try to identify the optimal knee joint actuator configuration despite a mild tendency toward the same compliance level/zone across different conditions. On the other hand, by combining the MACCEPA with the WA mechanism, the whole actuation system showed the potential to improve actuation efficiency while not compromising its output torque performance. This is possible because the WA mechanism exploits the spring-like behavior of the human knee joint during the early stance phase of gait, reducing the main torque requirements from the primary actuation unit.

6. Conflict of interest

The authors declare no conflict of interest.

7. Acknowledgment

The presented work was developed within the projects BioMot (EC’s 7th Framework Program, G.A. No. IFP7-ICT-2013-10-611695) and MIRAD (Flemish agency for Innovation by Science and Technology, IWT-SBO 120057).
References


doi:10.1109/ROBOT.2004.1307425


