A novel adaptive oscillators-based control for a powered multi-joint lower-limb orthosis

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***Abstract*— This paper introduces a novel control strategy for a multi-joint lower-limb exoskeleton during ground-level walking assistive tasks. It is aimed to define a physiologically consistent assistance to the human movements without engaging a complex sensory apparatus, hence providing a simple and comfortable human-robot interface. The control system is a two-block structure: one is based on adaptive oscillators (AOs) dedicated to the hip joint control, the other is a finite-state machine utilized for managing the assistive functions of knee and ankle joints. This control strategy was validated with two subjects walking on a treadmill wearing a hip-knee-ankle-foot (HKAF) exoskeleton. The tests were carried out at different speeds under both zero-torque and assistive control modes. Results presented repetitive and adaptive desired assistive torque profiles during all conditions and both subjects confirmed the benefits of gait assistance. We also analyzed the alterations of kinematics induced by assistive torques: the maximum-changed angle was 8.84 deg at ankle joint, and the time shifts of maximum/minimum angles were always lower than 2% of one stride cycle.**

Keywords*— *adaptive oscillators; state machine; HKAF exoskeleton

I. Introduction

Locomotion disability is a frequent problem troubling people of all ages, especially the seniors: population-based studies in [1] showed a 35% prevalence of gait disorders among persons over age 70, and 60% of 80-84 years old. The movement difficulties could be caused by chronic diseases and impairments from different functional systems [2]. To address the challenges succeeding to gait disorders, in the past decades, various assistive technological devices have been developed; one of the notable outcomes are lower-limb exoskeletons [3], some examples being ATLAS [4], Ekso (Ekso Bionics, US, formerly known as eLEGs [5]), MINDWALKER [6], HAL [7], and soft Exosuits [8].

The core issue of a lower-limb exoskeleton is to provide users with gait assistance as natural as their biomechanical features, while preserving their own residual voluntary movements. To this purpose, a variety of assistive strategies have been conceived, like body model-based control, predefined trajectory control and sensitivity amplification control; their detailed applications by exoskeletons and pros and cons analysis are described in our previous review work [9]. Nevertheless, the development of an efficient assistive strategy to reduce the user's energy efforts without increasing the complexity of the sensory system is still under exploration, and this is particularly true for multi-joint exoskeletons.

In [10], the authors proposed a model-free approach to aid the single-joint (hip) movements in ground-level walking tasks; the controller was grounded on adaptive oscillators (AOs, similar to human motor primitives [11]), and no other sensors were used besides the joints encoders integrated in the exoskeleton hardware. In this paper, we extend this concept to adaptively govern the assistive actions of a multi-joint exoskeleton. The AOs system is still specified to handle the hip joints. Based on the analysis of the biomechanics of ankle and knee joints, a finite-state machine is selected to deal with the assistive action for these two joints. In order to improve the accuracy of gait event detection for state transitions, the walking frequency acquired through the hip controller is input to the state machine. Thanks to this assistive strategy, we aim to simplify the sensory system required by a multi-joint lower-limb exoskeleton and finely tune the assistive actions by setting a few parameters to ensure the consistency with different walking patterns.

The AOs-based control system was preliminarily validated with a hip-knee-ankle-foot (HKAF) exoskeleton which is still under development at The BioRobotics Institute, Scuola Superiore Sant'Anna (Pisa, Italy). An overview of the mechatronics of the HKAF exoskeleton is introduced in Section II. Section III is devoted to explaining the structure of the adaptive controller and its validation results are presented in Section IV. Finally, Section V draws conclusions and prospects future works.

II. Mechatronics of the HKAF exoskeleton

The HKAF exoskeleton we used in this study as test bed for the developed controller is designed for favoring daily-life activities of elderly people or people who are affected by weakened muscles or mild lower-limb impairments. It consists of two modules: a bilateral active pelvis orthosis (APO) and a monolateral knee-ankle-foot orthosis (KAFO), as shown in Fig. 1. The following two subsections will

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separately give an overview on the mechatronics of both modules.

A. APO

The APO is an advanced portable version of the previous laboratory prototype presented in [12]. It is composed by a frame that stabilizes the entire structure on the user's trunk through an orthotic shell (Fig. 1). A backpack fixed onto the frame contains all the electronic components (also for the KAFO control), making the whole device portable and usable in an ecological environment, namely out of the laboratory.

There are three degrees of freedom (DoFs) on each side of APO: adduction/abduction and intra/extra rotation, passively mounted on the frontal and horizontal planes; flexion/extension, actuated by a series elastic actuator (SEA) –a solution widely engaged in the field of wearable robotics [13][14]. In order to minimize the lateral encumbrance of the APO, the actuation units are placed on the back side of the device (the design of this actuation system is under patent pending: Italian patent application n. FI/2015/A/000025). Thanks to this structure, users can move their hip joints within the following ranges of movement (RoMs): [-20, 100] deg, [-15, 20] deg and [-10, 10] deg in the sagittal, frontal and horizontal planes respectively

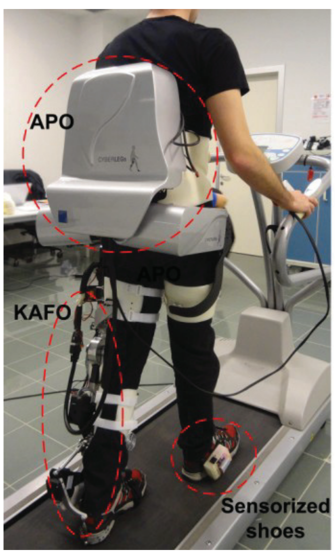

Fig. 1. The HKAF exoskeleton.

B. KAFO

The structure of KAFO is monolateral, limiting its utilization only for the left limb; KAFO is rigidly connected to the APO by a linkage parallel to the thigh, as displayed in Fig. 1. The exoskeletal leg is interfaced to the human thigh and shank through orthotic shells (customized on the user body), and to the foot through an insert placed on the shoe heel and a movable harness on the forefoot.

Active assistance from KAFO is only delivered in the sagittal plane. The flexion/extension of the knee is powered by a one-DoF SEA; the ankle is endowed with an actuator characterized by a non-linear elasticity varying with joint positions. Though only the foot plantar/dorsi flexion is actively actuated, the shoe connection mechanism owns a passive DoF, which allows the user to freely perform inversion/eversion during gait. The RoM of the knee joint is within [-150, 3] deg, and the ankle plantar/dorsi flexion is limited between [-50, 29] deg.

The low-level controllers of both APO and KAFO are based on a PID regulator acting on the error between the commanded torque from a high-level controller (i.e. control strategy) and the measured one on the joint side. Furthermore, a joint position control is realized for the knee joint through a PID regulator. The controller of the whole system is implemented on a sbRIO-9632 board (National Instruments, Austin, Texas, US), endowed with a 400-MHz processor running a NI real-time operating system and a field programmable gate array (FPGA) processor.

III. Assistive structure design

During ground-level walking, the human gait is typically segmented into stance and swing phases, depending on the foot in contact or off the ground [15], as shown in the upper plot of Fig. 2. The lower plot in Fig. 2 provides a concrete example of the torque envelopes of hip, knee and ankle joints within one stride cycle. During swing phase, hip joint torque is a quasi-sine waveform, while the other two articulations act like free damping mechanisms, represented by low torque values. During stance phase, the knee joint is greatly required to be responsible for the body support; then the propulsion, occurring at the late stance, involves a demanding effort from the ankle. Thus despite of the mutual dependences of these three lower joints, the dominant functions are separated in the timing scale.

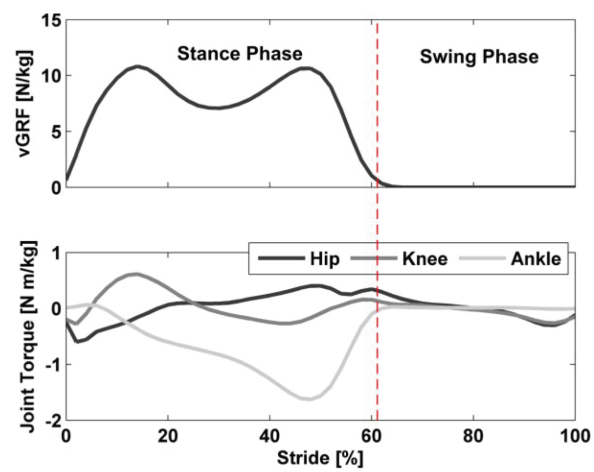

Fig. 2. The vertical ground reaction force (vGRF) and joints torques in one stride cycle during ground-level walking. The red dash line represents toe off moment. The positive torque is defined as to (dorsi-) flex the joint.

Based on the above biomechanical features, an adaptive control structure is proposed to separately manage the assistance of hip joint (APO) and knee-ankle joint (KAFO). Specifically, the APO is controlled by an AOs-based structure, and the KAFO is controlled with a finite-state machine. The walking frequency smoothly estimated by AOs is input to the KAFO state machine. The following two sections respectively provide the design of two assistive blocks.

A. APO control

The AOs utilized for the APO control can extract the fundamental frequency of hip joint angle and provide a real-time estimate of the angle phase, amplitude and offset [10], [16], [17]. For the sake of obtaining a smooth estimation and prediction of the hip joint angle $\theta(t)$, a kernel-based non-linear filter is appended to the AOs system, coupled together by the phase variable $\phi(t)$ learned by AOs. The frame of this AOs-based controller is shown in Fig. 3, where the lower graph gives an example of the real-time estimated and predicted hip joint angles $\hat{\theta}_*(t)$ and $\hat{\theta}_{*,\Delta}(t)$: a shifted phase $\phi(t) + \Delta\,\phi$ is input to the filter to have a real-time prediction of hip joint angle. More details of the working principles of AOs are provided in [10].

The assistive torque for the hip joint is then defined in such a way to attract the joint towards its predicted position by means of a virtual stiffness field:

$$\tau_{des} = K_v \cdot (\hat{\theta}_{*,\Delta}(t) - \hat{\theta}_*(t)) \tag{1}$$

where K_v is a tuneable virtual stiffness. The flexibility is left to the user to constantly adapt his or her gait pattern, and the desired torque profile will adapt with the joint movements automatically.

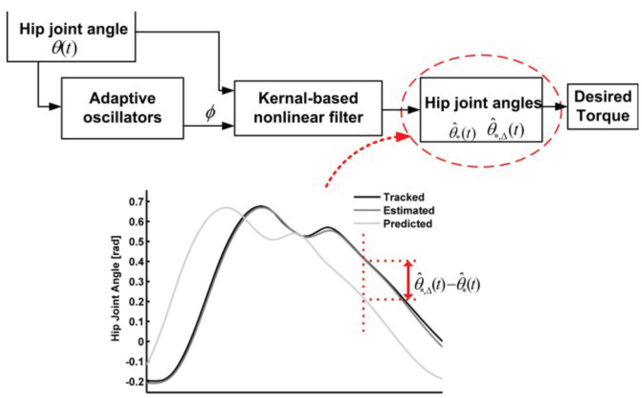

Fig. 3. The AOs-based control structure of APO. The black line in the lower graph is the measured joint angle, while the dark grey and light grey lines are respectively the real-time estimated and predicted angle positions.

B. KAFO control

1) Joint control

Referring to the biomechanical features displayed in Fig. 2, the knee is required to contribute to the weight acceptance, so it is blocked in an extended position before the leg enters into stance phase (black line in Fig. 4). After foot contacting the ground, the ankle joint starts delivering a time-increasing plantar-flexion torque (red arrows in Fig. 4) to bear the weight of the human-robot system together with the blocked knee.

At the end of the stance, for the sake of assisting subjects' forward propulsion, the knee is unblocked (absence of black line in Fig. 4) while the ankle is enabled to deploy a constant push-off torque. To unblock the knee, the low-level control is switched from the position control to the zero-torque control. It is worth noting that the ankle assistive torque starts from the heel strike (HS) of the contralateral leg and ends at the toe off of the orthotic leg. This timing is defined to optimize the benefits of plantar-flexion assistance of an ankle exoskeleton for level-ground walking tasks [18].

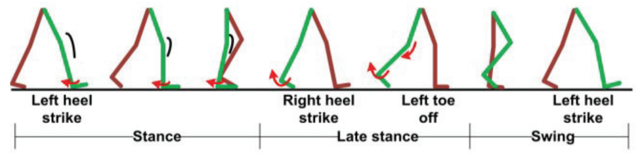

Fig. 4. Diagrammatic sketch of the KAFO joints control within one stride. The red arrows indicate the torques required from knee and ankle joints; the black line on the knee joint during stance phase represents the bloked knee.

During swing phase, both knee and ankle joints are expected to follow the user's movement transparently, thus they are controlled in zero-torque mode (ZTM) (absence of black line and red arrow in Fig. 4). At the beginning of swing phase, for the purpose of lifting the shank and foot and improving foot clearance, a constant flexion torque is commanded to the knee joint, being the duration of the knee flexion torque tuneable.

2) Finite-state machine

KAFO functioning is realized through a finite-state machine-based controller. It is designed to firstly recognize the quiet standing and ground-level walking, then the ground-level walking is further divided into stance, late stance –when the ankle deploys a push off torque– and swing states. The state machine diagram is depicted in Fig. 5.

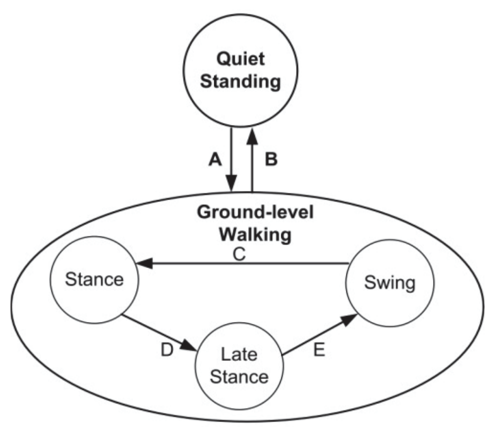

Fig. 5. State diagram of gait pattern recognition. The transitions between activities are labeled with letters A-E and the transition conditions are defined in TABLE I.

The gait pattern recognition depends on two types of biomechanical signals, namely the vertical ground reaction forces (vGRFs) and position of the centre of pressure (CoP), which are recorded by means of a pair of instrumented shoes (Fig. 1). Each shoe contains a sensitive insole represented by a matrix of 64 opto-electronic pressure sensors; a box attached to the shoe contains electronics for signal conditioning and wireless data transmission [19],

[20]. In the stance phase, the CoP is a real finite nsumber, while in the swing phase (being the vGRF equal to 0 N), the CoP is represented by a Not-a-Number (NaN). Consequently, the HS is detected when the CoP signal transits from NaN to a finite value. All recorded signals are sent to the real-time controller of the robotic system through an IEEE 802.15.4 wireless protocol.

Based on the vGRF and CoP, the state transitions the finite-state machine are defined in TABLE I. The transitions between quiet standing and ground-level walking are identified by means of settable thresholds on left and right vGRFs, referring to measured vGRFs during quiet standing. The three states under ground-level walking are distinguished mainly by the different positions of left and right CoPs.

TABLE I. FINITE-STATE MACHINE TRANSITION CONDITIONS FOR THE KAFO CONTROLLER.

Label	Transition Conditions
A	Right vGRF < vGRF Threshold OR Left vGRF < vGRF Threshold
B	Right vGRF > vGRF Threshold AND Left vGRF > vGRF Threshold
C	Left[a] CoP ! = NaN AND $t_{swing} > \alpha * T(t)$
D	Left CoP ! = NaN AND Right CoP ! = NaN AND left CoP < Right CoP
E	Left CoP = = NaN AND $t_{stance} > \beta * T(t)$

[a] Left leg is the one assisted by the KAFO module.

In TABLE I, the conditions:

$$t_{swing} > \alpha \cdot T(t), t_{stance} > \beta \cdot T(t)$$

are exploited to improve the accuracy of detected gait pattern. $T(t)$ is the stride period estimated with the fundamental frequency $\omega(t)$ learned by AOs in the APO controller; t_{swing} and t_{stance} are respectively the duration time of swing and stance states; α and β are coefficients in the interval $(0,1)$. Thanks to these two conditions, when the variables acquired through the instrumented shoes are affected by noise, such as discontinuous CoP at the beginning of stance phase, unexpected wrong state transitions can be avoided.

IV. EXPERIMENTS AND RESULTS

A preliminary validation experiment of the control system was carried out with two young healthy volunteers. They were asked to walk on a treadmill wearing the assistive HKAF exoskeleton. It is hypothesized that, under the control of the developed controller, the HKAF exoskeleton can provide the subjects with adaptive and comfortable assistance without hindering their movements; in addition we also wanted to prove that assistance is smoothly adapted as a consequence of the speed changes within a few strides.

To verify this hypothesis, both subjects were asked to continuously walk on a treadmill with zero slope under both ZTM and assistive mode (AM). In ZTM, the torque reference for each actuated joint of the HKAF system are set to 0 N·m, so the wearer could walk without any resistance from the robot. To test the performance of the controller under more conditions, we designed different walking speeds for the two subjects. Subject #1 was asked to firstly walk under ZTM at 2.4 km/h for 2 minutes, then under AM respectively at 2.4 km/h (2 minutes) and 1.8 km/h (2 minutes), and finally return to ZTM (1 minute) at 1.8 km/h; Subject #2 was asked to go through the same conditions, but with the higher speed being 2.6 km/h and the lower speed being 2.0 km/h. The experimental setup is shown in Fig. 1. During the experiments, angle and torque of hip, knee and ankle joints, as well as vGRFs and CoPs input to the controller, were recorded.

The experimental results are analyzed under both steady-state walking condition and speed transition condition. In the data analysis for steady-state walking, all recorded data were segmented into each stride and interpolated between 0 to 100% of stride cycle (starting form HS of left limb).

A. *Assistive performances during steady-state walking*

The torques and angles from all three joints are displayed in Fig. 6(a, b). Reported results were selected from the data of subject #1. Referring to Fig. 6(a), during ZTM, the residual torques from both the APO and KAFO were near to zero, with the maximum positive and negative torques being: [-0.69, 0.37] N·m and [-0.57, 0.38] N·m from right and left hips, [-0.77, 1.95] N·m from knee, and [-2.50, 2.86] N·m from ankle; the positive torque peaks in the ankle joint graph were due to sometimes the subject hitting the mechanical stop at the late stance.

In AM, the desired-torque parameters were kept to be the same in both higher and lower speeds: K_v was set to 17 N·m/rad, the desired flexion assistive torque from knee was 7 N·m and the constant push off torque from ankle joint was -4 N·m. Nevertheless, thanks to the inherent adaptability of the controller, the timing of assistance was adjusted automatically under different speed conditions, see Fig. 6(a).

According to Fig. 6(a), both the APO and KAFO showed the intended assistive behaviour: maximum hip flexion torque (6.01 N·m) and knee flexion torque (5.1 N·m) at the early swing were delivered to lift the leg; maximum ankle plantar-flexion torque (-3.0 N·m and -3.4 N·m respectively for higher speed and lower speed) at late stance helped to assist the push off; and passive knee torque were generated for weight acceptance at early stance (3.7 N·m for higher speed walking). Since the knee was blocked before HS, there was an extension torque in the late swing. This is a consequence of the fact that when the knee was blocked in the commanded position for weight acceptance, the elastic element in the SEA was compressed and provided the articulation with a stiffness similar to the anatomical one, ensuring a comfortable interaction between the wearer and the robot. Limited by the performance of KAFO actuation units, there are gaps between desired and actual torques of the ankle and knee joints. Nevertheless, both two subjects confirmed that the assistance actually helped them to walk with less perceived effort.

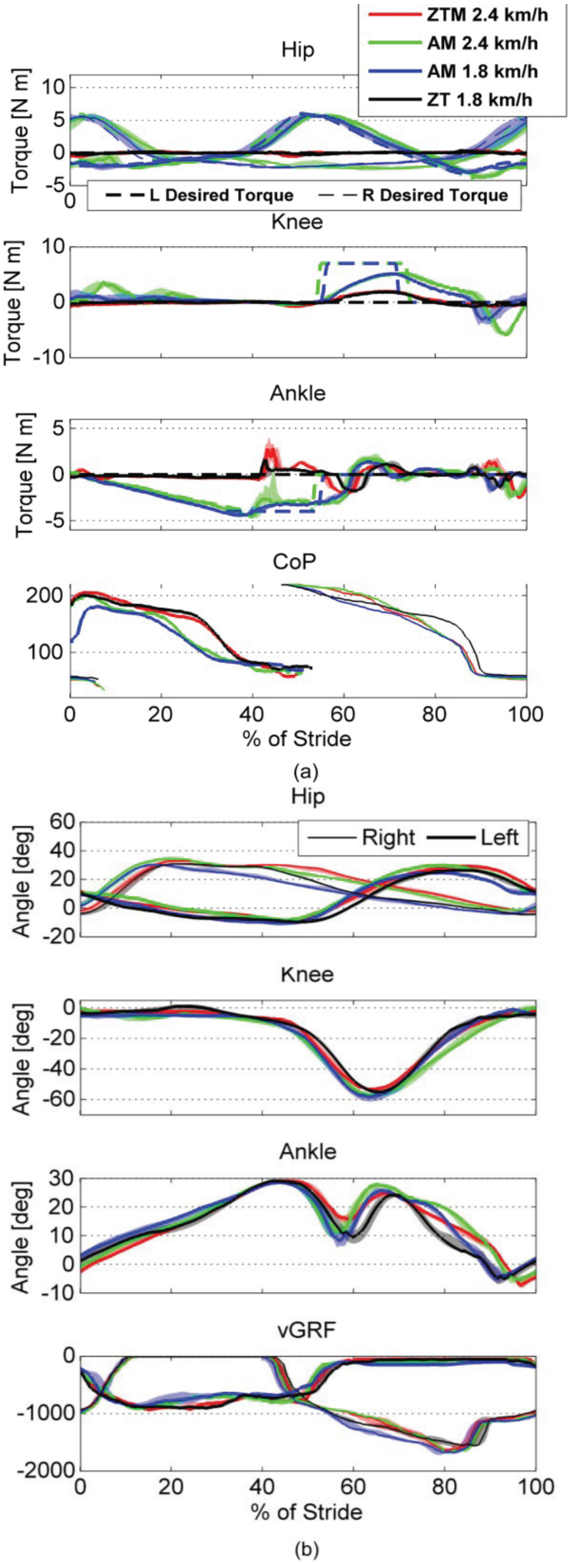

Fig. 6. The joints torques (a) and angles (b) of the HKAF exoskeleton under different walking conditions. In both (a) and (b), the solid lines represent the median value of the variables resampled over all the segmented strides, while the shadow represents the interquartile range; the thiner lines in the hip joint and torque plots represent results from the right hip joint; the dashed lines in plot (a) give the desired torque profiles.

The kinematics of hip, knee and ankle joints under different walking conditions were displayed in Fig. 6(b). All three joints generated smooth and repetitive trajectories under assistance. This means that during steady-state walking, the gait patterns were not modified by the assistive action of the HKAF exoskeleton joints.

To quantitatively evaluate the effect of the assistive torque on the human kinematics, we compared amplitude and timing of local maximum and minimum of joint angles in ZTM and AM modes. Results were reported in TABLE II. The up-warding arrow "↑" means the local maximum/minimum (max/min) of the joint angle was increased/ delayed within the stride; on the contrary the down-warding arrow "↓" means that the local max/min of the joint angle was decreased/anticipated within the stride. Looking at the data of both subjects, the maximum change in the kinematics was about 2.79 deg at the right hip, 3.49 deg at the left hip, 8.54 deg at the knee and 8.84 deg at the ankle. Maximum change in the timing of local max/min was always under 2% of the gait stride. The loading effect of the assistance on the kinematics was higher on Subject #2: this may be given to both/either a different physical and/or behavioural interaction between the user and the device.

TABLE II. DIFFERENCES OF HIP, KNEE AND ANKLE JOINTS MOVEMENTS INTROUDCED BY ASSISTANCE.

		Higher Speed		Lower Speed	
		Max[a]	Min	Max	Min
Subject #1	Right Hip	1.30↑ 0.33%↓	0.94↑ 0.13%↓	0.55↓ 0.42%↓	0.33↑ 0.39%↓
	Left Hip	0.62↑ 0.51%↓	0.40↑ 0.05%↓	1.67↓ 0.56%↓	1.07↓ 0.38%↓
	Knee	1.56↑ 0.1%↓	4.11↓ 0.05%↑	2.32↓ 0.27%↓	2.97↓ 0.18%↓
	Ankle	0.09↓ 0.11%↓	2.13↑ 0.05%↓	0.48↓ 0.17%↑	0.32↑ 0.19%↓
Subject #2	Right Hip	1.29↓ 1.38%↑	1.19↑ 0.59↓	2.79↓ 1.19%↓	2.41↓ 0.40%↓
	Left Hip	1.76↑ 0.06%↓	3.04↑ 0.09%↓	3.49↓ 1.45%↓	0.90↓ 0.70%↓
	Knee	0.64↑ 0.36%↑	4.57↓ 0.09%↑	5.33↑ 0.69%↓	8.54↓ 0.21%↓
	Ankle	0 0.09%↓	2.95↑ 0.05%↓	0 0.03%↓	8.84↓ 0.31%↓

a. In each cell, the upper and lower values are respectively the changed amplitude and timing of max/min angles.

B. *Assistive performances during speed transtions*

Performance of the controller during speed transitions (decreasing speed) recorded from both subjects are presented in Fig. 7. We showed hip joint angles before and after a change of speed change. In addition, we also showed how hip assistive torque took about 2 steps to adapt to the new speed: this is a consequence of the time necessary to let AOs to adapt to the new pattern, being the hip assistive torque dependant on the estimated/predicted hip angles. On the contrary, when we look at the knee and ankle joints, the assistive torque is not affected by the speed variations, being the algorithm in this case reliant on output of the finite-state machine. Hence, we can conclude that the proposed controller is capable to adapt to different walking patterns in a relatively short time.

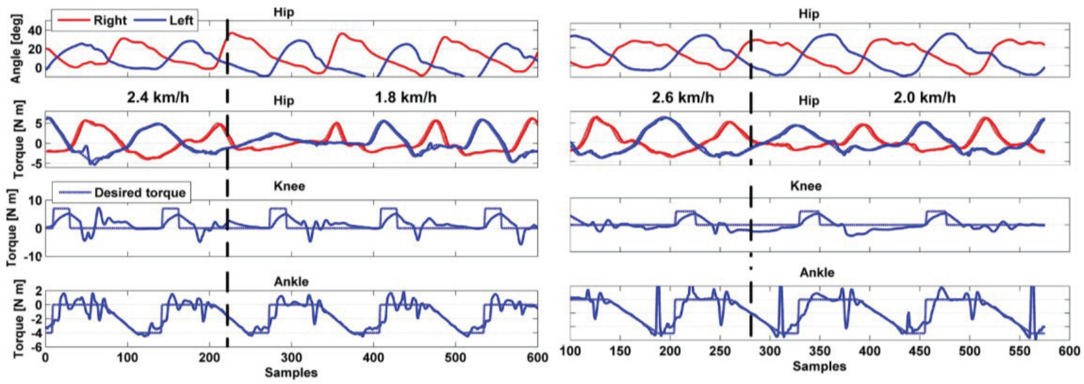

Fig. 7. The assistive performances during speed transitions. Black dashed lines represented the speed transition moments. The speed of subject #1 (left graph) changed from 2.4 to 1.8 km/h; the speed of subject #2 (right graph) changed from 2.6 to 2.0 km/h.

V. Conclusions

This paper proposed a novel AOs-based assistive strategy for a multi-joint lower-limb exoskeleton. A double-block structure was designed: an AOs-based controller for hip assistance and a finite-state machine controller for knee-ankle assistance. Thanks to the walking frequency estimated by AOs, it is possible to set a biomechanically consistent and adaptive assistive torque, with only a few parameters requiring tuning from the experimenter. Validation experiments were performed with two subjects walking on a treadmill and wearing a HKAF exoskeleton. Results confirmed the proposed controller is capable to provide stable and adaptive assistance with a limited loading effect on the user kinematics. Futures works will be aimed at carrying out a more extensive experimental validation with both healthy volunteers and patients with mild lower-limb impairments.

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