

Compliant Control of a Transfemoral Prosthesis by combining Feed-Forward and Feedback*

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Abstract—This paper reports the development and preliminary validation of a bio-inspired controller using an adaptive oscillator and artificial primitives for a powered ankle-knee prosthesis, namely the CYBERLEGsPlusPlus Gamma-prosthesis. The proposed controller combines a feed-forward reference torque component with a feedback impedance-based torque component for the prosthetic knee and ankle joints. A treadmill walking experiment was conducted with a transfemoral amputee to assess the relevance of the feed-forward torque component, compared to a pure feedback control approach. Results showed that adding a feed-forward torque component to both prosthetic joints allowed a decrease of the feedback gains, and thus a more compliant behavior of the prosthesis. This combined condition was well-perceived by the pilot transfemoral amputee.

I. INTRODUCTION

Lower-limb amputation is a seriously disabling condition that affects the overall quality of life and the locomotion performance. It was estimated that 1.1 million people were living with lower-limb loss in 2010 in the United States, and this number is expected to rise to 1.4 million in 2020 and 2.4 million in 2050 [1]. Locomotion of lower-limb amputees requires great energy expenditure and is characterized by slower speed and less stability, compared to able-bodied people [2], [3]. These co-morbidities are further enhanced in case of above-knee amputation, which concerns 27% of lower-limb amputees [4], due to the complexity of the knee system.

Designing efficient solutions to replace the amputees missing limb remains challenging. Current commercial transfemoral prostheses are mostly modular, passive devices, i.e. they are not enable to inject additional energy into the human-device complex. Among those are, for example, the Ottobock C-leg [5], the Ossur Pro-Flex XC [6] and the Blatchford Linx integrated prosthesis [7]. However, recent evidence showed that lower-limb amputees could benefit from using an active prosthetic device, for example by decreasing loading of the sound limb during walking [8] or reducing the metabolic cost [9]. To date, only a limited number of active devices is commercially available, such as the Ossur Power Knee [10] and the Ottobock Empower [11].

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Next to them, an increased number of research prototypes are being developed in order to address these challenges. These are e.g. the Vanderbilt prosthesis [12], the AMPRO3 [13] or the CYBERLEGsPlusPlus (CLs++) prosthesis [14], among others.

The potential benefit of using an active transfemoral prosthesis strongly depends on its control method. A wide range of control strategies are being investigated in different orthotic or prosthetic devices [15]. Most of them rely on finite-state machines, i.e. they decompose the gait cycle into distinct periods associated to different control laws (see for example [16], [17]). These controllers usually involve many parameters that must be tuned for each task and each user. On the other hand, several continuous control laws were recently explored on different devices. Many are gait pattern generators, i.e. they define a time- or phase-dependent trajectory that must be tracked by the prosthetic joints. This predefined trajectory can be based on healthy walking biomechanical data [18], [19], on the kinematics of the residual limb [20], on recorded muscle activity [21], or on other signals [22]–[24]. An emerging trend consists in making the trajectory adaptive to the subject's behavior and performance in real-time [25], [26], although most adaptive control methods are still at the simulation stage [27]–[29].

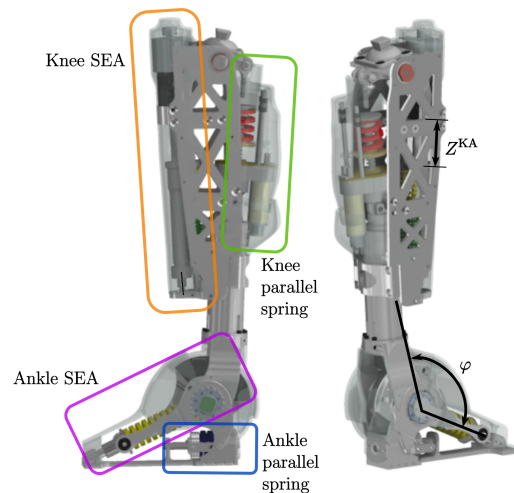


Fig. 1. CAD drawing of the CLs++ Gamma-Prosthesis. The prosthesis is equipped with knee and ankle series elastic actuators (SEA) and parallel springs. The ankle SEA drives the position of the moment arm denoted φ , while the knee SEA drives the position of the carriage along a linear axis, denoted Z^{KA} .

Some of the approaches outlined above combine feedback and feed-forward control. While pure feedforward controllers do not tolerate gait variability and environmental perturbations, pure feedback controllers may involve high gains, and thus possible high forces and instability [15]. Moreover, manual tuning of the feedback gains can be tedious, especially since it must be done for each participant and each locomotion task [30]. Consequently, there have been efforts to decrease the feedback contribution of prosthesis controllers and to design methods for automatic tuning of these gains, such as model-based optimization [31] or online tuning [32]–[34]. Interestingly, studies on the mechanisms involved in biological limb movement highlight the importance of using both types of control, and potentially for limb prosthesis [35]. According to [36], the cerebellum plays a crucial role in this framework, in particular for human walking. More precisely, this neural structure is putatively the place where both control modalities are combined: (1) feed-forward predictive control based on continuous task learning, and (2) feedback error correction due to the unpredictability of real-world environments and variability of the gait. Taking inspiration from this observation, we provide a bioinspired, primitive-based controller that combines feed-forward and feedback torque trajectories for the knee and ankle joints of the CLs++ Gamma-prosthesis. In particular, the present contribution evaluates the relevance of adding feed-forward control to both joints during a walking experiment with a

transfemoral amputee.

The paper is structured as follows. Section II provides the operating mode of the CLs++ Gamma-Prosthesis and a description of the proposed controller architecture. It also defines the experimental protocol. The experimental results are then reported in Section III, and discussed in Section IV. The paper ends with a conclusion.

II. MATERIALS AND METHODS

A. The CYBERLEGSPlusPlus Gamma-Prosthesis

The CLs++ Gamma-Prosthesis is a powered knee-ankle prosthesis equipped with Series Elastic Actuators (SEA) and parallel springs. Its CAD drawing highlighting its different modules is provided in Figure 1. It is an improved version of the CLs++ Beta-Prosthesis, for which a detailed description of the design and working principles is provided in [14], [18]. The ankle is powered by a variable stiffness actuator that can rapidly increase stiffness when displaced from the neutral position. It drives the moment arm angular position denoted φ (see Figures 1 and 2). The ankle parallel spring stores energy during plantarflexion and releases it during dorsiflexion to assist push-off, thus decreasing the actuator peak torque. The knee joint combines a SEA with a Weight Acceptance (WA) system. The WA system consists of a simple spring that can be engaged by a motor to provide an extension torque at the knee joint, e.g. during the WA phase of the gait cycle. Then it can be disengaged to allow knee flexion, e.g. during swing. The knee SEA consists of a motor

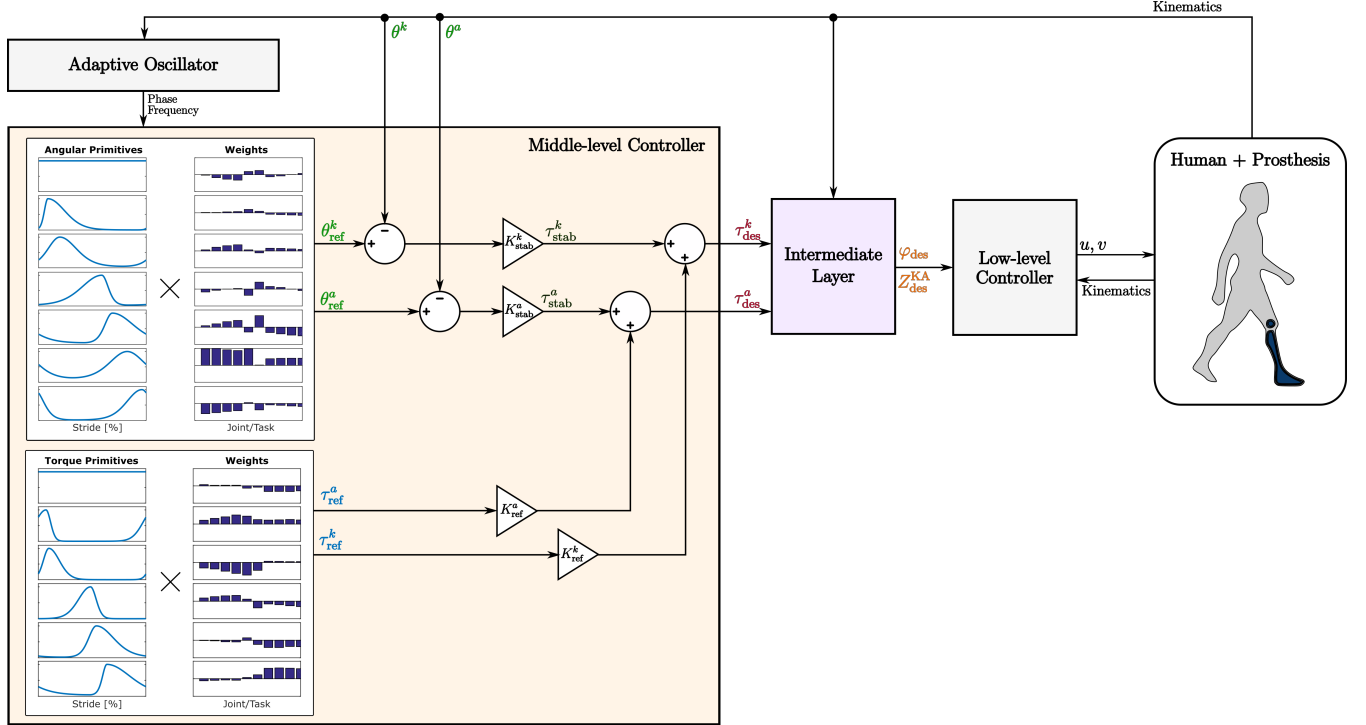


Fig. 2. Control architecture of the CLs++ Gamma-Prosthesis. The adaptive oscillator detects the global locomotion parameters, such as the gait phase and cadence. The middle-level layer uses these parameters to compute the desired torque patterns τ_{des}^k and τ_{des}^a for the prosthetic knee and ankle joints respectively. The intermediate layer converts the desired torque into reference positions for the joints actuators (φ_{des} for the ankle SEA and Z_{des}^{KA} for the knee SEA) [18]. Finally, the low-level controller is in charge of tracking the desired position at the level of each joint, by computing motor current setpoints u and v for the knee and ankle joints respectively.

that drives the position of a carriage along a linear axis. This position is denoted Z^{KA} (see Figures 1 and 2). The carriage embeds two opposite sets of springs: the baseline springs provide a flexion torque at the knee joint, and the extension springs provide an extension torque. Both prosthetic joints are able to provide compliant actuation during the gait cycle.

B. Controller Architecture

Figure 2 shows the block diagram of the prosthesis controller. The gait phase and frequency are estimated in real-time by means of an adaptive oscillator [37], [38]. Then, the middle-level controller uses these locomotion parameters to provide desired torque patterns for the prosthetic knee and ankle joints. The intermediate layer converts the desired torques into reference positions for the joints actuators. Finally, the low-level controller is in charge of tracking the desired position at the level of each joint. More details on the adaptive oscillator and the middle-level layer are given in the next paragraphs. Details on the intermediate layer can be found in [18].

1) *Gait phase and cadence estimation:* Adaptive oscillators are mathematical tools that capture the periodic characteristics of a signal in state variables [37], [38]. Therefore, they are suitable to detect the global locomotion parameters in real-time. In the proposed controller, a single adaptive oscillator was implemented based on the work of [39], using the prosthetic hip angle as input, and the ground reaction force of the prosthetic foot for heel strike detection. Its outputs are estimates of the gait phase ϕ and frequency.

2) *Bioinspired torque patterns:* The desired knee and ankle torques are computed in real-time by the middle-level layer as the sum of a feed-forward, primitive-based torque and a feedback, impedance-based torque. In physiology, motor primitives are basic signals that can be recombined to generate different locomotion patterns [40]. Using these primitives offers to reduce the controller dimensionality and ensures coordination between the prosthesis joints. The primitive-based torques (τ_{ref}^k and τ_{ref}^a) are computed as a weighted sum of precomputed, fundamental motor primitives, where the weights are modulated as a function of the estimated cadence. The impedance-based torques (τ_{stab}^k and τ_{stab}^a) are computed from the difference between the actual joint kinematics (θ^k and θ^a) and reference angular

trajectories (θ_{ref}^k and θ_{ref}^a). These angular trajectories are computed as a weighted sum of precomputed, fundamental angular primitives, where the weights are modulated as a function of the estimated cadence. Concretely, the desired joint torques are thus equal to:

$$\tau_{\text{des}}^a = K_{\text{stab}}^a(\theta_{\text{ref}}^a(\phi + \phi_{\text{off}}) - \theta^a(\phi)) + K_{\text{ref}}^a\tau_{\text{ref}}^a(\phi) \quad (1)$$

$$\tau_{\text{des}}^k = K_{\text{stab}}^k(\theta_{\text{ref}}^k(\phi + \phi_{\text{off}}) - \theta^k(\phi)) + K_{\text{ref}}^k\tau_{\text{ref}}^k(\phi) \quad (2)$$

where K_{stab}^a , K_{ref}^a , K_{stab}^k and K_{ref}^k are tunable gains. The motor and angular primitives were extracted beforehand from literature data (see Appendix for more details). The estimated phase ϕ ensures their synchronisation with the actual gait. Moreover, since lower-limb amputee gait is generally characterized by a shorter stance time compared to healthy walking [41], a configurable phase offset ϕ_{off} was added to the reference angle trajectories in order to: (i) shift the transition between ankle dorsiflexion and plantarflexion characterizing push-off; (ii) shift the knee, swing flexion peak immediately after toe-off to assist ground clearance; and (iii) make sure that the knee is fully extended at the end of swing, just before heel-strike (see Figure 4(a)-(b)).

C. Experimental Protocol

The experiment was conducted with a transfemoral amputee walking on a treadmill with the Gamma-Prosthesis and the proposed controller (see Figure 3). The subject gave informed consent before starting the experiment and the protocol was approved by the institutional medical ethics committee of UZ Brussel and Vrije Universiteit Brussel (Belgium, B.U.N. 143201732970) and the Federal Agency for Medicines and Health Products (FAMHP reference number: 80M0725). The pilot subject was 61 years old and underwent his amputation 30 years earlier due to osteosarcoma. His weight is 63 kg, without any prosthesis. During the experiment, the subject also wore a thigh IMU and a sensing insole on the prosthetic side to measure the hip joint angle and the ground reaction force respectively, similarly as in [42]. These data were used as inputs for the adaptive oscillator.

With the aim of assessing the relevance of the feed-forward torque component, the experiment consisted of three consecutive trials:

- (1) The desired torque of both joints was impedance-based only ($K_{\text{ref}}^k = K_{\text{ref}}^a = 0$);

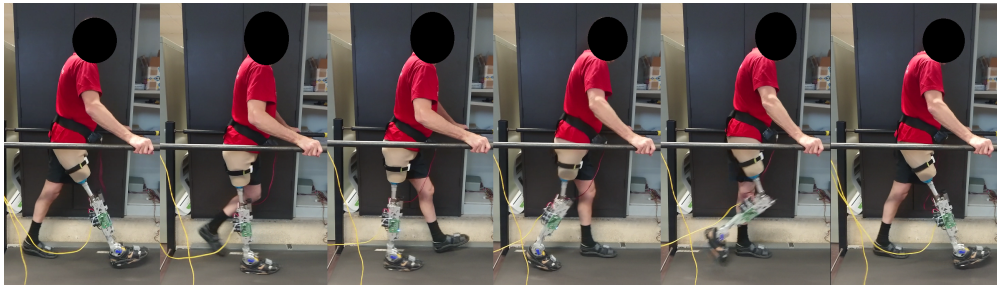


Fig. 3. Snapshots of the amputee walking on the treadmill with the CLs++ Gamma-Prosthesis.

FEEDBACK PARAMETERS				Trial (1)	Trial (2)	Trial (3)
Impedance gains K_{stab}^i (Nm/deg)	Knee	Stance	Flexion	0.025	0.025	0
			Extension	1.30	1.30	1.00
		Swing	Flexion	0.80	0.50	0.25
			Extension	0.80	0.75	0.55
	Ankle			2.50	2.50	2.50
	Phase offset ϕ_{off} (% stride)			20	20	20
Reference gains K_{ref}^i (% of τ_{ref}^i)		Knee	0	0	40	
		Ankle	0	40	40	

FEED-FORWARD PARAMETERS

TABLE I

VALUES OF THE CONTROLLER GAINS AFTER THE TUNING SESSIONS

- (2) The desired torque of the knee joint was impedance-based only ($K_{ref}^k = 0$), and the desired torque of the ankle joint was a combination of an impedance-based torque and a reference torque ($K_{ref}^a > 0$);
- (3) The desired torque of both joints was a combination of an impedance-based torque and a reference torque ($K_{ref}^k > 0$ and $K_{ref}^a > 0$).

Each trial was organized in two steps:

1) *Controller tuning*: Before the actual data recording, the subject walked on a treadmill while the controller gains were tuned until a comfortable gait was reached. The treadmill speed was set to 2.3 km/h. The parameters that were tuned were: the gains of the impedance-based torque (K_{imp}^a and K_{imp}^k , in Nm/deg), the gains of the reference torque (K_{ref}^a and K_{ref}^k , in % of the primitive-based torque) and the phase offset of the reference angle trajectories (ϕ_{off} , in % stride). For the knee joint, the impedance gain was independently tuned for the swing phase and stance phase (detected from the ground reaction force). Furthermore, it was independently tuned for the knee flexion and extension (determined from the sign of the angular error ($\theta_{ref}^k(\phi + \phi_{off}) - \theta^k(\phi)$) in Equation (2)). The final values are given in Table I for each trial.

2) *Data Collection*: After tuning the controller gains, the subject walked on a treadmill for 3 minutes while data were recorded. The treadmill speed was set to 2.3 km/h. The following data were collected: the prosthetic joint angles (θ^k and θ^a) and actuator positions (ϕ and Z^{KA}), the hip angle and the ground reaction force on the prosthetic side. The following data were computed by the controller and collected for data analysis: the estimated gait phase and cadence, the prosthetic joint reference positions (θ_{ref}^k and θ_{ref}^a), the desired torque trajectories (τ_{stab}^k , τ_{stab}^a , τ_{ref}^k and τ_{ref}^a) and the actuator desired positions (ϕ_{des} and Z_{des}^{KA}). The data acquisition rate was 100 Hz.

III. EXPERIMENTAL RESULTS

During the three trials, the amputee was able to walk with the CLs++ Gamma-prosthesis and the proposed controller, and the recorded data showed very repetitive profiles across the gait cycles. The results were averaged over (1) 84, (2) 64

and (3) 83 consecutive cycles, where (1), (2) and (3) refer to the trial number. The mean walking cadence was: (1) 0.57 stride/sec, (2) 0.55 stride/sec, and (3) 0.56 stride/sec.

Figure 4(a)-(b) compares the averaged joint angles over the gait cycle that were measured during the three trials, for the knee and the ankle respectively. Figure 4(c)-(d) shows the corresponding actual joint torques that were computed from inverse kinematics relationships [18]. Figure 5 shows the ankle mechanical work per step that was measured in the three trials, i.e. the product between the ankle torque and its velocity, normalized to the subject's weight. Figure 6 shows the mean contribution of the feedback and feed-forward torque components to the total torque, for the ankle joint during Trials (2) and (3), and for the knee joint during Trial (3). Finally, a movie showing representative cycles of the three trials is provided as supplementary multimedia material.

A. Addition of the Ankle Feed-forward Torque Component

Compared to Trial (1), a reference torque component $K_{ref}^a \tau_{ref}^a$ was added to the desired torque at the level of the ankle in Trial (2). The reference torque component accounted for 32.0% of the total ankle torque on average during stance, and 16.6% during swing (see Figure 6). As shown in Figure 4(b), this resulted in an increased plantarflexion angle at push-off (from 7.7 deg to 20.5 deg). Although the observed maximum plantarflexion angle is higher than the reference one (equal to 4.5 deg), this was well appreciated by the subject, because it facilitated foot clearance. Consequently, the impedance gains of the knee joint during swing could be decreased as compared to Trial (1) (see Table I). This led to a decreased maximum knee flexion angle during swing, from 76.3 deg to 66.2 deg. In this way, the knee behavior was perceived by the subject as being smoother, and thus more comfortable. Furthermore, the mean stance/swing ratio was 0.79 in Trial (1) and 0.81 in Trial (2). Finally, the averaged ankle work per step increased from 0.17 J/kg to 0.39 J/kg between Trial (1) and Trial (2), as pictured in Figure 5. As a reference, a healthy ankle is characterized by 0.21 J/kg per step at normal speed [43].

B. Effect of the Knee Feed-forward Torque Component

Compared to Trial (2), a reference torque component $K_{ref}^k \tau_{ref}^k$ was added to the desired torque at the level of the knee in Trial (3). In Trial (3), on average, the reference torque component accounted for 30.9% and 17.1% of the total ankle torque during stance and swing respectively, and for 94.1% and 44.4% of the total knee torque during stance and swing respectively (see Figure 6). Figure 4(b) shows that this resulted in a further increased plantarflexion angle at push-off (from 20.5 deg to 24.7 deg), and it was again well appreciated by the subject since it facilitated foot clearance. Consequently, the impedance gains of the knee joint K_{stab}^k during swing could further be reduced compared to Trial (2) (see Table I), increasing the knee joint compliance. Although the observed maximum knee flexion angle during swing was increased compared to Trial (2) (from 66.2 deg to 72.7 deg),

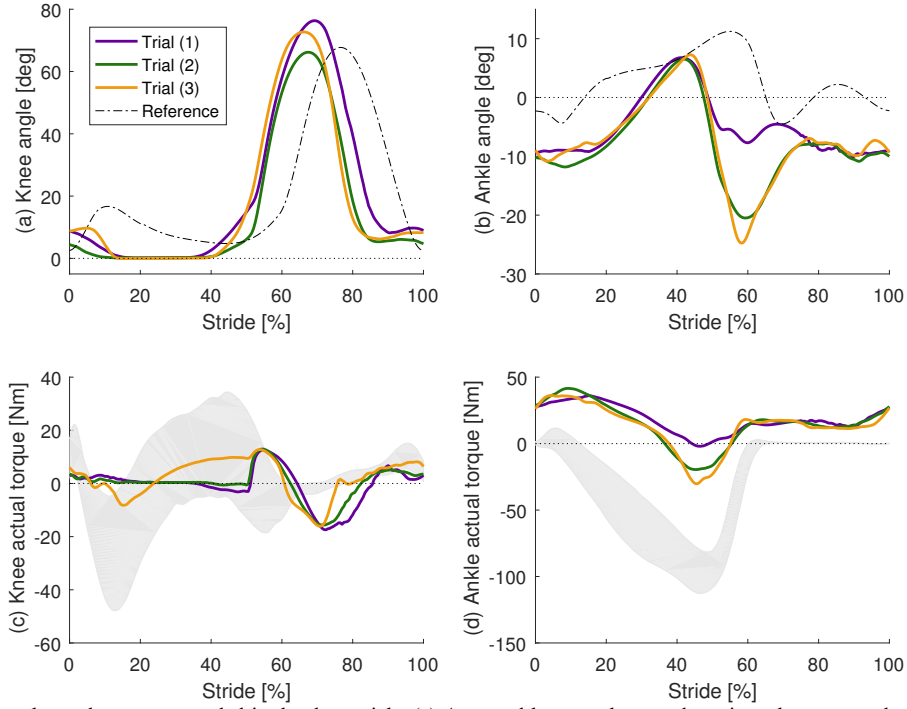


Fig. 4. Averaged joint angles and torques recorded in the three trials. (a) Averaged knee angle over the gait cycle, compared to the primitive-based knee reference angle; (b) Averaged ankle angle over the gait cycle, compared to the primitive-based ankle reference angle; (c) Averaged torque at the knee joint over the gait cycle, compared to a reference torque profile (in grey) taken from [43]; (d) Averaged torque at the ankle joint over the gait cycle, compared to a reference torque profile (in grey) taken from [43].

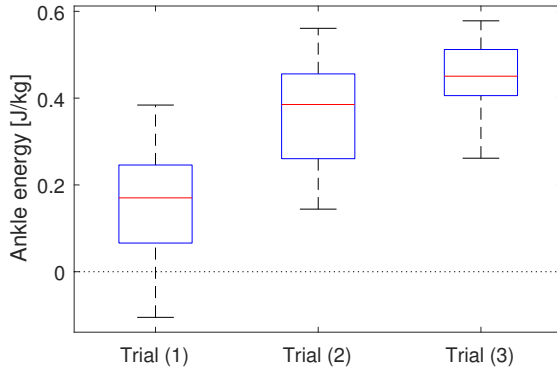


Fig. 5. Ankle work per step, normalized to the subject's weight, measured during the three trials.

the knee behavior was again perceived as being smoother by the subject, and thus more comfortable. Moreover, the mean stance/swing ratio increased to 0.88 in Trial (3). Note however that this is still far from the ratio of 1.5 reported for healthy walking [43]. Finally, Figure 5 shows that the averaged ankle work per step increased from 0.39 J/kg to 0.45 J/kg between Trial (2) and Trial (3).

IV. DISCUSSION AND CONCLUSION

We developed a bioinspired controller combining feed-forward primitive-based and feedback impedance-based torque trajectories for the CLs++ Gamma-Prosthesis. We

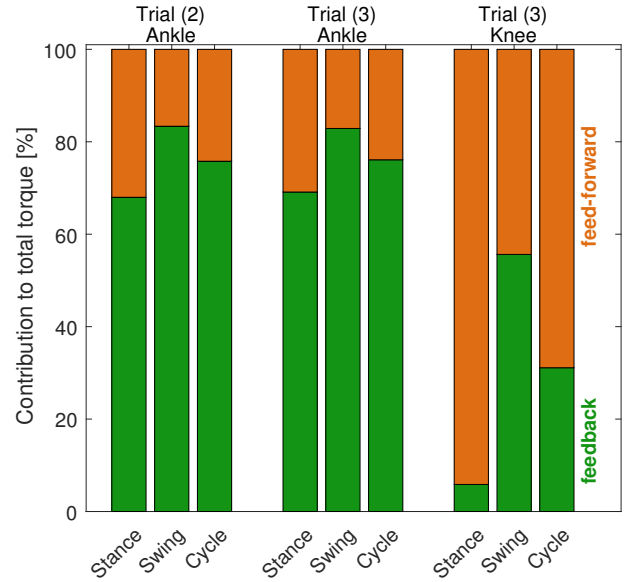


Fig. 6. Averaged contribution of the feedback and feed-forward torque components to the total joint torque, for the ankle joint in Trial (2), and for the ankle and the knee joints in Trial (3).

conducted an experiment with a transfemoral amputee walking on a treadmill and compared three cases: (1) using only feedback control for the ankle and the knee joints; (2) adding feed-forward control to the ankle joint; and (3)

adding feed-forward control to both joints. Results showed that the contribution of the reference torque component is relevant for both joints, compared to an impedance-based torque only. Indeed, it was well perceived by the amputee subject and allowed to have a more compliant behavior of the prosthesis. In particular, we observed that adding a feed-forward torque component at the ankle level increased the push-off torque, as well as the plantarflexion angle at push-off. We postulate that these extra torque peaks facilitated the ballistic movement of the leg, so that we could decrease the impedance gains of the knee joint during the swing phase.

Interestingly, the pilot subject appreciated the decrease of the knee impedance gains when a feed-forward torque component was added (either at the ankle only or at both joints), but he preferred keeping the ankle impedance gains unchanged. As consequence, in Trial (3), the feedback torque component accounted for more than 76% of the total ankle torque, compared to 31% of the total knee torque. In other words, the prosthetic knee was mostly controlled by a feed-forward signal, whereas the prosthetic ankle was mostly controlled by a feedback signal. Interestingly, this observation is consistent with the hypothesis postulating the existence of a proximo-distal gradient in the vertebrate leg. This hypothesis states that biological distal muscles (i.e. muscles close to the feet) are mostly controlled by feedback-based signals such as reflexes, whereas proximal muscles (i.e. muscles close to the hips) also receive efferent inputs from central pattern generators, i.e. following a feed-forward pathways [29], [44].

Our study faces some limitations that should be pointed out. First, the walking experiment was performed on a treadmill, and the results may vary in an ecological, uncontrolled environment where balance issues would be more prominent. Then, the current version of the controller requires manual tuning of several parameters that are subject-specific. This is acceptable giving the main goal of the experiment, but improvements should be done towards (semi-)automatic tuning in future developments. Finally, the primitive-based reference torque and angular trajectories adapt neither to the subject's performance nor behavior, since they were pre-computed from healthy walking. Indeed, the hypothesis that healthy walking is the best way of walking for lower-limb amputees, or is even attainable by them, is still unverified. Consequently, future work will focus on the development of performance-based reference trajectories for the prosthetic knee and ankle joints.

APPENDIX - PRIMITIVE EXTRACTION METHOD

The motor and angular primitives used in the proposed controller were extracted from a set of reference torque and angular trajectories respectively. The database included reference patterns for the knee and ankle joints during treadmill walking at different speeds, provided in [45]. An asymmetric Gaussian shape was imposed to the primitives, i.e. each primitive can be expressed as:

$$\mathcal{G}(\text{phase}) = \exp\left(\frac{\cos 2\pi f(\text{phase} - \mu, \frac{d+1}{2}) - 1}{\sigma}\right) \quad (3)$$

$$\text{where } f(x', d') = \begin{cases} \frac{x'}{2d'} & \text{if } x' < d' \\ 1 - \frac{1-x'}{2(1-d')} & \text{otherwise} \end{cases}$$

Therefore, each primitive is fully described by three parameters: its mean μ , its standard deviation σ and its asymmetry index d . An "offset primitive" was added to each set of primitives, defined as $\mathcal{G}(\text{phase}) = 1$. We used a constrained Particle Swarm Optimization (PSO) to extract the primitives from the set of reference signals. The evaluated cost function was the global root mean square error (RMSE) between the reference signals and the reconstructed ones. For each primitive, we constrained the maximum absolute value of its associated weights to be lower than the maximum absolute value of the signals of the defined database. The selected number of primitives corresponds to the minimum number of primitives associated to a maximum normalised RMSE lower than 4%. We obtained 5 torque primitives and 6 angular primitives. They can be seen in Figure 2. The primitive associated weights across the different walking cadences were interpolated by second-order polynomials.

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