6th IEEE RAS/EMBS International Conference on
Biomedical Robotics and Biomechatronics (BioRob)
June 26-29, 2016. UTown, Singapore

Effects of a neuromuscular controller on a powered ankle exoskeleton during human walking

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Abstract— **Wearable devices to assist abnormal gaits require controllers that interact with the user in an intuitive and unobtrusive manner. To design such a controller, we investigated a bio-inspired walking controller for orthoses and prostheses. We present (i) a Simulink neuromuscular control library derived from a computational model of reflexive neuromuscular control of human gait with a central pattern generator (CPG) extension, (ii) an ankle reflex controller for the Achilles exoskeleton derived from the library, and (iii) the mechanics and energetics of healthy subjects walking with an actuated ankle orthosis using the proposed controller. As this controller was designed to mimic human reflex patterns during locomotion, we hypothesize that walking with this controller would lead to lower energetic costs, compared to walking with the added mass of the device only, and allow for walking at different speeds without explicit control. Preliminary results suggest that the neuromuscular controller does not disturb walking dynamics in both slow and normal walking cases and can also reduce the net metabolic cost compared to transparent mode of the device. Reductions in tibialis anterior and soleus activity were observed, suggesting the controller could be suitable, in future work, for augmenting or replacing normal walking functions. We also investigated the impedance patterns generated by the neuromuscular controller. The validity of the equivalent variable impedance controller, particularly in stance phase, can facilitate serving subject-specific features by linking impedance measurement and neuromuscular controller.**

I. INTRODUCTION

A widely accepted practice in gait rehabilitation by physiotherapists and robotic devices entails active assistance of lower body movement [4], [12]. This yields several positive effects, such as soft tissue stiffness reduction, increase in muscle strength, and increase in brain plasticity by providing somatosensory stimulation that correlates with motor output [21], [15], [16]. As robotic devices increase in popularity for both gait training and gait assistance, interaction control (i.e. shared control) between the exoskeleton and the human becomes important for recovery, understanding user intention, and adaptation to outside environments. Some researchers have shown that lack of interaction control can reduce the recovery capacity, such as if the assistance encourages slacking (slacking hypothesis [18]). To address the negative consequences of the slacking hypothesis, researchers have developed a new class of controllers to provide "assistance-as-needed", such as strategies based on impedance modulation [19], [1] - helping the subject only when away from a reference pattern (kinematic in most cases) - or proportional myoelectric control [17], [7], [8], [11], where the control output is directly proportional to the magnitude of surface electromyography (EMG) signals [11]. One advantage of proportional myoelectric controllers is that they do not require a reference, as their control signals are directly derived from muscle activation patterns. This type of interaction control may be robust to environmental changes and thus suitable for wearable devices. Myoelectric controller also promotes neural plasticity, as device wearers can actively initiate and modulate the device's actions. However this method relies on clean and reliable EMG signals from existing and functional muscles, which may limit their application for certain patient groups.

As the main long-term goal of our control approach is to design walking controllers for wearable orthoses that assist altered and/or pathological gaits, developing such a controller requires a deeper understanding of the complex dynamics that involves both the biomechanical (i.e. body-environment dynamics, musculo-skeletal structure) and the neuro-physiological (i.e. muscles, sensors, neural networks) level. Therefore the controller should ideally not only have a good dynamics fidelity but also emulate the properties of the neuro-physiological components (e.g. muscles, sensors, neural networks). Our proposed control approach utilizes models of human muscles and tendons in order to generate motion that is compliant with both the user's abilities and natural walking dynamics. The hypothesis is that the functional biomimetics will promote active recruitment of the user's own neuromuscular system.

The controller we propose, referred to as neuromuscular controller (NMC), has the capacity to work in parallel with a healthy ankle without impeding its function while only using few sensory inputs. The approach relies on the use of neuromuscular models (NMM) [10], [5] - muscle-tendon models used in neuromechanical simulation of human gait - as the basis for the design of the exoskeleton controller. The novelty of the control paradigm is to base the controller on functional models of dynamic elements in the human leg (i.e. coming from the NMM) rather than observed properties of human gait, such as libraries of desired joint trajectories. This offers several advantages compared to other approaches:

- Robustness & adaptability: With no predetermined pat-

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tern, walking emerges as a result of the interaction of the body with the environment. A consequence is that the exact same controller (with same parameters) can be used under a variety of conditions, such as different speeds and different terrains [5].
- Modularity: As a consequence of the use of a physiologically realistic controller framework, the structure of the model (with use of local feedback/feedforward control) enables ease of control and assistance of specific components of the walking gait. This can be either joint-based (e.g. ankle controller to assist push-off as in this paper, or hip controller to assist a patient with weak hip function) or muscle-based (e.g. assist only ankle extensors, or assist only mono-articular muscles components).
- Simple sensors: Due to the dynamics of the muscle models, modules only need ground contact detectors and joint angles to reproduce walking. The hip modules also require trunk angle relative to gravity.

Reproducing the dynamics of human locomotion does not imply that the neuro-physiological structure of the extended model is valid, but demonstrates that it would be sufficient to produce the desired dynamics. This dynamic fidelity makes the models good candidates for the design of controllers for robotic devices, where the robustness and flexibility of this approach have been successfully demonstrated in the control of a prosthetic ankle device [6]. The bio-inspired aspects of the controller enabled amputee subjects wearing the device to walk on different terrains (e.g. flat ground, ramp ascent / descent, stairs) in a manner comparable to healthy subjects, without the need for explicit terrain sensing [6]. These results demonstrated the potential of NMM in the controller design of lower limbs rehabilitation devices. Indeed the flexibility and robustness of the controller implies compliance with both healthy subjects and patients with remaining function.

Here we demonstrate the first example, to our knowledge, of the application of this principle for the control of orthotic devices. We tested our control paradigm by implementing the ankle module on a powered ankle exoskeleton and investigated the effects of the controller on the gait of healthy subjects. We expect that the NMC would not adversely affect walking mechanics. Instead, we anticipate less plantarflexor activity around push-off, lower joint power at the ankle, and less energetic cost overall.

II. MATERIALS AND METHODS

A. Neuromuscular Controller

We developed the NMC based on an neuromechanical simulation inspired by Geyer's model [10]. This model consists of seven Hill-type muscle units per leg: hip flexor, gluteus, hamstring, vastus, gastrocnemius, tibialis anterior, and soleus. The forces generated by these muscles, depending on muscle length and velocity, combine to produce ankle, knee, and hip joint torques. Different reflex loops are used depending on the gait cycle, and ground contact is used to switch between stance and swing. During stance, the reflex loops induce higher activity in extensor muscle, in order to favor weight bearing support and prepare for push off. When the swing phase is initiated, reflexes induce a reduction of extensor activity and an increase of flexors activity (see Figure 1 for a general description of the state machine and feedback signals driving the model). Although the spinal architecture responsible for locomotion proposed by Geyer is neuro-physiologically incomplete, the dynamics of the produced gaits are in striking agreement with physiological observations of healthy human walking. This agreement is observed at the muscle level, where similar patterns to those recorded in humans with surface EMG are observed [5] at normal walking speeds.

By taking advantage of the modularity aspects of Geyer's model, we created a library of reflex and CPG control modules that can be combined to generate different types of assistive controllers. The library permits controller designs, such as the one implemented in this article, in a drag-and-drop fashion with Simulink (Mathworks, Natick, MA, USA). The Simulink models will be released as part of the libnmm library[1].

1) Ankle module for exoskeleton control: Our controller was implemented on an ankle exoskeleton, and therefore only the distal modules were used. In particular, the control loops of the tibialis anterior and soleus muscles were implemented. During the stance phase, the soleus muscle is driven by positive force feedback to increase tension while negative force feedback decreases the tension on the tibialis anterior (Figure 1). During swing and throughout the gait cycle, a positive length feedback on the tibialis anterior prevents overextension. The virtual gastrocnemius, a bi-articular muscle, was omitted to avoid a shared control issue, as torques would have been created at both the ankle and absent knee joint. Both the tibialis anterior and soleus control loops were used in pure reflex mode, without the CPG component, as previous studies showed the CPG component is more critical for proximal joints, in particular, the hip joint [5]. The parameters used for the control were the same as presented in [10] except for the soleus muscle resting length, which was decreased to 0.3 cm to account for the difference in morphology between the model and the subjects. At the time of testing, we did not scale the controller in respect to the subjects' body weight or height, and we applied the same controller to different subjects at various speeds as an indirect test of robustness.

A gain multiplying the normal torque output of the ankle controller was used to control the level of assistance of the controller. The gain was either 0%, 50%, or 100%, corresponding to Zero, Low, and High levels of assistance. A High level of assistance corresponded to the contribution of the tibialis and soleus muscles required for the neuromechanical simulation of the lower limb model walk in steady-state at 1.3 m/s.

[1]More information on libnmm can be found here: https://bitbucket.org/efx/libnmm/src.

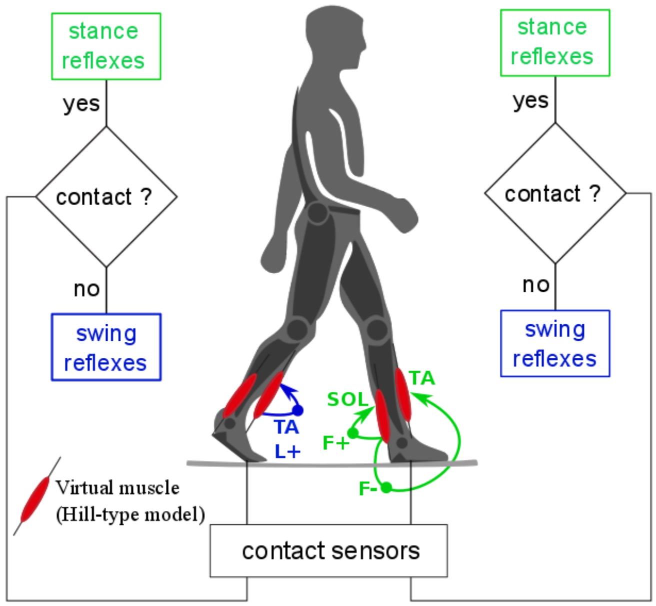

Fig. 1: Schematic view of the ankle module based on the reflex walking model from H.Geyer. Ground sensors are used to detect whether the limb is on stance or swing phase. Then, depending on whether the limb is in contact with ground or in swing, different reflex rules are generated, stance reflexes in green and swing reflexes in blue (Table 1 in [5] shows the different reflex loops acting depending on the gait phases). Here only the soleus and tibialis muscle modules are used. During stance, a positive force feedback (F+) increases the tension in the soleus muscle, and a negative force feedback (F-) decreases the tension on the tibialis anterior to prepare for push off. During the entire gait cycle, a positive length feedback (L+) tries to return the tibialis anterior muscles to a predefined length. Figure inspired from [10].

2) Experimental setup: We investigated the mechanics, energetics, and muscle activity of young, healthy subjects walking with ankle exoskeleton controlled by the NMC. Subjects (N=2) wore the Achilles [13], a motorized ankle exoskeleton which provided various amounts of bilateral torque assistance. Subjects walked at their preferred speed (1.06 m/s, 1.08 m/s) and at a slow speed (0.58 m/s) on an instrumented split-belt treadmill. At self-selected speeds, three different levels of torque assistance were provided (Zero, Low, and High), and two were provided at slow speeds (Zero, Low). The Zero condition corresponded to the transparent mode of the device, with no input from the NMC. The High condition for slow speed was not performed by one of the subjects due to time constraints, and therefore that trial is not included here. The trials were six minutes each and randomized. Subjects' age was 28 and 37 years, their body mass was 80 kg and 77 kg, and their leg length was 0.92 m and 0.92 m. All subjects provided written informed consent prior to the study, according to Institutional Review Board procedures.

We measured metabolic power, electromyography, and gait biomechanics using standard procedures. We recorded the rate of oxygen consumption and carbon dioxide production (CareFusion Oxycon Pro, San Diego, CA USA) and calculated the steady-state metabolic power (in W)

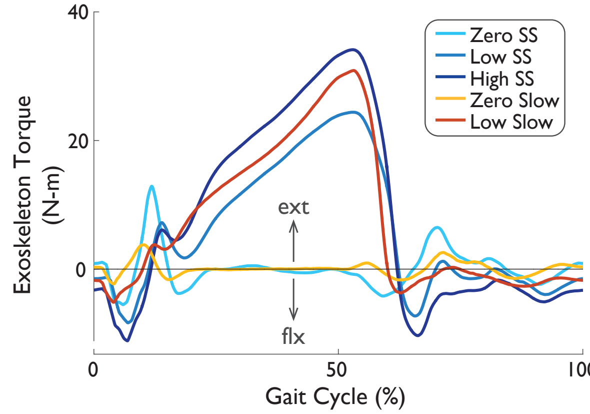

Fig. 2: Measured exoskeleton torque from Zero to High condition at self-selected speeds and for Zero and Low at slow speeds from Subject B. At Zero, the commanded torque from the neuromuscular controller was not provided, and thus the torque command defaulted to zero torque mode. Peak plantarflexion and dorsiflexion torques increased as controller contribution increased. Gait cycle is defined as a full stride starting from heel-strike.

from the last two minutes of each trial using standard conversion factors [2]. Net metabolic power was calculated from subtracting the metabolic power for quiet standing (108 W, 106 W) from the gross metabolic power. We also recorded electromyography (EMG) in the tibialis anterior (TA), soleus (SOL), and medial gastrocnemius (MG) of the left leg (Delsys, Boston, MA, USA). All signals were high-pass filtered with a 20 Hz cutoff frequency (fourth-order Butterworth filter, zero-lag), full-wave rectified, and then low-pass filtered (fourth-order Butterworth filter, cutoff frequency 6 Hz, zero-lag) to obtain the linear envelope of each EMG signal. Kinematic data was measured for one of the subjects with motion capture (Phoenix Technologies Visualeyez, Canada), and 6DOF ground reaction forces were measured from both subjects with an instrumented dual-belt treadmill (Motekforce Link, Amsterdam, the Netherlands). Gait event detection provided to the controller was calculated from the ground reaction forces. Kinematics and inverse dynamics (OpenSim, Stanford, CA, USA) yielded ankle angles, moments, and powers. As with metabolic power, only the last two minutes of the trial were used for analysis.

III. Results

Preliminary results indicated that the controller reduced the energetic cost of walking and lowered soleus and tibialis anterior muscle activities. Subjects' overall walking dynamics were not greatly altered by the controller. In particular, the ground reaction forces and joint angles were qualitatively similar, and no systematic changes in step length or step time were observed.

As the gain on torque assistance increased from Zero to High, the measured torque from the exoskeleton also increased (Figure 2). Peak plantarflexion torque occurred at around 55% of the gait cycle, and peak dorsiflexion occurred around 6% and 65% of stride. Since the NMC torque is dependent on stance and swing phases and ankle angles, walking with different levels of assistance induces gait changes that inherently influence the commanded torques. Speed-related changes in ankle kinematics, such as lower peak plantarflexion angle [22], also affect the commanded torque. Hence, the peak torque at Low is not expected to be half the amount at High, and indeed the peak torque at Low is 70% of the High torque (24 N-m for Low, 34 N-m for High).

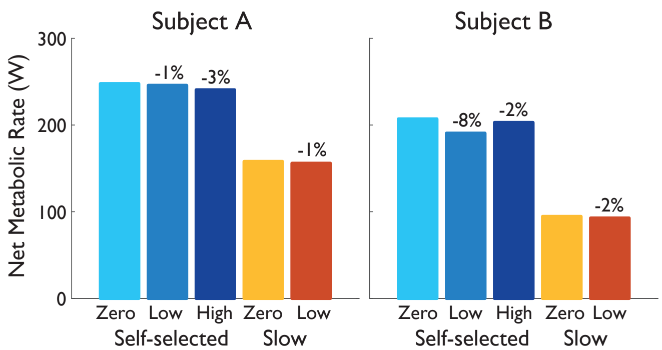

Fig. 3: Net metabolic power (N=2) with Zero to High levels of torque assistance. Both subjects decreased their energy expenditure rate with NMC at both slow and self-selected speeds. The reduction ranged from 1 to 8%, compared against the Zero condition.

Subjects also expended less energy to walk with the reflex controller. Compared against the Zero condition (transparent mode of the device), the NMC conditions reduced the net metabolic rate from 1% to 8% for self-selected speeds and up to 2% for slow speeds (Figure 3). Hence the overall cost of walking with Achilles was reduced, and this could be reflected in the human-like torque profiles provided by the controller. Presumably if the exoskeleton provided some of the ankle torque the subjects would normally exert, then the subjects could choose to use less of their own ankle muscles.

As an indication of muscular effort, muscle activity of the tibialis anterior, soleus, and medial gastrocnemius were also measured. We found decreased activity at the tibialis anterior and soleus (Figure 4), which could partially explain the observed metabolic reduction. Subjects' tibialis anterior (TA) activity decreased near heel-strike and after toe-off. Hence the controller assisted in subjects' dorsiflexion during heel-strike and for ground clearance during swing. Both subjects' soleus activity also decreased near push-off, and as the controller provided more assistance, the plantarflexion generated by the subjects decreased. A less clear trend was observed in the medial gastrocnemius. However, this muscle assists in both ankle plantarflexion and knee flexion, and therefore the effect of an ankle exoskeleton on this biarticular muscle is less straightforward.

While the controller reduced metabolic cost and EMG activity, it did not greatly alter subjects' overall walking dynamics. In particular, ground reaction forces (Figure 5) were relatively unchanged. Qualitatively, the characteristic double hump of the vertical forces was intact with no apparent changes in force loading. Speed had a greater effect, as the initial vertical loading was shallower at slow than at self-

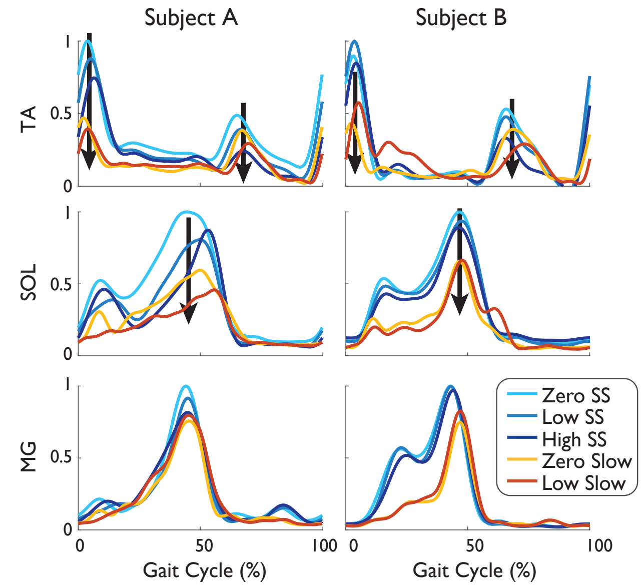

Fig. 4: Muscle activity (N=2) of the tibialis anterior (TA), soleus (SOL), and medial gastrocnemius (MG) over a range of torque assistance conditions. TA activity decreased near heel-strike, and SOL activity decreased near push-off. EMG signals were normalized to peak activation among the muscles for each subject.

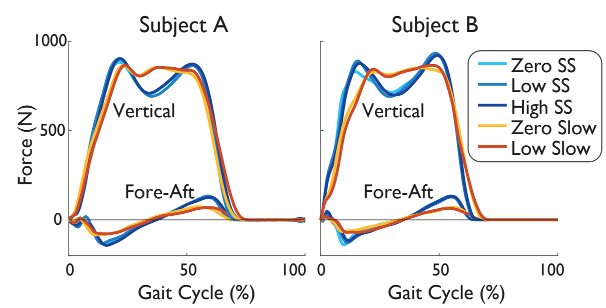

Fig. 5: Fore-aft and vertical ground reaction forces (N=2) at different torque levels over two different speeds. Qualitatively minimal changes were observed with increased torque gain. Greater changes occurred between the two speeds, especially in the loading slopes.

selected speeds. Similarly, there was also no observable trend for changes in step parameters among conditions, except for speed-related changes of shorter step length and longer step time with slower speeds. At self-selected speeds, mean step length was 0.64 m and mean step time was 0.60 s, and at slow speeds, mean step length and step time was 0.50 m and 0.86 s, respectively.

Observations of total and exoskeleton ankle moment and power revealed that subjects' own contribution decreased as exoskeleton assistance increased (Figure 6). Biological ankle moment and power were calculated from subtracting the exoskeleton measurements from the total provided by inverse dynamics. With greater exoskeleton assistance, both the subject's peak biological ankle moment and push-off power decreased by a maximum of -43% and -54% respectively (Figure 7). Hence, with increased assistance by the exoskeleton, the subject provided less push-off power, which could partially explain the lower metabolic cost we

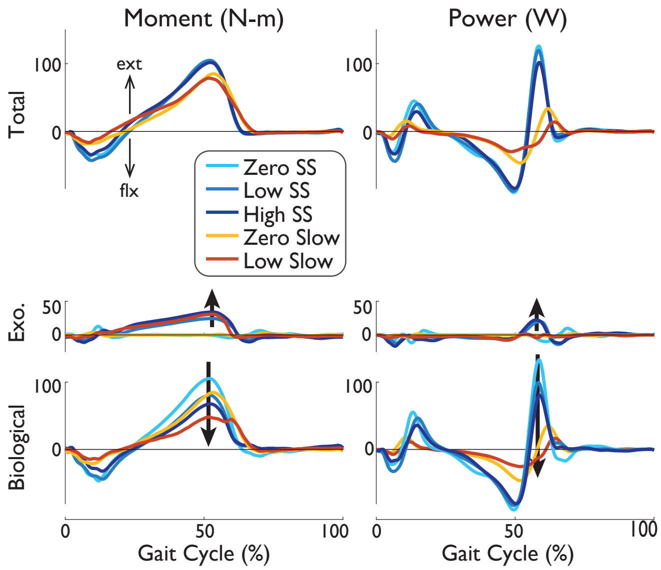

Fig. 6: Total moment and power, as determined by inverse dynamics, measured exoskeleton torque and power, and the difference as the subject's own contribution (Subject B) for a range of torque assistance conditions. With greater assistance, the subject provided less ankle moment and power near push-off.

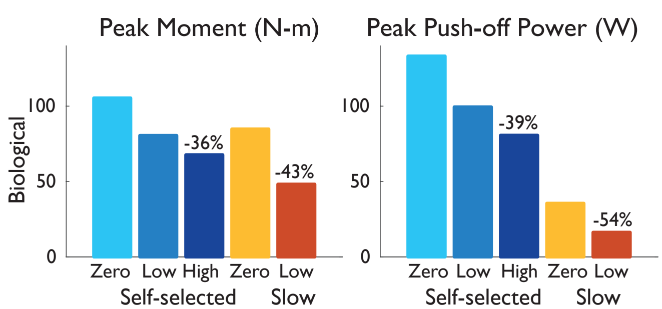

Fig. 7: Peak moment and push-off power produced by Subject B's ankle for different assistance levels. With greater assistance, the biological ankle provided less moment and power near push-off. Percentage decreases compared against Zero are at maximum -43% for moment and -54% for power.

measured.

IV. DISCUSSION

A. The NMC facilitates walking at two different speeds

We had hypothesized that the bio-inspired controller would be intuitive for subjects and be adaptable without the need for extra sensors. To test this hypothesis, we measured healthy subjects walking at two different speeds with the controller on a powered ankle exoskeleton. Preliminary results showed that as the controller provided more torque, the overall cost of walking was reduced for both speed conditions. Reductions were observed in net metabolic cost and in soleus and tibialis anterior muscle activity. The controller also did not adversely affect walking dynamics, as little changes were observed in the ground reaction forces and in step parameters.

While the experiment had a limited number of subjects, the results indicated a reduction in metabolic cost of walking and lowered EMG activity. The implemented controller was also not tuned for specific subjects, for specific speeds, and did not account for the weight of the Achilles device. Hence, the reductions could be even greater than those observed here. In addition, we measured a third subject, whose results were not included as the subject' anthropometry were not fully compatible with the device. However, we also observed less soleus activity with greater torque assistance for that subject. This suggests that the bio-inspired controller proposed could have an inherent capacity to provide patient-specific assistance by automatically adapting to the environmental conditions and subjects' state.

It may be difficult to compare the energetic savings of our controller and the Achilles against other devices and control algorithms, as exoskeleton weight and actuator limitations also impact metabolic cost. One possible measure is the exoskeleton performance index of 0.25 times net metabolic power savings over average exoskeleton positive mechanical power [17], but other measures are needed to evaluate controller robustness or complexity. Here the NMC only needed a minimal set of sensors to reliably and robustly provide human-like ankle power at two different speeds.

B. Neuromuscular controller modeled as an impedance controller

While the modular nature of the NMC design enables control for patient-specific pathologies, controller parameters for subject-specific tuning, such as muscles properties and reflex loops connection weights, are still hard to optimize, mainly due to lack of reliable in-vivo measurement tools. To facilitate subject-specific tunability, we also investigated the modulation of stiffness and damping by comparing the NMC ankle module to an impedance controller.

To facilitate tuning of subject-specific parameters, the NMC was compared against impedance controllers derived from a simulated perturbation experiment. The underlying motivation for using a variable impedance model comes from the fact that the stiffness and damping of muscles can be modified based on their intrinsic mechanical properties and the regulations from central nervous system (e.g. reflexes) [3], [9]. These variable muscle impedances can be transferred to the joint level and estimated using torque and angle measurements in perturbation experiments. Such subject-specific estimation of impedance can help to adapt the NMC to the subjects with different muscle mechanical properties and level of motor deficits.

Figure 8 shows the estimated stiffness and damping of the ankle across the gait cycle. During the stance phase of the gait, the stiffness damping model was a good approximation of the NMC (correlation of estimated torque with simulated torque $>$ 0.9 except at 2.5% and 17.5% of gait cycle). However, the impedance model was not as representative for the swing phase (0.4$<$correlation coefficient$<$0.7).

Comparison of the estimated stiffness and the amount of joint torque both in simulations (the linear fit $R^2 > 0.87$) and in the Achilles experiments (Figure 2) verified the existence of a strong linear relationship. This observation is in in harmony with previous findings in upper limb impedance [20].

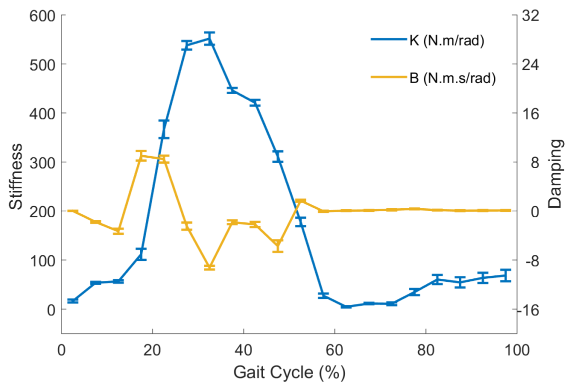

Fig. 8: Estimated stiffness and damping during gait cycle for the KB-model (the expected value and 3 times standard deviation of estimates). The impedance model was obtained through torque perturbations experiments of different amplitudes applied to the ankle to provide a displacement in joint angle. The perturbations produced deviations in joint angle and torque. These deviations were then processed and used to identify the ankle joint impedance. The impedance model fitted to the reflex controller is the following: KB-model with both variable stiffness and damping components ($\tau = -K(t)\cdot(\theta(t)-\theta_t(t))-D(t)\cdot(d\theta(t)-d\theta_t(t))$). The correlation (defined as the Pearson correlation between the response torques produced by the reflex controller and by the model) was used to assess the capability of the model to reproduce the response torque produced by torque perturbation.

V. CONCLUSION

In this paper, we presented a bio-inspired modular controller robust to different walking conditions and requiring very few sensors. The controller we developed is part of a control library for neuromechanical simulation and was tested at the ankle joint level on healthy subjects wearing a powered ankle exoskeleton. The effect of the NMC in stance phase could also be approximated by an impedance controller, which facilitates subject-specific tuning of the NMC.

The similarity between the bio-inspired model and the intact neuromuscular system could also have the potential to improve patient's recovery by adapting the control to the specificity of the patient's remaining functions. We wish to test this further by extending the neuromechanical models of human locomotion to account for specificity of spinal cord injury subjects and test the derived controllers in simulation, on existing devices (e.g. Achilles and gait trainer LOPES [14]), and on a novel exoskeleton currently under development within the framework of the Symbitron project.

ACKNOWLEDGMENT

The work presented here was performed as part of the SYMBITRON project which is supported by EU research program FP7, FET-Proactive initiative "Symbiotic human-machine interaction" (ICT-2013-10) under project contract #611626. The SYMBITRON project is coordinated by University of Twente. The authors thank Annerieke Nijman and Gijs van Oort of University of Twente for their help in conducting this study.

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