A Neuromuscular Model for Symbiotic Man-Machine Exoskeleton Control Accounting for Patient Impairment Specificity

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***Abstract*— Millions of people worldwide live with impaired locomotion. The degree of impairment is highly variable and the causes are multiple. This variation necessitates the design of a new generation of exoskeleton controllers for personalised, symbiotic man-machine interaction. One of the characteristics of such a controller is the ability to realistically include the characteristics of both normal and neurologically impaired human locomotion. The information can be used to recover only the relevant missing features of locomotion. In this paper, we describe the main characterisation tools used to describe human movement and discuss possible ways to include the resulting information in a neuromuscular model in order to create a personalised controller for a wearable exoskeleton.**

I. Introduction

Many neurologically impaired people would benefit from a personalised exoskeleton able to reproduce a walking gait as close as possible to normal human walking. One challenge of such a controller is the ability to recover only the missing torques of the impaired gait without injuring the wearer. In addition to standard safety criteria such as joint angles and torque limits, impairment-specific features should be accounted for. Indeed, impaired gait often exhibits features that act to constrain the locomotion, for example, owing to the increased muscle or joint stiffness frequently observed in stroke and spinal cord injured patients. Such features, if not taken into account, can result in discomfort for the patients or even injury. Therefore, impairment characterisation is a crucial step in the generation of a personalised exoskeleton controller.

One means of characterising the impairment is by examining the net effect of the impairment on muscle contraction. In this context, an impairment can be classified as one of two main categories: remaining locomotor features (RLF) and features constraining the locomotion (FCL). RLF refers to features inducing an overall reduced muscle activity due to muscle weakness and decreased activity level of the spinal networks responsible for locomotion. FCL refers to contracture and the increased overall joint stiffness often present in injury and disease that affect ambulation [10], [15]. FCL constrains the range of motion that can be realised by the limb and the torques that can be applied at the different joints. In order to guarantee safety, the correction of the RLF should only be performed within the constrained range of motion induced by the FCL, which is likely to be markedly different from healthy gait owing to typical gait impairment features.

This paper presents a method for modelling FCL using gait data of spinal cord injured patients. By utilising data from walking trials of a spinal cord injured patient, a model is produced which accounts for the specific features of neurologically impaired gait.

II. Material and Methods

Our proposed strategy to include information regarding the structure of gait impairments within the controller comprises two elements. First, a kinematic and electromyography (EMG) gait analysis of the patient is used to characterise the RLF and FCL components. The impairment characteristics are then used to optimise a constrained gait corresponding to the best possible gait within the bounds of the specific FCL. We propose the use of a neuromuscular walking model since the structure offers a meaningful basis for modelling FCL. Indeed, the advantages of structures such as bio-inspired actuators, sensors, and neural networks make them suitable candidates for modelling impaired locomotion. All such structures can be adapted to model different features. Muscle or tendon impairment can be modelled by adapting the properties of the muscle model, while spinal impairment can be modelled by adapting the structure of the network that generates muscle activation; for example, a stiff joint can be modelled by increasing the basal activity of the motoneurons spanning the joint.

Fig. 1 summarises the different steps in the design of the exoskeleton controller. First, the patient impairment is analysed to extract the FCL. Second, these features are used to optimise a 'constrained neuromuscular model' representing the best possible gait given the specific FCL. Finally, the exoskeleton controller subtracts the contribution of the patient to the movement such that only the missing features are generated. Depending on the application, the contribution of the patient can be either estimated from intramuscular EMG [19] or torque sensors. Both can be directly used in the neuromuscular model.

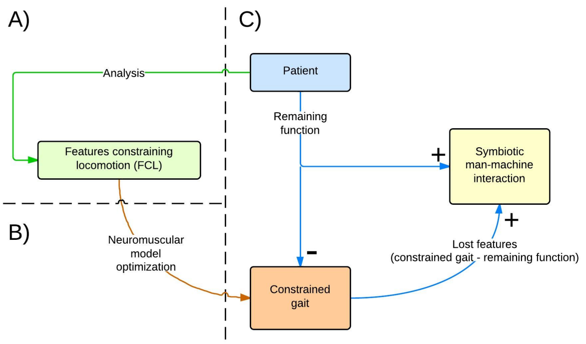

Fig. 1. Exoskeleton controller generation based on the impairment analysis. A) The patient impairment is characterised to extract the FCL. B) A neuromuscular model is optimized to generate a constrained walking gait that exhibits the identified FCL. C) The patient contribution to the movement (i.e. remaining function) is subtracted from the constrained gait to create symbiotic man-machine interaction. The contribution is estimated using EMG or torque sensors.

A. *Spinal cord injury gait impairment characteristics*

Disease involving disruption or weakness of neural pathways such as spinal cord injury has multiple impacts on gait, owing to muscular weakness among other underlying factors. A lack of sufficient foot clearance during the swing phase due to inappropriate knee flexion has been noted in several investigations [9], [11], [16]. Furthermore, excessive plantarflexion of the ankle can occur, leading to foot drag [16]. During the stance phase, the knee is often flexed more in spinal cord injured than healthy subjects, leading to a 'crouched' gait and problems related to the support of the body weight against gravity. The hip joint typically shows greater excursion for spinal cord injured subjects [13]. A less frequently observed deviation from the normal gait kinematics is pelvic drop caused by weakness of the hip abductor muscles [16]. EMG recordings have revealed that the muscle activation of spinal cord injured subjects is different to healthy subjects and tends to have a longer duration[13]; the limb coordination patterns are also different between healthy and neurologically impaired subjects [1], [13].

The various impacts at the hip, knee and ankle joints are manifest in several gait characteristics including reduced walking speed, cadence and stride length of spinal cord injured subjects [8], [12], as well as a longer duration of the double support phase [12], [17]. Moreover, many of the negative impacts on gait are strongly related to the degree of spasticity [8].

B. *Impairment modelling*

The first stage in the creation of a constrained gait model is the choice of parameters characterising the impairment such that the complexity of the problem is reduced. The parameters are then optimised to generate a model exhibiting the FCL. The optimisation algorithm used is a multi-objective extension of the particle swarm optimisation (PSO) algorithm [7] based on lexicographic ordering [2].

The neuromuscular model proposed in this study is an extension [4], [14] of the reflex-based neuromuscular model developed by Geyer [5]. This model was chosen because of the properties of the generated gaits allow muscle activation, joint angle and torque patterns close to human gait while also providing stability to perturbations [3], [18].

Depending on the nature of the impairment, modelling the FCL can be performed at the musculoskeletal or neuronal levels.

- Modelling FCL at the musculoskeletal level can be achieved by modifying the properties of the muscle models using Hill type actuators (details can be found in [6]). Muscle weakness can be modelled through adjustment of the maximum force of the contractile element and changes in muscle passive properties such as increased passive muscle tightness can be modelled by changing the stiffness of the parallel element. Similarly, changes in the tendon properties can be modelled by changing the stiffness of the serial element.
- Modelling FCL at the neuronal level can be done by introducing feedforward signals to the network; these can, depending on the property of the FCL, also be state dependent. The shape of the feedforward signal is parameterised and optimised to reproduce the observed FCL.

C. *Exoskeleton controller*

The exoskeleton controller requires information regarding the remaining function of the patient so that it generates only the torques actually needed. For exoskeleton control using EMG sensors, the correction is directly applied at the muscle stimulation level.

$$A_j = A_j^{model} - A_j^{patient} \tag{1}$$

Here, A_j is the j^{th} muscle activation level, A_j^{model} is the activation level in percentage of maximum muscle activity generated by the model's neural network and $A_j^{patient}$ is the activation level as a percentage of maximum muscle activity generated by the patient.

For exoskeleton control using torque sensors, the correction is applied at the torque level.

$$\tau_i = \tau_i^{model} - \tau_i^{patient} \tag{2}$$

In the above equation, τ_i is the torque applied to joint i, $\tau_i^{model} = \sum_{j\epsilon i} r_{j,i} \cdot F_j$ is the torque generated by the model, where $r_{j,i}$ is the moment arm of muscle j with respect to joint i (for a mono-articular muscle-tendon unit $r_{j,i} = r_i$), and F_j is the actual force generated by the muscle-tendon unit j. $\tau_i^{patient}$ is the torque contribution of the patient.

EMG measurement can be combined with torque sensing by removing the torque contribution of the already included muscle. This can be achieved by having an internal model of the muscle used to estimate the torque contribution of that muscle.

D. Experimental Data

As a first step towards realisation of the previously described control methods, a model is developed in this paper that incorporates some of the specific features of impaired gait. The model is developed using data from treadmill walking of a male spinal cord injured subject aged 64 years with an incomplete C1 injury. The subject's height and weight are 1.86 m and 95 kg, respectively. The subject walked without assistance at a speed of 1 m/s while kinematic data from the hip, knee and ankle joints were collected. The angle ranges for the three joints corresponding to the collected data are shown in table I.

	Hip	Knee	Ankle
Min. (°)	-5.62	61.8	-11.0
Max. (°)	43.6	10.3	13.5

TABLE I
ANGLE RANGES FOR GAIT OF SPINAL CORD INJURED SUBJECT.

A time history plot of the hip, knee and ankle joint angles for the SCI subject and healthy human (in which the healthy gait data is from [20]) is provided in figure 2. We can see small asymmetries between the left and right limbs as a result of the injury. The ranges of the hip angles for both right and left limbs are substantially higher than in the healthy subject (caused by a bending of patient's trunk). Ankle angles show the same trend than hip angles though the differences are larger in stance than in swing. While the knee angles resemble the healthy kinematics, a reduced maximum extension can still be noted. The kinematic data are used to generate the FCL that the constrained gait should fulfil (see Fig. 1). The FCL are modelled here as minimum and maximum ranges of the different joints.

III. Results and Discussion

Fig. 3 shows the resulting hip, knee and ankle angles of the optimised model throughout the gait cycle. It can be seen that the angles fall within the specified angle range constraints; these angle ranges are significantly less than those of the typical gait pattern of an able-bodied person.

The results demonstrate that the applied method can reproduce some aspects of impaired gait. In the particular implementation of this paper, the reduced angle ranges encountered in spinal cord injured gait are reproduced by including additional constraints in the optimisation phase of model development. However, there are other typical gait features of persons with neurological impairment, including reduced joint torques, asymmetry and altered timing of joint kinetics. These aspects will be incorporated in future versions of the gait model by extracting more comprehensive kinematic and kinetic data from walking tests of spinal cord injured subjects.

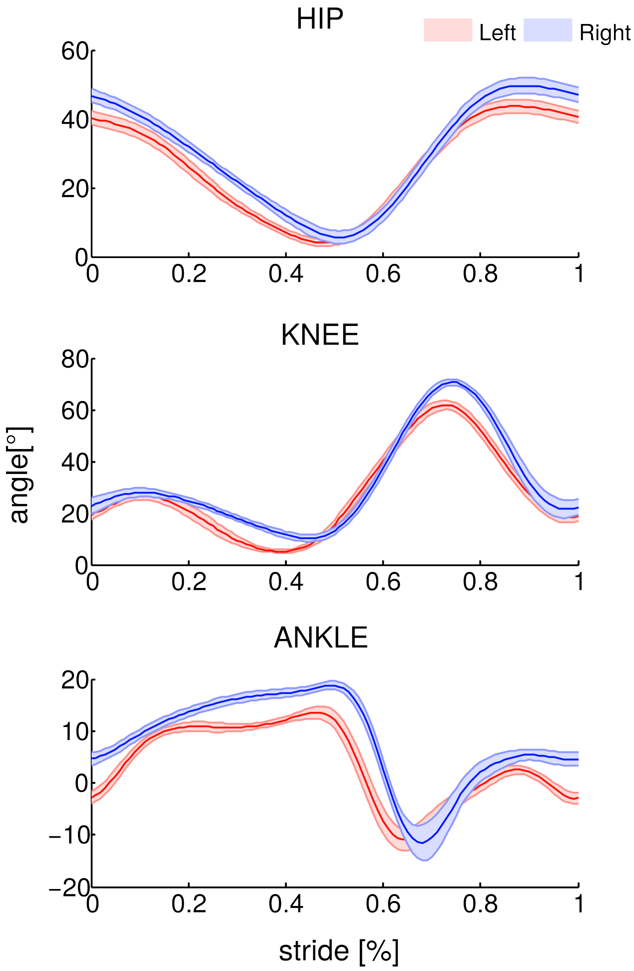

Fig. 2. Averaged time histories of joint angles during gait of the patient walking on a treadmill. Thick lines show the average angles +/- standard deviation. Angles from left/right leg of the patients in red/blue respectively, human data (taken from [20]). Top: hip angles, Middle: knee angles, bottom: ankle angles.

Future work will, therefore, be focused not only on developing more detailed models of spinal cord injured gait, but also on how the models could be integrated within exoskeleton controllers. This realisation must take into account the additional constraints imposed by the spinal cord injured gait characteristics in addition to the interaction between the human and the exoskeleton. The overall goal of the method will be to create personalised controllers for wearable exoskeletons which will allow neurologically impaired poeple to walk in the exoskeleton hardware while also taking into account possible limitations of the specific user such as reduced joint angle ranges, thereby reducing discomfort and the risk of injury.

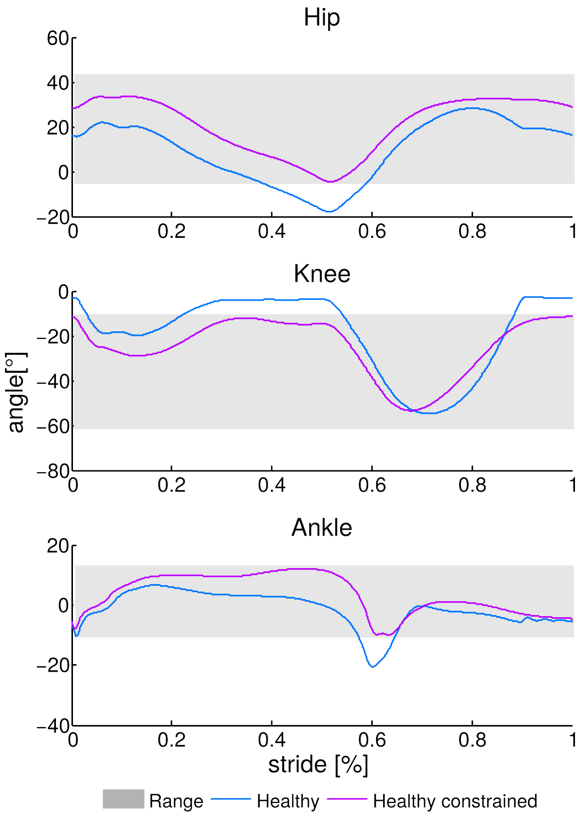

Fig. 3. Averaged time histories of joint angles (as percentage of stride) for healthy model in dashed blue (i.e. no FCL) and healthy constrained model in purple (i.e. with FCL). The FCL, corresponding to results with range constraints extracted from patient kinematics (see Fig. 2), are shown in grey. It can be seen that the healthy constrained gait fulfils the constraints imposed by the FCL.

IV. Conclusions

A method for incorporating specific walking characteristics of spinal cord injured patients into gait models has been presented. In this particular implementation, the focus is on the reduced angle ranges typical of spinal cord injured gait; future realisations will include additional aspects such as asymmetry and reduced maximal joint torques. The presented modelling techique will be used as part of a control strategy aimed at creating personalised control for wearable exoskeletons taking into account possible limitations of the user in terms of joint ranges and other characteristics.

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