Control of posture with FES systems

Zlatko Matjačić, *Kenneth Hunt, *Henrik Gollee, #Thomas Sinkjaer

Institute for Rehabilitation, Republic of Slovenia, Ljubljana, Slovenia
*Centre for Systems and Control, University of Glasgow, Glasgow, United Kingdom
#Center for Sensory – Motor Interaction, Aalborg University, Aalborg, Denmark

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Correspondence address:

Zlatko Matjačić, PhD
Institute for Rehabilitation of Republic of Slovenia
Linhartova 51
SI-1000 Ljubljana
SLOVENIA

Tel.: + 386 1 47 58 159
Fax.: + 386 1 43 72 070

e-mail: zlatko.matjacic@mail.ir-rs.si

Abstract

One of the major obstacles in restoration of functional FES supported standing in paraplegia is the lack of knowledge of a suitable control strategy. The main issue is how to integrate the purposeful actions of the non-paralysed upper body when interacting with the environment while standing, and the actions of the artificial FES control system supporting the paralyzed lower extremities. In this paper we provide a review of our approach to solving this question, which focuses on three inter-related areas: investigations of the basic mechanisms of functional postural responses in neurologically intact subjects; re-training of the residual sensory-motor activities of the upper body in paralyzed individuals; and development of closed-loop FES control systems for support of the paralyzed joints.

Keywords: natural postural control; artificial postural control; neural prosthesis; paraplegia

Introduction

One of the major consequences of an injury to or disease of the central nervous system (CNS) is impaired postural control. The impairment can range from moderate (hemiparesis, paraparesis, tetraparesis) to severe (hemiplegia, paraplegia and tetraplegia). When dealing with moderate impairments the emphasis in the rehabilitation process is on facilitation of the recovery and development of alternative strategies and motor programs in impaired people by means of physiotherapy and various biofeedback training techniques [1, 2]. A prerequisite for application of these techniques is the ability of an impaired individual to stand, i.e. to adequately control all the biomechanical degrees of freedom. This is clearly not the case in paraplegia where the paralyzed joints of the lower extremities and to some extent also paralyzed muscles of the trunk must be supported by external, artificial means. Frequently, functional electrical stimulation (FES), artificially activating skeletal muscles, has been employed for this purpose [3, 4].

The simplest FES system for standing in paraplegia provides bilateral, open-loop stimulation of the knee extensors, which maintains the knees extended [3]. The hips are held hyperextended passively (C-posture) while the ankle joints are free to move. The paraplegic person maintains an upright posture by means of the arms holding on to a suitable support and thus effectively acts as the postural controller. This type of FES supported standing is clinically widespread. However, two fundamental limitations are associated with it. The first is the problem of muscle fatigue, which limits the achievable duration of standing. Posture switching technique [5] and closed-loop control of knee extension with conventional PID controllers [4] and artificial reflex controllers [6] were proposed and with some success experimentally tested to circumvent the fatigue of artificially activated muscles. The more fundamental second limitation, directly related to the existing control of posture, is that the arms of the paralyzed individual are engaged in stabilizing activity. This makes standing nonfunctional and consequently renders FES supported standing exercise, which has many beneficial therapeutic and physiological effects [7], unattractive for a paralyzed person. Further, as regulation of posture during standing is an integral part of movement, standing should be bipedal and not quadripedal. These deficiencies motivated several research groups to work on the problem of restoring functional, i.e. “unsupported” or “arm-free” standing in paraplegia. Here, the term functional standing implies that at least one arm is freed from balancing activity and available for manipulation of objects and purposeful interaction with the environment [8].

The first theoretical study was undertaken by Jaeger [9] who developed the simplest mathematical model of arm-free standing, i.e. the single-link inverted pendulum model. The only degree of freedom was the ankle joint, assumed to be under the control of closed loop FES via stimulation of calf and dorsiflexor muscles. A simple PD (Proportional-Derivative) controller was used in simulation studies, which showed that at least theoretically it might be possible to stabilize the body.

In an experimental study of "single-link standing" Hunt et al have achieved successful periods of unsupported standing with paraplegic subjects [10-12]. For reasons discussed in detail below, it was found that the simple PD structure is not sufficient for stable postural control. The control approach in [10-12] is based upon a nested loop structure, as proposed by Donaldson [13]. An outer loop is concerned with stabilization of the inverted pendulum, while the inner loop enhances the ankle torque tracking. The results of this work have demonstrated experimentally how challenging the task of restoration of arm-free standing in paraplegia is when the control of posture relies solely on the artificial FES control system. The scope of the achievable performance and limitations to applicability of the approach are discussed below. One key conclusion of the work, however, is that it is important to integrate the voluntary motor control skills of the upper body within the overall control scheme for posture stabilisation.

From the results of the above research it became clear that the actions of the neurologically intact neuromuscular system of the upper body need to be incorporated in the control schemes in order to synthesize functional postural control during restored standing in paraplegia [14]. The first control scheme which aims for functional FES supported standing in paraplegia by attempting to integrate the actions of the upper body and the artificial system was proposed by Matjačić and Bajd [15, 16]. The control strategy is based on voluntary and reflex activity of the non-paralyzed upper body and artificially controlled mechanical stiffness in the paralyzed ankles. The paralyzed knees and hips were assumed to be maintained in the extended positions by means of FES. Analysis of the linearized closed-loop mathematical model has shown that with properly selected ankle stiffness postural stability could easily be maintained by the so-called "hip strategy" [17], except that the postural activity of the paralyzed hips is replaced by trunk movement. The approach, confined to the sagittal plane, was tested experimentally by using a custom made mechanical brace that held ankles, knees and hips of a paralyzed individual (T12) immobilized and provided an artificial servo-controlled ankle joint where a selected level of mechanical stiffness was maintained. After a few days of training the tested subject learned how to use the upper body to maintain a selected posture with adequate stiffness support from the mechanical brace. However, it was initially unclear whether such an approach would be viable should FES be used to support the knees and hips in extended positions and to regulate adequate stiffness in the ankles. A recent pilot study implemented ankle stiffness control using FES, and this allowed a T5 paraplegic subject to maintain stable posture [18]. Further details are given below.

The approach described in [15, 16] was an engineering attempt, based on biomechanical analysis of single and double inverted pendulum stabilization requirements, addressing several crucial questions related to integration of the actions of the intact physiological system (upper body) with the actions of the FES system supporting the paralyzed physiological system (lower extremities). These questions are:

- How can the user be in continuous control of posture regulation?
- What sensory feedback should be provided to the user and the artificial control system?
- What should be the role of the artificial control system in the whole control scheme? Should the artificial controller be concerned only with actuation issues, i.e. the regulation of specified variables at the level of a single joint leaving the overall postural activity to the "re-trained" upper body, or should it try to minimize the upper body effort or fatigue in stimulated muscles?
- Should the upper, non-paralyzed part of the body be "re-trained" to be able to continuously regulate selected posture, what kind of training should the user be subjected to and how can this training take the actions of the artificial control system into account and, conversely, what should be the actions of the artificial control system in order to always work in synergy with the re-trained central nervous system (CNS)?

This paper is composed of two parts. In the first part we review our work carried out in the past three years addressing the questions stated above. Our methodology was based on investigation and identification of functional postural control following controlled delivery of perturbations in neurologically intact individuals in conditions that resemble functional paraplegic standing. Our aim was to gain an insight into control strategies developed by a biological system, which subsequently served as a basis for the synthesis of a life-like control scheme suitable for restoration of functional FES supported control of posture in paraplegia. The findings of these basic studies provided directions towards efficient "re-training" methodologies and the scope of the control scheme for the artificial FES system supporting the paralyzed joints of lower extremities. Innovative control algorithms for the implementation of the closed-loop FES support were developed and tested. Finally, we review the experimental demonstration of the feasibility of the proposed approach, combining the results of the studies into i) basic biomechanical mechanisms underlying balance control in neurologically intact individuals, ii) re-training of the residual sensory-motor abilities of a paraplegic and iii) low-level, closed-loop FES control algorithms. The second part of the paper discusses our findings and the proposed control strategy in the light of natural control of movement, advantages and limitations of the proposed approach and instrumentation requirements necessary for implementation in a clinical environment. Further, robustness and stability issues of the overall postural control scheme in the face of fatigue of artificially activated muscles and purposeful manual interactions with the environment are discussed.

Postural control – studies into basic mechanisms

Our first step in deriving a suitable control strategy for restoration of functional postural control in paraplegia was to study responses in a group of neurologically intact individuals under the relevant experimental conditions. Specifically, we were interested in identification of functional postural responses by studying the net mechanical outcome in particular joints after the action of mechanical perturbations. Unlike the vast majority of studies done in the area of postural control, where moving or rotating platforms upon which an examinee was standing imposed perturbations, we developed a new mechanical apparatus called the Multi-purpose Rehabilitation Frame (MRF), shown in Fig. 1a, which imposes perturbations at the level of the pelvis [19]. In this way, after the commencement of perturbation, the body is inevitably "broken" into two parts that are accelerated in the opposite directions, very much like a mechanical double inverted pendulum. The subsequent recovery of vertical posture requires coordinated activity of the sensory-motor apparatus of the upper and the lower body. The question arises as to what is the interaction of the two postural activities. Is the whole body governed by one "central" controller, which estimates the location of the center of mass (COM) and center of pressure (COP) and does the difference of the two signals drive the recovery and necessary functional activity [20, 21]? Or can functional postural activity be decoupled between the actions of the upper and the lower body? As we have pointed out in the Introduction we are particularly interested in the functional responses of the lower extremities, as we assume that the intact upper body of a paraplegic person can be adequately re-trained. If we further assume that the knees and hips (sagittal plane) of a paraplegic person will always need to be held extended, the focus is placed upon the functional postural responses in the ankles and the hips (frontal plane).

Figure 1

We have examined these issues by undertaking a study where we assessed functional postural responses by analyzing the net joint torques (NJT) in the ankles and the hips resulting from perturbations delivered in multiple directions to subjects standing quietly [22]. A total of eight subjects were standing on two force platforms while the MRF apparatus randomly delivered controlled perturbations (torque pulses; duration 200 ms) at the level of the pelvis in eight directions: anterio-posterior (AP), medio-lateral (ML), and four combinations of these principal directions (Fig. 1a, b). Perturbations were repeated five times in each direction for six conditions (i.e., three different perturbation strengths and three different feet orientations). NJT in each ankle (AP and ML projections) and hip joint (ML projection) were calculated. We have examined time courses of individual responses in each joint as well as the sums of NJT that act in AP (both ankle joints) and ML directions (both ankles and hips). The comparison of the averaged ankle sum NJT (AP) responses showed that the time courses of the responses elicited by a perturbation acting only in the AP (Forward, Backward) direction were identical to those elicited by a combination of two corresponding AP and ML perturbations (Forward group, Backward group), as shown in Fig 2.a. In contrast the observed averaged ankle NJT (ML) responses did not show the same similarity (Fig. 2b). Comparison of the averaged ankle and hip sum NJT (ML) responses revealed that the time courses of the responses elicited by a perturbation acting only in the ML (Left, Right) direction were identical to those elicited by a combination of two corresponding AP and ML perturbations (Left group, Right group), as shown in Fig. 2c. These findings were invariant of the experimental conditions and were consistent among all eight subjects. Thus, we concluded that the ankle sum NJT (AP) and the ankle and hip sum NJT (ML) were the global variables being controlled during the recovery of the lower body from perturbations. These results show that the CNS controls the recovery from the multiple-direction perturbations of moderate strength by decoupling the AP-ML postural space into two orthogonal directions (AP and ML). Furthermore, we can also conclude that the control of the upper body (predominantly trunk muscles) and the described control of the ankles and hips of the lower body is decoupled as well, since despite changes in perturbation strength and orientation of the feet (change of the sensory-motor map), the described control law, governing the recovery of the lower body to vertical posture, did not change.

Figure 2

Another very important observation was that the time courses of the calculated NJT and the corresponding inclination angles (ankles – sagittal plane, ankles and hips – frontal plane) bore close resemblance, indicating the possibility of a rather simple and close relationship between the kinematics and kinetics of the recorded postural responses. This observation motivated another study [23] where the procedures employed in [22] were repeated in a group of six neurologically intact individuals with mechanically locked knees by means of long leg bracing. The objective was to investigate whether a simple static stiffness model could adequately relate the angles and NJT developed in the ankles and hips that constituted postural responses following a series of random perturbations applied at the hip by means of the MRF. In a similar way to the previous study we examined the responses in each individual joint as well as summed responses. Ankle sum stiffness was found to be invariant to the perturbation directions for the group of forward and backward directed perturbations while the hip sum stiffness was invariant to left and right directed perturbations. The correlation coefficients of the linear regression model were in all cases higher than 0.95 indicating the suitability of the simple linear model. Even though the reviewed relationships were valid only for the summed responses, both studies [22, 23] showed that the individual contribution to the summed responses either in the ankles (AP) or ankles and hips (ML) varies approximately linearly with the relative loading of the particular extremity. A similar study by Mihelj et al. [24] investigated the validity of a static stiffness model in recovery from perturbation imposed only in the sagittal plane. Their results were similar to those in [23].

These findings have direct implications for restoration of functional arm-free standing in paraplegia and provide an answer to the most important question: How to re-train the residual sensory-motor system (essentially the upper body) in a way which will be compatible with the action of the neuroprostheses (FES supported lower body) and what should be the appropriate action of the neuroprostheses in order to work in synergy with the re-trained CNS. The answer and consequently the control strategy are rather simple and straightforward. The upper body should be re-trained in conditions where the MRF emulates the action of the neuroprostheses, i.e. the level of supporting forces around the pelvis should be proportional to a suitable stiffness constant. On the other hand the action of the neuroprostheses should be limited only to i) maintain the knees and hips (sagittal plane) in the extended positions by means of FES of the knee and hip extensors and ii) to provide stiffness around the ankles in the sagittal plane and around hips in the frontal plane. The ankle stiffness should be controlled by closed-loop FES of the dorsi-/plantarflexors while the stiffness around the hips in the frontal plane should be controlled by closed-loop FES of the abductor muscles. The reference values for left and right ankles should depend on relative limb loading, while the left or right leg abductors should be stimulated in dependence on the inclination of the legs in the frontal plane. Note that in the proposed strategy the actions of the re-trained upper body, which is under the voluntary and reflex control of the re-trained CNS, are completely independent of the actions of the artificial control system and vice versa. Naturally, upper and lower body are coupled mechanically and thereby influence motion of each other, however, since the action of the artificial control system is limited only to regulation of impedance in the paralyzed ankles and hips both control systems will always act in synergy toward the same objective - maintenance of upright posture. Even though the proposed scheme is conceptually rather simple its implementation is a very challenging task.

Postural control – re-training of the intact sensory-motor apparatus

The first step toward the implementation of life-like postural control in paraplegia is appropriate training of residual sensory-motor abilities during standing in an environment that is fall-safe and adequately emulates the action of an artificial FES control system. We have investigated whether the MRF apparatus, when configured to provide a selected static stiffness support, transferred to supporting forces acting on the pelvis of a standing subject through the bracing frame, can serve as a suitable training environment facilitating the development of suitable postural activity of the non-paralyzed upper body [25]. Two complete paraplegic (T6 and T8) and two incomplete tetraplegic (C5-6, C5-6) subjects participated in a 9-day balance-training program. Every day three consecutive standing sessions were performed. The duration of each session was approximately 5 minutes. Before initiating each training session, the stiffness support of the MRF apparatus was varied in order to determine the level at which the subject was comfortable and able to maintain vertical posture in the sagittal and frontal planes. Each subject effectively determined a suitable level and was therefore in charge of the training sessions. Both paraplegic subjects supported their trunks by holding onto the bracing system of the MRF due to rather high lesions. This, however, did not simplify the task of balancing by means of the upper body. At the beginning of the program the initial levels of stiffness support provided by the MRF was around 15 Nm/degree for all four subjects. In the course of the following three days both paraplegic subjects were able to maintain balance at almost half this value, i.e. around 7-8 Nm/degree in both planes of motion, while both incomplete tetraplegics were able to balance without the MRF's support. The results of this study clearly demonstrated the ability of the residual sensory-motor system to re-learn and re-train the abilities necessary for maintenance of posture when the lower body is supported in a stiffness-like manner.

Postural control – innovative closed-loop FES control algorithms

The work which was reviewed above shows that ankle stiffness plays a key role in control of postural stability *in the situation where control action can be applied at both the ankle and at the hip joint*. On the one hand, the studies of the basic mechanisms of postural control show that the ankle response to postural perturbations can be approximated with a high degree of accuracy as a static stiffness. Secondly, the preliminary results on re-training of the intact sensory-motor control show that if an appropriate degree of stiffness is provided artificially at the ankle joint, then paralysed individuals are able to balance using their intact upper body. In this section we review results which complement these findings - our focus is on the development and evaluation of closed-loop FES control systems which aim to provide suitable ankle characteristics, in order to achieve stable unsupported (arm-free) standing in paralysed persons.

It is important at the outset to recognise that the structure of the control problem is a primary determinant of the ankle properties required for stability. The discussion in the preceding two sections considered the case where the subject can initiate control activity around the hip joint, as well as at the ankle joint (as realised in the MRF apparatus). In this setup the subject can be modelled ideally as a two-link inverted pendulum. We will consider closed-loop FES control for this situation below.

However, we begin by reviewing our work on a simpler case, i.e. the situation in which the subject is free to move only around the ankle joint. Experimentally, this has been achieved using an apparatus known as the "Wobbler", which braces all joints above the ankles. This is modelled in the ideal case as a single-link inverted pendulum. It may at first appear paradoxical, but this simpler bio-mechanical structure requires a more complex *artificial* control algorithm than the two-link case. This is because a simple static stiffness is not sufficient for stabilisation of a single-link pendulum. As well as a minimal level of stiffness (whose value depends on the bio-mechanical parameters of the system), a certain level of viscous damping is also required. It is well known that, at least theoretically, stiffness and damping can be realised for the single-link model of standing using a simple PD (Proportional-Derivative) controller which acts on the measured angle of inclination (this was illustrated in simulation by Jaeger [9]). Unfortunately, a PD control strategy is not appropriate for a real *implementation* of artificial postural controllers. A physical implementation requires measurement of the inclination angle, which introduces measurement noise into the feedback loop. Pure derivative action then leads to amplification of the noise at high frequencies, and this readily results in system instability. An important constraint in the control of standing is that the system is open-loop unstable, and therefore a lower limit on the closed-loop bandwidth exists for stability. Sufficient bandwidth is also required in order to reject postural perturbation disturbances. Thus, as a result of extensive experimentation with human subjects, we have found that a fully dynamic, higher-order, controller is required in order to provide sufficient flexibility to achieve the unavoidable trade-offs in closed-loop frequency responses for good disturbance rejection and low noise sensitivity, while maintaining a closed-loop bandwidth high enough to maintain stability. We consider the single-link and two-link cases individually.

{xe "Closed-loop FES posture control\: single-link case"}Closed-loop FES posture control: single-link case

We have previously completed an experimental study on control of unsupported standing in paraplegia. The study used an apparatus called the "Wobbler" in which the standing subject is free to move only around the ankle joint, and stabilising torque is generated by Functional Electrical Stimulation (FES) of the calf muscles. The Wobbler apparatus is shown in figure 3, and it is described in detail by Donaldson *et al* [26]. While standing in the Wobbler the subject wears a custom-fitted body shell which locks the knee and hip joints, allowing motion only around the ankle joint (figure 3).

Figure 3

For safety four light ropes are attached to the shoulders of the body brace and from there to a frame attached to the ceiling. When the ropes are tight the body cannot move. The ropes can be slackened sufficiently to allow movement in the sagittal plane within predefined limits. A string attached to the body brace at shoulder level is wound round a pulley attached to a potentiometer placed well behind the subject. This potentiometer is used to measure the inclination angle. The subject's feet are positioned in foot boxes connected to a shaft aligned with the ankle axis. Sensors in the shaft allow independent measurement of left and right ankle moments. Measurement of inclination angle and ankle moments allowed us to implement a nested loop structure for control of standing (see figure 4): a high-bandwidth inner loop provides control of the ankle moments via stimulation of the calf muscles; the angle controller in the outer loop regulates the inclination angle, and its control signal is the desired ankle moment for the inner loop.

Figure 4

The nested-loop structure for unsupported standing allows the overall feedback control system to be designed and tested in several steps, starting with the ankle moment control loop and moving then to the body angle controller. The steps involved in system design and test are:

1. The muscle dynamics are identified using an open-loop PRBS (pseudo-random binary sequence) test. This establishes a dynamic model between the pulsewidth p and the ankle moment m. This step also involves validation of the identified models.

2. The closed-loop controller for ankle moment is designed. The moment controller is designed using an analytical approach which utilises the approximate dynamic model identified in the previous step. This step establishes a desired closed-loop response between the reference moment m_{ref} and the measured moment m. Following controller synthesis, the moment loop is verified by examining the key closed-loop

frequency responses and then by testing the performance in experiments. When these tests are judged to be satisfactory we proceed to the next step.

3. The closed-loop controller for body inclination angle is designed. The plant model for angle controller design is taken as the transfer function between the desired moment m_{ref} and the angle θ, i.e. this is a combination of the ankle moment loop and the open-loop body dynamics. Angle controller design establishes a desired closed-loop response between the reference angle θ_{ref} and the measured angle θ. The frequency response functions of the overall closed loop are verified, and then the system is tested in experiments.

Details of the approach used for muscle dynamics identification are given elsewhere [27, 28]. For design of the moment and angle control loops both optimal control [10, 11, 27] and pole assignment approaches [29] have been used.

Preliminary results showed that periods of stable standing for 30-40 s were possible [10, 11]. The feedback control structure was subsequently re-designed and improved [29, 30], and paraplegic subjects are now able to stand for up to seven minutes at a time [12]. Results of a typical standing experiment are shown in figure 5. Here, the top plot shows the stimulation pulsewidth p, the middle plot depicts the total measured ankle moment m and the reference moment m_{ref}, and the bottom plot shows the inclination angle θ together with the reference angle θ_{ref}. A constant reference angle was set while external disturbances were applied by pulling anteriorly at chest level with a moment of approximately 6 Nm. The disturbance is applied at 5 s and at 25 s, each time for a period of 10 s. In both instances, the disturbance is compensated for by an increase in the stimulation pulsewidth which causes an increase of the ankle moment.

Figure 5

Our study concluded: (i) that significant periods of standing can be achieved, and that the length of standing is limited only by muscle strength and the rate of fatigue; (ii) that the ability to reject external disturbances (such as pulling or pushing the subject) is defined by the available muscle strength, which is limited, and (iii) most importantly, that further work must integrate the voluntary motor skills of the upper body within the overall control scheme for posture stabilisation. This major step forward will lead towards true functional standing, with greatly improved standing times, and will offer enhanced and flexible therapy options.

{xe "Closed-loop FES posture control\: two-link case"}Closed-loop FES posture control: two-link case

We have recently completed a pilot study which addresses the problem of posture control in the case when the voluntary motor skills of the upper body are to be integrated with an artificial closed-loop FES control system. The feasibility of ankle stiffness control via FES was first proven [31, 32] (that work used the Wobbler apparatus). To investigate control of standing via FES-controlled ankle stiffness we then utilised the MRF apparatus described previously. Thus, the upper body is free to move, and the lower limbs are controlled via FES to give a sufficiently high ankle stiffness. Preliminary standing experiments were carried out with one paraplegic subject with a complete lesion at level T5. With FES-controlled ankle stiffness, and voluntary motor control action from the upper body, the subject was able to stand repeatedly for periods of one minute [18], after which stimulation was switched off and postural stability was immediately lost.

The MRF's frame was used to brace the subject's knee and hip joints and to constrain motion to the sagittal plane. Thus, the upper half of the body was under voluntary control, while the ankle joint was controlled using closed-loop FES. In particular, the dorsiflexor and plantarflexor muscle groups of both legs were stimulated in an attempt to achieve a desired level of ankle stiffness. The total ankle moment was measured using a forceplate, and the ankle angle was measured using sensors on the MRF frame.

The control strategy is depicted in figure 6. Here, the required total ankle moment $m_{ref,total}$ is the product of measured ankle angle and desired stiffness. This total required moment is then distributed between the left and right sides using a simple load balancing approach. The reference moments for the left and right sides are then fed to individual closed-loop moment controllers for each side, i.e. the blocks labelled "left ankle" and "right ankle" in figure 6 are each dynamic closed-loop moment controllers. The design methodology for each moment controller is similar to that outlined above for ankle moment control with the Wobbler apparatus, i.e. the approximate linearised dynamics of each ankle joint are determined in an open-loop identification test, and the models so obtained are then used to design closed-loop controllers (the pole assignment procedure was used in the pilot study).

A typical experimental test of standing using this approach is shown in figure 7. In this test, the desired stiffness level was set to 10 $Nm \cdot deg^{-1}$ (this value was chosen based on data from intact subjects from previous studies [15,16]). The left column of the figure shows data for the left leg and the right column shows the right leg. The top two rows show the stimulation pulsewidths for the plantarflexor and dorsiflexor muscles, respectively. The plots in the third row show desired and actual left and right ankle moments (recall that the desired moment is the product of ankle angle and the pre-specified desired stiffness). The bottom plot (identical for left and right sides) shows the ankle angle. It is clear that the subject not only maintains upright posture during this test, but that he also manages to progressively decrease the excursions of the sway, indicating a learning effect. These results are typical of many other successful trials with this subject.

Figure 6

Figure 7

Discussion

<u>*Life-like restoration of postural control vs. natural control of movement*</u>

Different parallel control systems are involved in the regulation of a single limb movement, which is an example of a seemingly simple motor task, as well as in regulation of bipedal standing, which requires carefully orchestrated sensory-motor activity of the whole body musculo-tendon systems. These parallel systems range from single joint to whole limb impedance regulation accomplished through activation of antagonist muscles [33], followed by peripherally mediated reflex mechanisms [34] and finally concluded by centrally mediated responses. Posture regulation during standing is a typical motor task where, unless required for execution of a manipulation and interaction of the upper limbs with the environment, no planned movement is taking place. Rather, deviations caused by internal or external disturbances are continuously counteracted in ways that are not yet completely understood. There is experimental evidence suggesting that postural activity during quiet stance [35] and perturbed standing [23, 24] can be adequately modeled as regulation of a suitable level of mechanical impedance, where the stiffness or compliant component plays the major role.

Despite some doubts [36], these findings and suggestions do not come as a surprise if one considers the prohibitive computational complexity needed for specific trajectory control of a multi-link inverted pendulum, requiring substantial central activity. If we further take into account the magnitude of neural delays within the neuro-musculo-tendon systems, centrally driven regulation of posture would encounter serious difficulties when attempting to control deviations from the desired posture. In this respect it is easy to imagine the efficacy of the properly tuned impedance regulation acting in a decentralized manner, locally in each joint of the biomechanical system. One can intuitively imagine that a mechanical model resembling the bipedal structure of a human and having a simple mechanical spring mounted in each mechanical degree of freedom would be able to withstand external perturbations applied at different segments without tipping over the feet. However, at the same time one has to acknowledge that such a passive solution suffices only for limited magnitudes of perturbations. As soon as this threshold is exceeded, the structure would lose balance. Therefore, the described regulation of posture applies only when counteracting the effects of perturbations of moderate size. As soon as these are exceeded, a different strategy must be adopted, which does not utilize impedance regulation, but rather makes use of mechanical dynamic coupling, as in the "hip" control strategy [17]. What has been discussed so far applies for the early and medium latency responses. However, in the later, conclusive phase of the postural response, voluntary activity is needed. Therefore, the simple impedance regulation approach must be complemented with central mechanisms providing decision making capacity and acting as an adaptive, fine-tuning higher level of control.

In the light of the previous paragraph the proposed life-like control strategy for restoration of functional regulation of posture during standing in paraplegia incorporates all the important aspects of natural, biological control. The lower body is supported by means of artificial impedance regulation in the ankle and hip joints, while the upper body is re-trained in a manner which provides the fine-tuning and decision making capabilities to the whole postural control scheme. In this way the task posed for the artificial control system is rather simple, i.e. decoupled regulation of stiffness (ankles, hips) and position (knees, hips).

There is a very close interplay between the studies reported above which investigated the basic mechanisms of *natural* posture control, and those which seek to design *artificial* systems for control of posture in impaired individuals. Results from the former give simple approximate models of ankle and hip control during standing, and these provide design guidelines for the latter.

We emphasise, however, that the control strategy in an artificial design will depend on the task in hand. For stable "single-link standing", for example, a static stiffness control is not sufficient. We have nevertheless shown that several minutes of quiet standing can be achieved, even in the face of postural disturbances, while the arms and upper body are not involved in the posture stabilisation task. In this case the arms and hands are free to concentrate on some functional task.

In "two-link standing", the upper body is actively concerned with maintenance of balance, and the control strategy at the ankle can then be simpler, i.e. static stiffness control. This kind of standing is qualitatively different, and is highly useful for tasks such as re-training of balance in a range of patient populations. Further, such a control scheme inherently incorporates the “posture switching” technique, as the user can voluntarily change posture, thereby relaxing and engaging different muscle groups. Finally, the stability of the artificial control system is not compromised when the user manually interacts with environment as the actions of the FES system are essentially “passive”.

Advantages and limitations of the proposed artificial control system

The closed-loop FES control approaches we have presented require a simple approximate model of ankle dynamics. This is obtained in a fast and simple identification procedure. It is important to note that although a model is required, it does not necessarily have to be of high accuracy because, ultimately, it is used for the purpose of feedback control design. In fact, the basic properties of feedback mean that the model can be highly inaccurate at low and high frequencies. This is because feedback controllers are normally designed with integral action, resulting in infinite gain at zero frequency, and hence large robustness against model uncertainty at low frequencies (such as constant offsets and disturbances). At high frequencies, on the other hand, the plant gain will naturally go to zero, and the loop is thus protected against noise and model uncertainty at high frequency. The critical area is the crossover region, i.e. for frequencies close to the closed-loop bandwidth. Here, the model needs to be sufficiently accurate to ensure robustness of stability and performance against model error.

Thus, we strongly recommend analytical control design procedures based on simple linear models, rather than heuristic approaches such as trial and error tuning of PID controllers. Simple empirical models can be obtained very easily, and can lead to large benefits for achievement of desired closed-loop properties.

Limitations to the application of the closed-loop FES control approach may arise in the paraplegic population due to secondary medical complications including low muscle tone, excess spasticity, and joint contractures. Furthermore, limitations may arise because of muscle weakness and fatigue.

It is not obvious that adaptive control is a natural solution in the context of standing control. In standing, we expect the loop gain to decrease progressively because of fatigue. Tracking gain changes and adapting the artificial controller to maintain a given closed-loop response will conversely require the controller gain to be progressively increased, thus making the system more sensitive to measurement noise and to model error. Also, adaptive control usually requires sufficient excitation of the system which can be difficult to ensure in a quiet standing situation. Thus, adaptive control must be applied with caution. The overall control scheme, however, incorporates adaptive control, via the upper body, in the regulation of posture. This resides in the actions of the re-trained residual sensory-motor system.

Future steps

Based on extensive experimental work and tests with paralysed individuals we are confident that the proposed control approach, which is based on the re-trained postural abilities combined with simple artificial control algorithms for FES support, is the most appropriate and might become successful in the future, once the technological limitations are overcome. Namely, even though the control principles underlying the proposed control scheme for the artificial FES system are simple, decoupled and straightforward we need to recognize the difficulties associated with reliable implementation. One has to acknowledge that the successfully accomplished experimental tests were performed in laboratory conditions, where the sensory information required for stiffness control was accurate and reliable. Even though the proposed control scheme incorporates adaptive capabilities (in the form of the re-trained upper body) that can compensate for less than optimal performance of the artificial control system, there is clearly a lower bound on this performance, which is still acceptable.

At the moment there are no artificial or natural sensory systems at our disposal, of acceptable performance and reliability that could be used in the implementation of the proposed control scheme for practical standing. The development of suitable sensory systems, being entirely artificial or tapping directly into signals produced by the physiological sensors, is an active area of research [37]. It is also likely that the future technological solution that might implement the findings reviewed in this paper will have to be fully implantable. In this respect we mention the BION^{TM} system for distributed neural prosthetic interfaces [38], which is capable of delivering precise stimulation and picking-up a variety of sensory signals within the body, as a possible technological platform for the future implementation of the proposed control scheme for posture control with FES.

Before such an attempt can take place we first need to test the viability of the proposed control scheme in its entirety, as for the time being only artificial systems for ankle control have been designed and tested. We also need to test stiffness control of the hips in the frontal plane, as well as maintenance of the knees and hips in the extended positions in the sagittal plane. Finally, the minimum quality of the required sensory information needs to be assessed experimentally in order to provide the design specifications of the future implantable neuroprosthesis for functional standing in paraplegia.

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Figure captions

Fig. 1. Perturbing apparatus and perturbation directions. **a)** The Multi-purpose Rehabilitation frame (MRF) was used to generate perturbations to a standing subject. It consists of two 2-DOF rotational joints, two 1-DOF rotational joints, two vertical supportive rods and a bracing system. Subjects stood with each foot on an aluminum block (the top surface contains a grid of 480 symmetrically drilled holes). Cylindrical pegs were used to constrain the position and orientation of feet. Torque impulses, delivered by two servo-controlled hydraulic motors through the bracing system, which was put around subject's pelvis were used to induce perturbations. **b)** Perturbations were delivered in the eight directions: four principal (Forward and Backward in the AP direction and Left and Right in the ML directions) and four combinations of the principal directions (Forward & Left, Forward & Right, Backward & Left and Backward & Right). **c)** NJT conventions are illustrated.

Fig. 2. The representative NJT responses (one subject). An arrow shows a time instant of perturbation commencement. **a)** The ankle NJT (AP) responses are shown for each perturbation direction. At the right hand side the ankle sum NJT (AP) are plotted together for the perturbation directions: Forward, Forward & Left, Forward & Right (upper row) and Backward, Backward & Left, Backward & Right (lower row). **b)** The ankle NJT (ML) responses are shown for each perturbation direction. **c)** The hip NJT (ML) responses are shown for each perturbation direction. At the right hand side the ankle and hip sum NJT (ML) are plotted together for the perturbation directions: Left, Forward & Left, Backward & Left (upper row) and Right, Forward & Right, Backward & Right (lower row).

Fig. 3. Subject standing in the Wobbler apparatus.

Fig. 4. Nested loop control structure. θ is the inclination angle, m is the ankle moment and p the pulsewidth of the stimulation. C_m is the moment controller and C_θ is the angle controller. The desired values for ankle moment and inclination angle are m_{ref} and θ_{ref}, respectively.

Fig. 5. Wobbler standing result. Solid lines indicate measured values, dashed lines are reference values. See text for details.

Fig. 6. Block diagram of the ankle stiffness control and standing strategy. The blocks denoted “left ankle” and “right ankle” are closed-loop controllers for the left and right ankle moment, respectively.

Fig. 7. Standing control, T5 paraplegic subject. The decreasing amplitude of the sway angle θ_S indicates a learning effect.

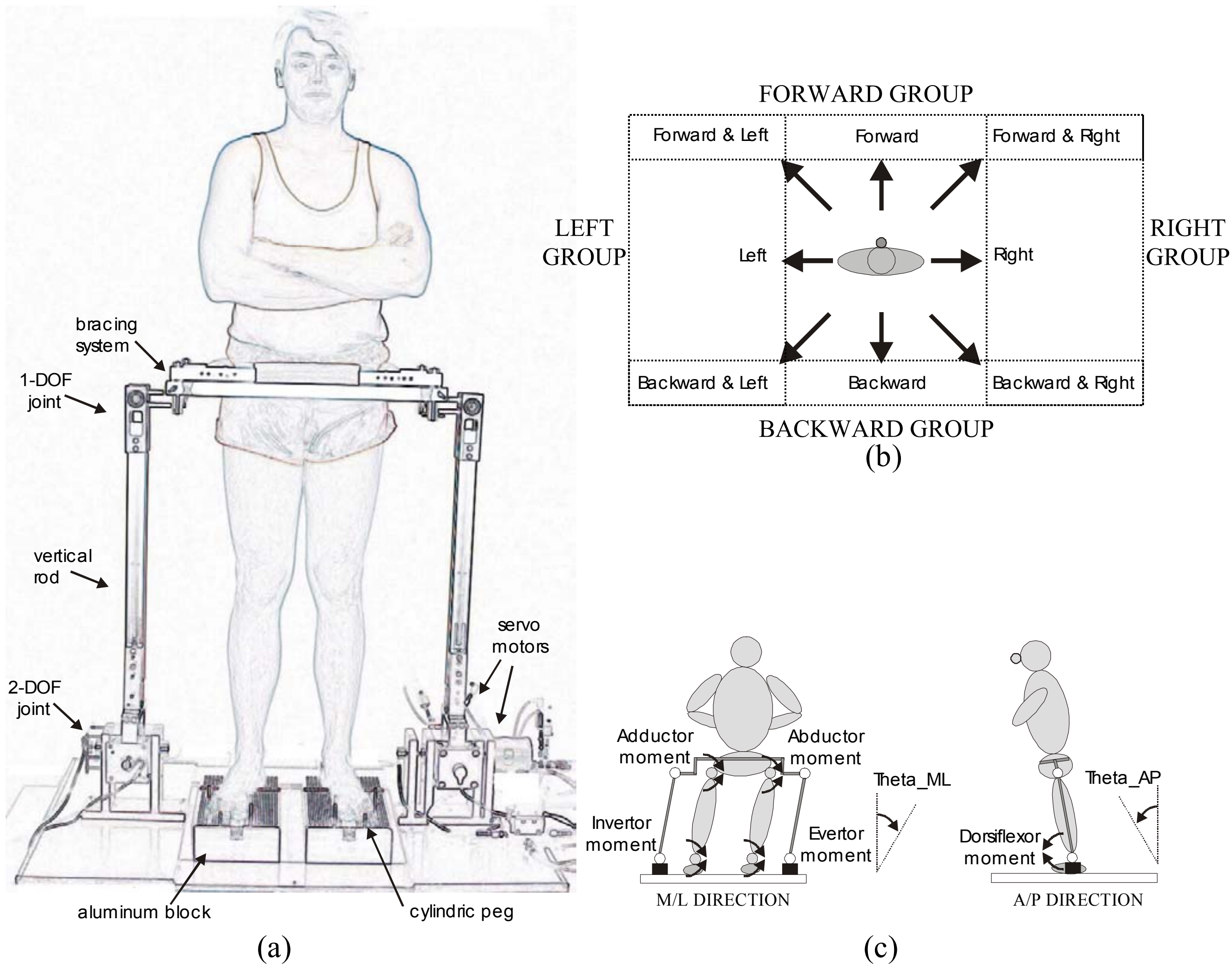

bracing system
1-DOF joint
vertical rod
servo motors
2-DOF joint
aluminum block
cylindric peg
(a)
FORWARD GROUP
Forward & Left
Forward
Forward & Right
LEFT GROUP
Left
Right
RIGHT GROUP
Backward & Left
Backward
Backward & Right
BACKWARD GROUP
(b)
Adductor moment
Abductor moment
Theta_ML
Invertor moment
Evertor moment
M/L DIRECTION
Theta_AP
Dorsiflexor moment
A/P DIRECTION
(c)

Fig 1

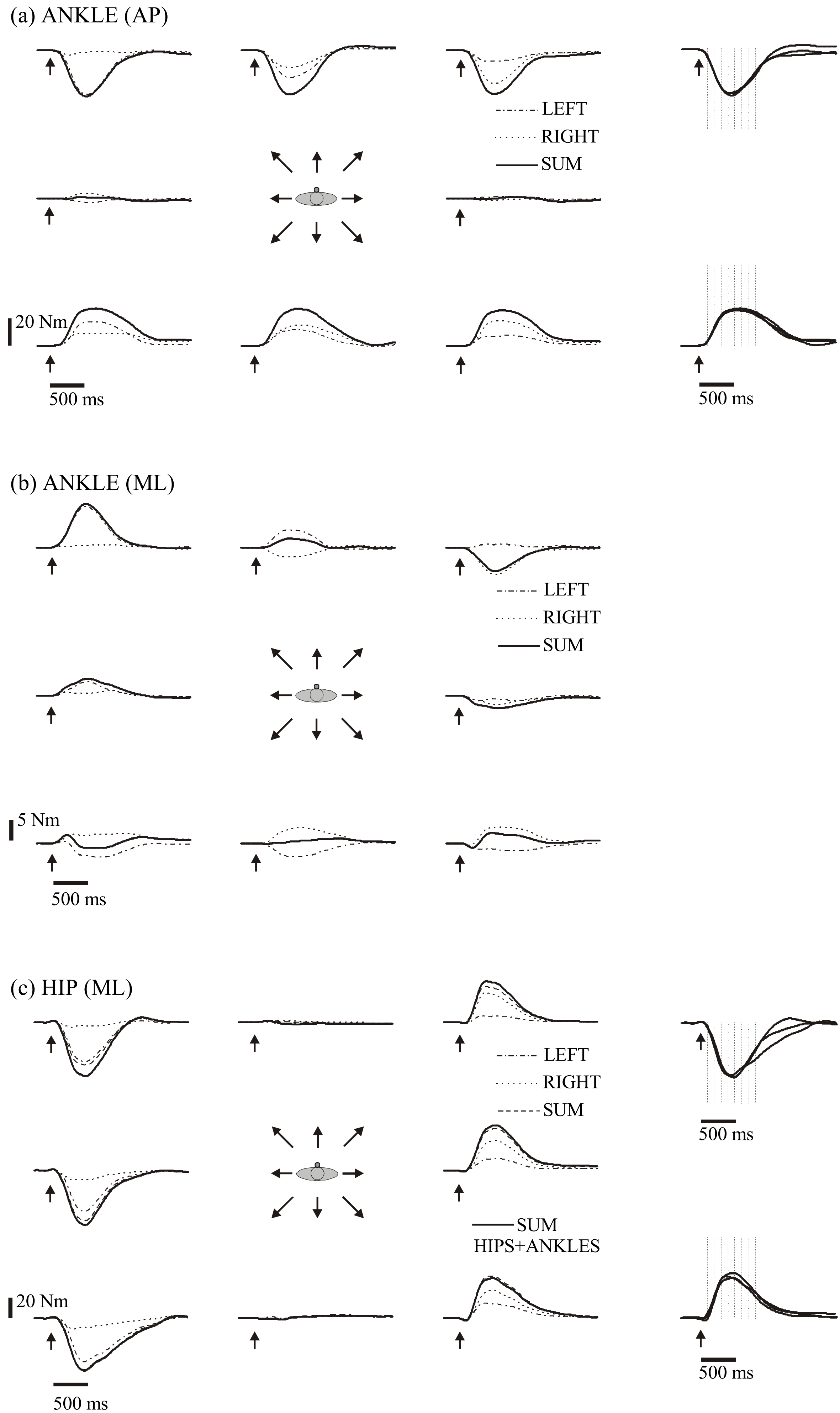

(a) ANKLE (AP)
LEFT
RIGHT
SUM
20 Nm
500 ms
500 ms
(b) ANKLE (ML)
LEFT
RIGHT
SUM
5 Nm
500 ms
(c) HIP (ML)
LEFT
RIGHT
SUM
500 ms
SUM
HIPS+ANKLES
20 Nm
500 ms
500 ms

Fig 2

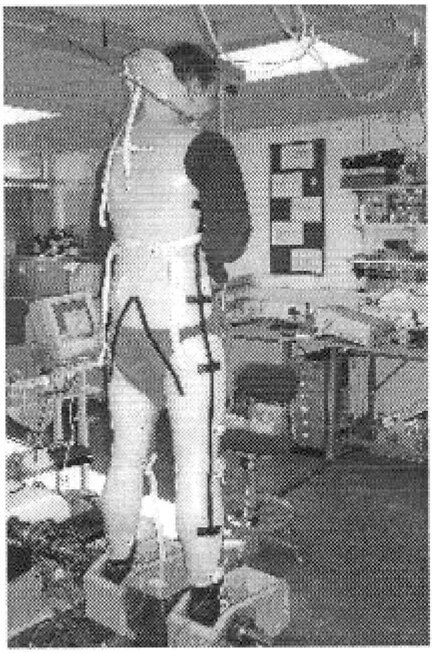


Fig 3

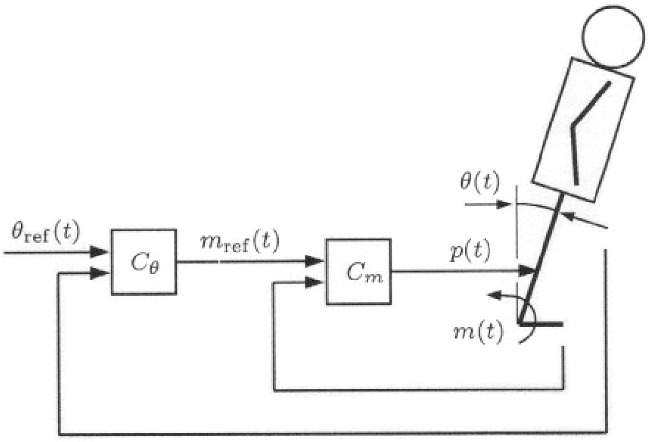
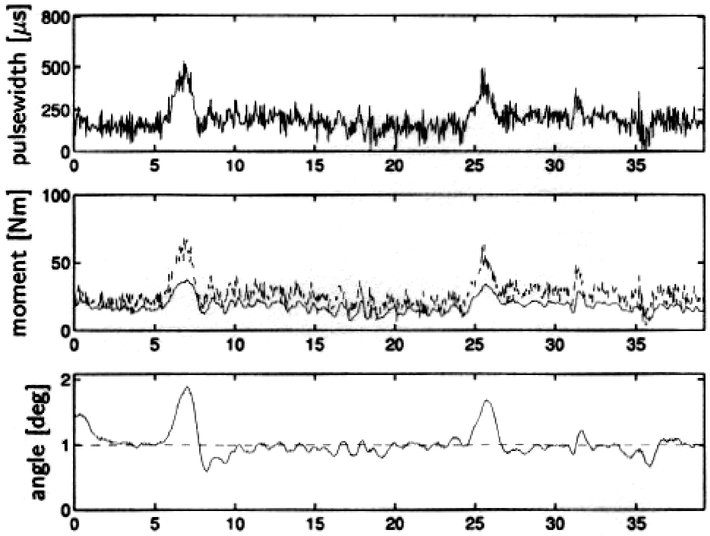


Fig 4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500
250
0
0
5
10
15
20
25
30
35
moment [Nm]
100
50
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angle [deg]
2
1
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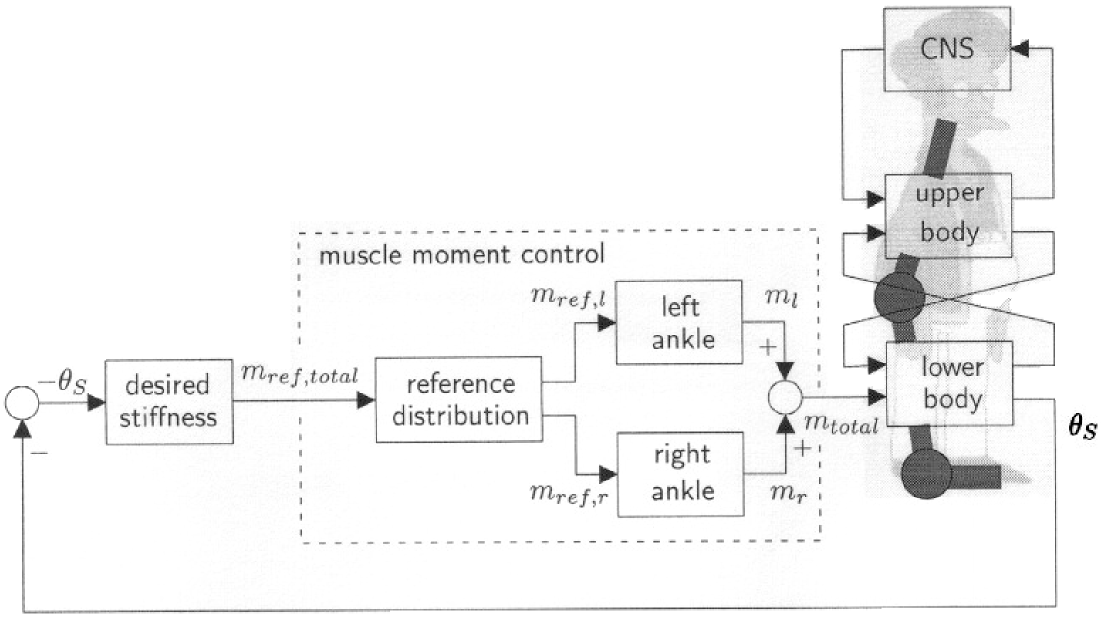
Fig 5


CNS
upper body
lower body
muscle moment control
desired stiffness
reference distribution
left ankle
right ankle
$-\theta_S$
$m_{ref,total}$
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+
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Fig 6

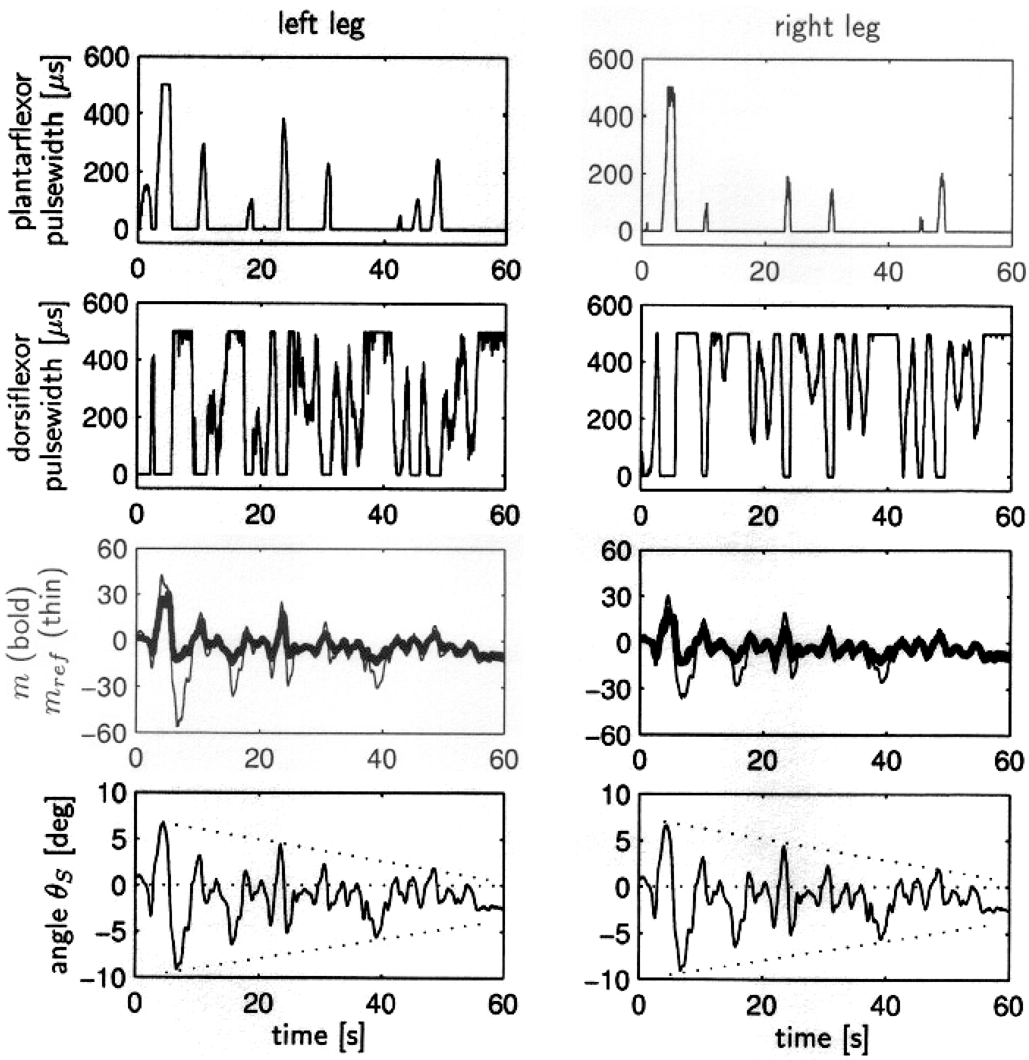
left leg
right leg
plantarflexor pulsewidth [μs]
dorsiflexor pulsewidth [μs]
m (bold) m_{ref} (thin)
angle θ_S [deg]
time [s]
time [s]

Fig 7