# A New Controller for FES-Assisted Sitting Down in Paraplegia

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***Abstract*—A new control strategy that supports the sitting down phase in FES-assisted standing in paraplegia is proposed. It is an adaptation of a well established closed-loop On/Off controller that uses trajectories within the state space of knee angle and knee angular velocity, but defines a zone between the On and Off sub-spaces for gradual change in the stimulation levels. This new controller has been experimentally tested on two paraplegic patients and it is concluded that compared to the conventional On/Off controller it better controls the lowering of the person toward the seat, by reducing knee-end velocity and handle reaction forces. It has a certain degree of interaction with the user's voluntary effort. Keeping the system simple in respect to the required number of sensors, only the knee angle measurements are required as feedback signal making the strategy simple and, thus, particularly useful for daily standing exercises in home environments.**

***Index Terms*—biomedical engineering, control strategies, neuromuscular stimulation, paraplegia, sitting down**

## I. INTRODUCTION

It is estimated that the annual incidence of spinal cord injury (SCI) in the USA, not including those who die at the scene[1] of the accident, is approximately 40 cases per million of population. The number of people living with a SCI in the USA in 2008 was estimated to be approximately 259,000 [1]. For the nations of the EU (803,850,858 population estimate in 2009 [2]) there is a lack of data, but some sources estimate that the annual incidence of SCI is approximately 14 cases per million of population [3]. Spinal cord injury results in damage to the upper motor neurons. If there is no damage to the lower motor neurons, the muscles themselves retain their ability to contract and to produce forces and movements. Functional electrical stimulation (FES) is a technology that uses small electrical pulses to artificially activate peripheral nerves causing muscles to contract, and this is done so as to restore body functions. The ability to stand SCI patients by means of FES has been reported since the early 1970s [16, 17]. Stimulation of the quadriceps muscles is regarded as the minimum to achieve standing, though other muscle groups have been used in the different phases of the sit-stand-sit manoeuvre [23], [25] and in controlling posture while standing [11,12,13]. Regular standing in spinal cord injured subjects is thought to help in preventing osteoporosis, preventing contracture by preserving the range of movement at lower limb joints, improving digestion, respiration and urinary drainage, reducing the chance of decubitus ulcers by relieving pressure and contributing to the psychological benefit by enhancing personal esteem [4], [14].

The devices that deliver electrical stimulation and aim to substitute for the control of body functions that have been impaired by neurological damage are termed 'neuroprostheses'. Several portable microprocessor or microcontroller-based neuroprostheses (for example *Compex Motion* [15], *ExoStim* [24], *Stanmore Stimulator* [19]) have been developed using a set of sensors combined with a control algorithm to trigger the pre-programmed stimulation sequences. These devices can provide the clinician with the flexibility to program subject specific applications and to apply advanced control strategies. However, the acceptance of such devices into the daily lives of SCI persons depends on the measured and perceived benefits from using the device and the ease in donning and doffing of the worn parts. With the benefits is associated their performance to control the standing phase, as well as the transition phases of standing up and sitting down. The success in the design of a control strategy for FES-assisted standing therefore requires attention to details such as robustness, ease in tuning and understanding the purpose of the control parameters, and the possibility to choose between different control strategies appropriate to that subject while performing a motion task. Controllers to assist in the different phases of standing (i.e. standing up, standing and sitting down) in paraplegia have been judged according to the following criteria that:

- They minimise the hand forces exerted by the patient and enable supported standing,
- They assure a comfortable speed of movement,
- They avoid excessive impact with the seat during sit-down,

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- They minimise metabolic energy consumption, and
- Patients and physiotherapists can use them.

The performance of neuroprostheses for standing strongly depends on the control strategy and the reliability of the feedback signals. As said, standing comprises three main tasks: standing-up, standing and sitting-down, each of them requiring adequate control. Basically, there are two approaches: *patient-centred* and *controller-centred* strategies. Patient-centred strategies [6], [7], [22] try to incorporate the voluntary contribution of the patient in the control decision; whereas controller-centred strategies [5], [18], [20], [21], [25] dictate that the patient has to comply with the controller.

In one patient-centred approach, CHRELMS (Control by Handle Reactions of Leg Muscle Stimulation) [6], [7], the arm forces applied by the patient are measured and the subsequent stimulation pattern is calculated to substitute the load by the lower limb while assuming a quasi-static movement. The required leg joint moments (called moment deficits in [6]) are calculated on a one-legged humanoid model basis. However, in a series of experiments [20, 21] differences between the left and right legs were observed in paraplegic persons and therefore the CHRELMS approach requires further attention if implemented into rehabilitation practice. Additionally, CHRELMS requires a substantial set of sensors measuring handle forces and joint positions, which limit practical application.

An alternative is the PDMR (Patient Driven Motion Reinforcement) strategy [22], where the joint positions and velocities are used as feedback signals to an inverse dynamic model that predicts the stimulation pulsewidth needed for the movement. The strategy does not require estimation of handle reaction and the computed stimulation pattern aims to maintain the movement initiated by upper body effort. Compared to CHRELMS, this strategy requires fewer sensors since only the position and velocity of the body segments is needed, but it has only been experimentally validated for standing up and sitting down movements in a simplified situation with the patient sitting on a see-saw construction. In addition, a major disadvantage of this controller is that awkward parameter adjustment is required prior to the experiments.

In conclusion, patient-centred strategies developed to date have limited practical application because of the amount of time required in clinical assessment to measure or identify the high number of parameters and the high number of sensors required; limitations that are avoided in the controller-centred strategies.

The On/Off controller described in [18] provides maximum (On) or minimum (Off) quadriceps muscle activation according to a switching line in the state-space of knee angular velocity against knee angle. However, this controller was tested in experiments only with the patient in the supine position and, therefore, perturbations and hand force reactions of the patient were avoided. It is our opinion that the controller parameter settings are difficult to visualise by a non-expert, because it is not clear how the slope of the switching line, which can be set for each patient, is related to the observable motion.

Dolan *et al.* [5] proposed a switching curve controller to support FES-assisted standing up and sitting down. The switching curve is again plotted in the knee angular velocity against knee angle state-space and derived from able-bodied subjects. The main objective of this technique was to control the terminations of the manoeuvres, i.e. the locking of the knee joints on standing up to avoid high knee angular velocities that may cause damage to the knee joint over time and the contact with the seat on sitting. Its advantage is that it does not require any parameters of the plant. Only one knee angle was used as a control input to the controller by assuming a symmetrical response to stimulation, but this has been shown to be inadequate in practice [20, 21]. We found that in sitting down, the termination angular velocities were 1.7 times greater than those for able-bodied individuals, while in standing up the angular velocities at full flexion were 3 times higher than those for able-bodied individuals, almost the same as open-loop stimulation of the quadriceps alone.

Sitting down is far more difficult to control: the patient uses their hands to support an unpredictable fraction of their body weight and the final knee angular velocities should be limited in order to protect the insensate tissues of the buttocks from injury. This paper proposes a new controller (ONZOFF or ON-ZONE-OFF) to support FES-assisted sitting down in paraplegia. As with the On/Off controllers described in [5] and [18], it works according to a switching curve in knee angle against knee angular velocity state-space, but with a gradual increase or decrease in stimulation pulsewidth between the On and Off sub-spaces, the so-called ‘Zone’. In order to perform standing exercises, a paraplegic person has to accomplish the three phases of standing up, standing and sitting down. Our intention has been to build a neuroprosthesis that integrates the control strategies for all three phases. Standing up can be easily performed by simply ramping up the stimulation intensity applied to quadriceps and the effectiveness of control strategies for the maintenance of standing has been shown [8], [10], [25]. In this paper we compare this new controller ONZOFF to the On/Off controller by measurement of kinematic data, stimulation levels and support handle reaction forces.

## II. METHODS

### A. Selection of candidates

The selected candidates were at least one-year post injury, had a complete mid-to-low thoracic lesion (T6-T12) and had completed a hospital rehabilitation programme. The strategy selected for supporting standing up, standing and sitting down determines the muscle groups that require retraining. In the controller described in this paper the quadriceps (knee extension), gluteal (hip extension) and hamstring (knee flexion) muscles are stimulated. Electrical stimulation is delivered by means of surface electrodes placed over the bulk of each of these muscle groups, but in the case of the quadriceps the electrodes were positioned to avoid stimulating rectus femoris muscle. Before being assessed for whether their muscles are strong enough to support standing, each candidate performed at least three months of FES training of the quadriceps muscles [8]. When the candidate is considered strong enough to stand, palpation tests were performed to find the respective pulsewidth for both threshold and maximal contractions; these were loaded as parameters into the control program. Standing was attempted, after the clinical team had instructed the patient about the postural movements required to complete the manoeuvre with minimal effort. Each candidate was required to be able to use their upper limbs to balance in the transverse plane while the controllers mainly provided the movement in the sagittal plane.

### *B. Clinical tests*

The clinical studies to test the proposed control strategy have been conducted in two different locations: Grosshadern Hospital of Munich, Germany, and Salisbury District Hospital, UK. Ethical approval at both centres and informed consents from participants were obtained prior to performing these experiments. Two candidates were selected for clinical study: (1) female from Germany, aged 33 years, 55 Kg, T9 motor and sensory complete, 3 years post-injury and (2) male from the UK, aged 36 years, 69 Kg, T7 motor complete and sensory incomplete, 4 years post-injury. Each candidate participated in three experimental sessions, standing up to ten times in each session. In each stand, knee angle and knee angular velocity data for left and right legs were recorded as well as stimulation pulsewidth for the quadriceps, gluteal and hamstring muscles. For the control strategies discussed in this paper the knee angle is measured as shown in figure 1. The knee angular velocity was calculated from the knee angle inter-sample difference (16-bit quantization, sampled at 100 Hz) and smoothed by a digital low-pass finite impulse response filter (cut-off frequency between 25 and 30 Hz). Handle reaction forces were also measured in the German trials.

### *C. Hardware*

The FES system used in the UK tests consists of an eight channel stimulator (Stanmore Stimulator) and two servo-potentiometers housed in neoprene knee cuffs to measure bilateral knee angles. The stimulator can deliver current-regulated stimulation pulses of maximum 150 mA and has four analog and four digital input channels.

The WALK! neuroprosthetic system used in the German tests comprises an eight channel neurostimulator (ProStim8, Neuromedics, Montpellier, France), a multi-channel sensor system, a sensory substitution system, all controlled by a process control software [10]. Ground reaction forces from insole pressure sensors (Zebris GmbH, Germany), joint angles from electrogoniometers at ankle, knee and hip joints, and rotational velocities from gyroscopes at pelvis, thighs, shank and feet are recorded at 200 Hz by a custom-built sensor system [9].

Stimulation is applied via self-adhesive surface electrodes. On both systems the stimulation current is kept constant and set for each channel individually. Pulsewidth is used to modulate muscle forces. Stimulation is applied to the quadriceps muscles to extend the knee, and may be applied to the gluteal muscles to extend the hip and to the hamstrings to both realize knee flexion and support hip extension. Both stimulators communicated with a computer via RS232 interfacing.

### *D. Control strategies*

Control strategies supporting standing up, standing and sitting down in paraplegia are implemented within an application program embedded in the stimulator, figure 1. Standing up is accomplished by stimulation of the quadriceps muscles in open-loop control with a ramped up stimulus until standing upright is achieved, as described in [17], [20], [25]. Once the subject reaches the standing position an automatic calibration is made and then, until the subject initiates sitting down, the stand is controlled according to one of two strategies; the Knee Extension Controller (KEC, used in the German tests, see [10]) or the Proportional-Integral-Derivative controller (PID, used in the UK tests) as described in [25]. During standing, the gluteal muscles are stimulated for hip extension and improved support at the hip and pelvis; ramped from zero to maximum towards the end of the standing up phase, kept constant during standing and ramped down to zero during the sitting down phase, all in open-loop. The standing up and standing phases are not the interest of this paper, but rather the sitting down phase from this upright posture. In the application program three sitting down control strategies have been implemented: /1/ open-loop control by simple ramping down of the pulsewidth of the stimulated muscle groups, /2/ the On/Off controller and /3/ the new strategy, proposed here, called ONZOFF.

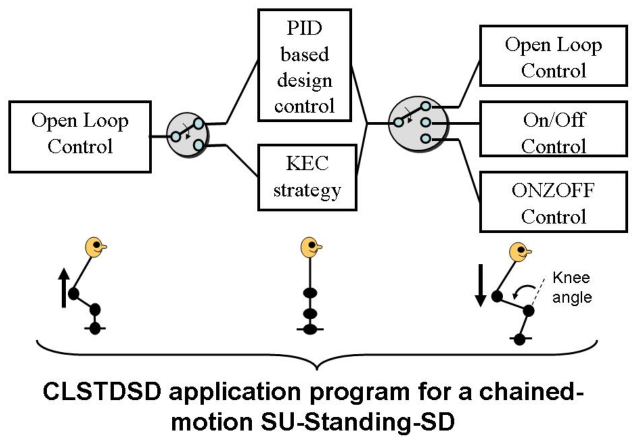


Fig. 1. Application program diagram for FES-assisted standing up (SU), standing and sitting down (SD).

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down phase from this upright posture. In the application program three sitting down control strategies have been implemented: /1/ open-loop control by simple ramping down of the pulsewidth of the stimulated muscle groups, /2/ the On/Off controller and /3/ the new strategy, proposed here, called ONZOFF.

*1) On/Off Controller*

The On/Off controller provides maximum (On) or minimum (Off) quadriceps muscle activation according to a linear switching line in state-space of knee angle against knee angular velocity, figure 2. The switching line is determined from its Ox intersection (point 'A', figure 2; 90 to- 100 °) and gradient -1.8 to -2.2 $s^{-1}$). These suggested values (in brackets) are from [18] and have been verified during our experiments. The controller enters state-space when the knees are unlocked (point 'B', figure 2; 0 to 10 °) and exits when the knee angles indicate sitting (point 'C', figure 3; 70 –to 90 °). Sometimes there is a need to unlock the knees at the beginning of sitting down by briefly stimulating the hamstrings.

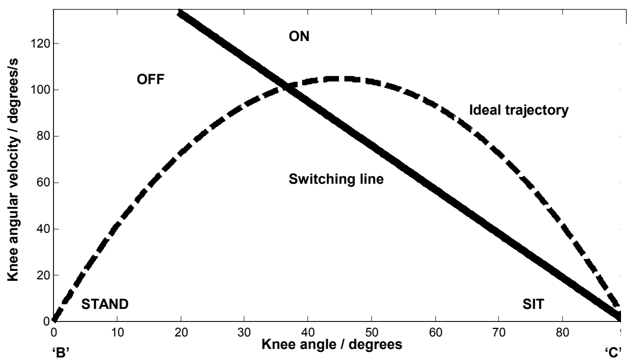


Fig. 2. State-space diagram for knee angle against knee angular velocity as for the On/Off controller (adapted from [18]). The switching line defines the On and Off regions, whereas the ideal trajectory would represent a smooth transition from standing to sitting. Points 'A', 'B' and 'C' are described in the text.

*2) ONZOFF controller*

The ONZOFF controller uses two polynomial second-order switching curves in the state-space of knee angle against knee angular velocity, figure 3. These two curves define a zone between the ON and OFF subspaces, where slightly decreasing or increasing pulsewidth for the quadriceps and hamstrings muscles can be taken into account during the sitting down phase. Therefore, sitting down is divided into three motion sub-phases: Buckle1 (knees are unlocked), Buckle2 (controlled lowering) and SitDown (ramp down stimulation to zero).

When sitting down has been initiated (i.e. by pressing a button), i.e. the start of *Buckle1*, stimulation levels to the quadriceps and gluteal muscles ramp down to a minimum and, simultaneously, the hamstring muscles ramp up to maximum in a set time; these levels are then maintained. This sequence continues until the knees are deemed to have unlocked by reaching a pre-determined angle (point 'A', figure 3) at which time the state-space working point enters within the ONZOFF controlled region, indicated as *Buckle2* in figure 3. In that region, the

1. Glueal muscles are stimulated with constant pulsewidth,
2. Quadriceps muscles are activated at maximum pulsewidth into the ON subspace, at smoothly decreasing pulsewidth (depending of actual position into the zone in the velocity direction) in ZONE and at minimum pulsewidth into the OFF subspace, and
3. The hamstrings are activated at minimum pulsewidth in the ON subspace, at smoothly increasing pulsewidth in ZONE and at maximum pulsewidth in the OFF subspace.

The ONZOFF strategy could be applied only to the quadriceps stimulation within the *Buckle2* region, but we argue that the hamstrings co-activation within the ZONE region provides a smoother motion. Co-activation of the quadriceps and hamstring muscles is expected to increase the knee joint stiffness and, therefore, control flexing of the knees as the body is lowering towards the seat. By tending to maintain the knee angular velocity within the ZONE region, a smoother lowering is expected. When the knees again reach a pre-determined angle (point 'B', figure 3), the final sub-phase of the program is entered, where all stimulation levels ramp down to zero within a predetermined time.

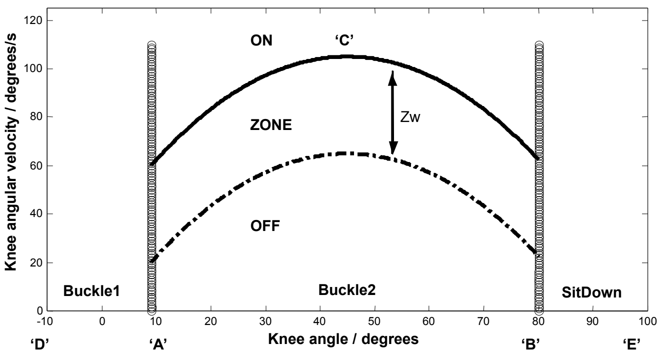


Fig. 3. State-space diagram for knee angle against knee angular velocity as for the ONZOFF controller. Points 'A' through to 'E' are described in the text.

The controller parameters are tuned from observation of each individual sitting down and the suggested values below are from our results. First, the knee angles at points A and B are determined (0 to 10 ° and 80 to -90 ° respectively). The form of the polynomial switching curve in the second sub-phase (*Buckle2*) is defined by the maximum knee angular velocity (point 'C', figure 3) and the Ox intersections (points 'D' and 'E'). Suggested values for these are 110 to 130 °/s, -10 to 0 ° and 90 to -100 ° respectively. The zone is defined by the zone width (Zw, figure 3) and a suggested value is between 30 and 70 °/s. Patients showing increased spasticity require a zone parameter value closer to the upper bound of the proposed interval. The parameters defining ZONE allow us to choose the subspace in which we would like to keep the state-space trajectory, making a smooth motion, avoiding jerky movements, and, therefore, assuring comfort for the patient. The maximum knee angular velocity parameter of the switching curve can be reduced if the descent speed is considered too high and similarly the zone width parameter can be increased (ZONE made wider) if switching between the On (maximum pulsewidth) and Off (no stimulation) states results in jerky movements.

## III. RESULTS

The sitting-down controllers and the required software and hardware have been intensively tested (113 trials on healthy subjects and 77 trials on SCI patients). The WALK! neuroprosthetic system [10] that has been used in Germany provides a great amount of information concerning ground reaction forces, hand forces, joint angles at ankle, knee and hip joints, and rotational velocities. It helped us to earn a deeper understanding of the control process. Moving towards clinical application requires ease in donning and doffing, reducing the number of sensors and ease in tuning the controller parameters. We translated the overall control strategy on a Stanmore Stimulator and the required number of sensors has been kept as low as possible (two servo-potentiometers housed in bilateral neoprene knee cuffs). A laptop has been used to collect data during each trial and present it graphically after each trial.

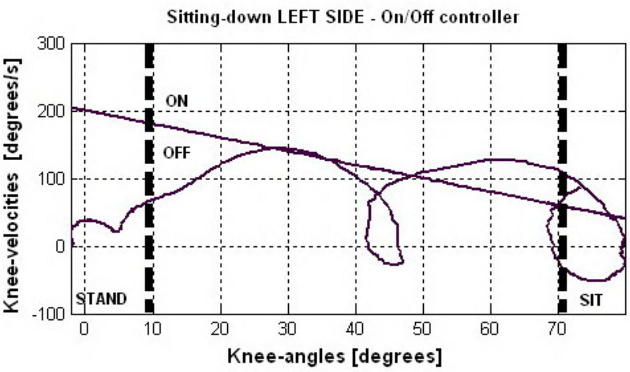


Fig. 4. Knee angle against knee angular velocity state-space trajectory using an On/Off controller to support sitting down (recorded data for the left side). The controller is defined by Ox = 100 °, gradient = -2 s-1 and operating between 10 ° and 70 ° (c.f. solid straight line).

### A. *On/Off Controller*

Fourteen trials (two experimental sessions) were performed with subject 1 to test this sitting down control strategy.

It was found that the parameters for left and right controllers had to be tuned separately; data defining the controller, along with resulting state-space trajectory and recorded data with an On/Off controller for sitting down, is indicated in figure 4.

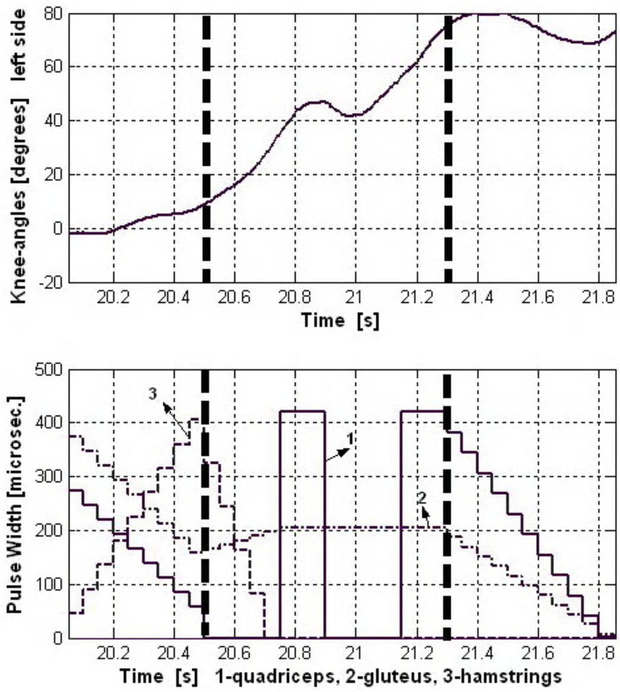


Fig. 5. Recorded data for the left side during the On/Off controlled sitting down motion task: knee angle (top plot) and stimulation pulsewidth (bottom plot) against time. The controller is operating between 20.5 and 21.3 seconds.

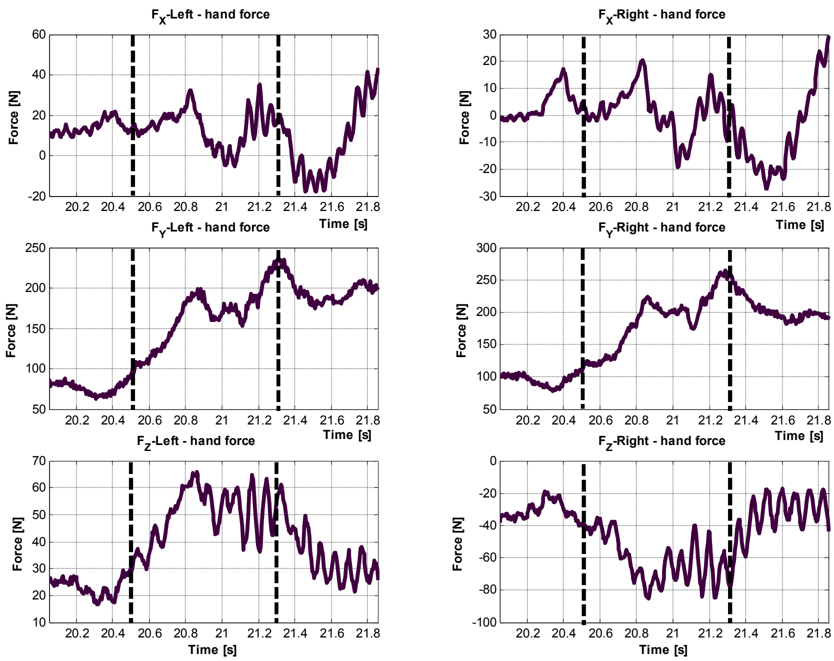


Fig. 6. Recorded data for handle reaction forces (positive values for FX [forwards], FY [vertically up], and FZ [lateral]) during On/Off controlled sitting down; the controller is operating between 20.5 and 21.3 seconds.

Recorded data for knee angle and stimulation pulsewidth and the handle reaction forces during sitting down are illustrated in figures 5 and 6 respectively. The current intensity has been kept constant at 110 mA.

It is shown that as the knees buckle, then the controller modulates sitting down. However, a high knee angular velocity occurred, when entering the SIT subspace. Mulder et al. [18] focused more on standing up but only in the supine position, and even in such a context they remarked a high non-reproducibility during experiments. During our experiments the same behaviour was observed. We conclude that this is due to spasticity of the knee flexors and differences in the voluntary handle reaction forces during different trials. The On/Off controller does not show any robustness in the presence of these perturbations. Furthermore, this determines a great variability in knee end velocities at the sitting moment, mostly between 90 °/s and 115 °/s.

Figure 4 shows the knee angle against knee angular velocity state-space trajectory while using an On/Off controller to support sitting down. The patient is practically falling down (OFF subspace) until the trajectory intersects the switching line (knee angle at about 27°). The patient reacts by exerting vertical forces $F_z$ with his/her hands (see figure 6, at about 20.7 s).

Once the trajectory reaches the switching line (ON subspace) the quadriceps are stimulated at maximum (figure 5, at about 20.75 s), hence reducing the falling speed. The lower limb support enables the patient to reduce the upper limb involvement (see figure 6, at about 20.9 s to 21.2 s). The trajectory re-enters the OFF space, thus, quadriceps stimulation is switched off, and the patient descends further (knee buckling, see figure 4, knee angle at about 42° to 48°), so they increase upper limb involvement (increased hand reaction forces) and again the quadriceps are stimulated at maximum pulsewidth. As sitting is approached at a knee end velocity of about 110 °/s, the stimulation is switched off and the patient finally sits. There is no thin modulation in quadriceps stimulation level around the switching line that in most experiments determine knee buckling and high knee end velocities.

*B. ONZOFF Controller*

Figures 7, 8 and 9 show the results that have been obtained while sitting down has been performed by means of an ONZOFF control strategy with the same subject and on the same day. Other parameters that have been chosen at the very beginning of the trials and have been kept constant during the experiments are: pulsewidth stimulus at 320 μs for the quadriceps muscles within the ON region, pulsewidth stimulus at 250 μs for the hamstrings muscles within the OFF region and pulsewidth stimulus at 250 μs for the gluteal muscles. The current intensity has been kept constant at 110 mA. Within the Buckle1 phase the hamstrings are stimulated at the maximum pulsewidth in order to unlock the knee.

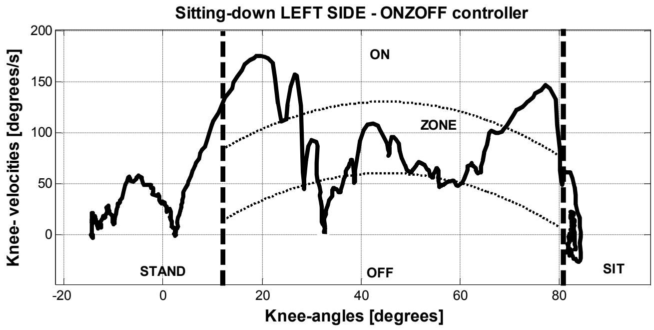


Fig. 7. Knee angle against knee angular velocity (data for the left side) in a state-space trajectory using an ONZOFF controller to support sitting down. The controller is defined by Ox = -10 and 100 °, maximum knee angular velocity = 120 ° s-1, a zone width of 70 ° s-1 and operating between 10 and 80 °.

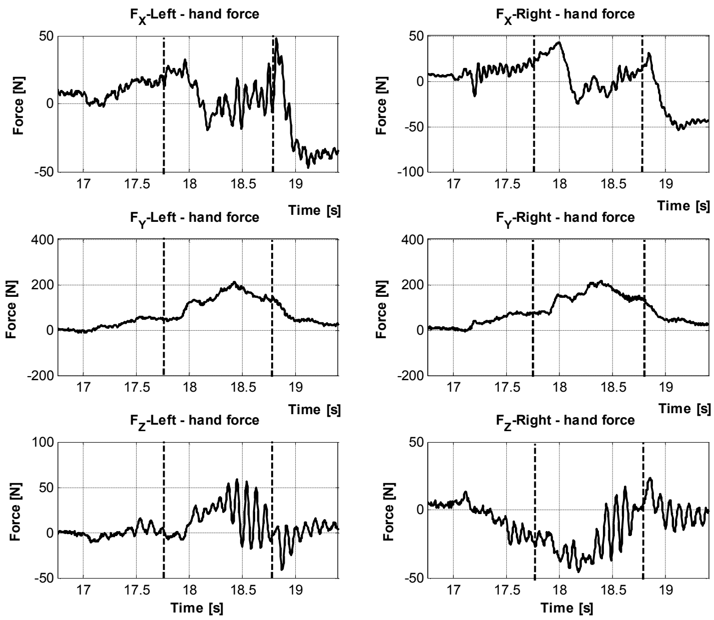


Fig. 8. Recorded data for handle reaction forces (positive values for $F_X$ [forwards], $F_Y$ [vertically up], and $F_Z$ [lateral]) during ONZOFF controlled sitting down; the controller is operating between 17.8 and 18.75 seconds.

Figure 7 shows the knee angle against knee angular velocity state-space trajectory while using an ONZOFF controller to support sitting down. While entering the controlled area (*Buckle2*) there is usually an increase in knee angular velocity. The controller reacts by bringing the trajectory within the ZONE subspace where the quadriceps stimulation varies more smoothly. The patient feels confident and hence decreases their exerted vertical hand forces (see figure 8, at about 18.1s); it is near to knee buckling (figure 7, knee angle at about 32°). The ONZOFF controller reacts and smoothly brings the trajectory within the ZONE subspace (figure 9, time: 18.3s to 18.7s). Being confident, the patient reduces again the exerted vertical hand forces, and the controller switches most of the support to the lower limb by stimulating the quadriceps at the stimulation level settled for the ON subspace, causing a decreased the knee end velocity of about 58 °/s as the *SitDown* phase is entered.

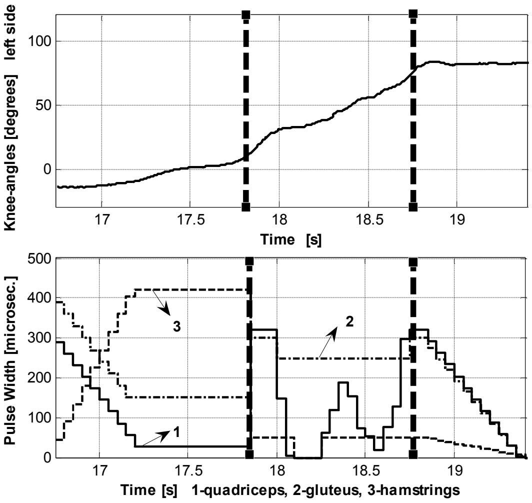


Fig. 9. Recorded data for the left side during the ONZOFF controlled sitting down motion task: knee angle (top plot) and stimulation pulsewidth (bottom plot) against time. The controller is operating between 17.8 and 18.75 seconds.

### C. Data Analysis

There have been observed differences between left and right legs, so the parameters for left and right controllers have been tuned separately. Among the total number of 77 trials on these two SCI patients, some trials have been performed to test the equipment response, to calibrate, to find the stimulation parameters intervals for a proper contraction and to adjust the graphic user interface characteristics. The parameters have been tuned in a number of twenty ONZOFF controllers and fourteen On/Off controllers, the overall trials being accomplished by the SCI patient. Among the huge amount of the collected data (numerical data, video data) we analysed the knee end velocity for both controllers. It is an important parameter to be analysed while the lack of control on the end velocity may lead to medical complications by repeatedly striking the seat. The mean of the knee end velocity was 67.6 °/s (standard deviation of 33.1°/s) with the ONZOFF controller and 106.9 °/s (standard deviation of 16.7 °/s) with the On/Off controller, and the difference was statistically significant. A hypothesis test has been used to quantify the test of normality. Since each sample is relatively small, a Lilliefors test is recommended. The significance level of the Lilliefors test is 5%. The logical 1 returned by each test didn't indicate a failure to reject the null hypothesis that the samples are normally distributed. The non-parametric Wilcoxon rank sum test has been used to quantify the test of equal medians. It tests if two independent samples come from identical continuous (not necessarily normal) distributions with equal medians, against the alternative that they do not have equal medians. The test rejects the null hypothesis of equal medians at the default 5% significance level. The probability of observing the given result, or one more extreme, by chance if the null hypothesis ("medians are equal") is true, is $p = 0.0002$. Finally, a t-test has been performed. It tests if two independent samples come from normal distributions with equal but unknown standard deviations and the same mean, against the alternative that the means are unequal. The null hypothesis has been rejected at the default 5% significance level. The significance is 0.00018. The computation is based on an assumption of normality in the data, but the comparison is reasonably robust for other distributions.

## IV. DISCUSSION AND CONCLUSIONS

A FES-based system able to support transfers such as wheelchair-to-toilet and wheelchair-to-bed has the potential to improve paraplegics' quality of life. Making it as simple as possible and minimising how much equipment must be worn by the user increases compliance. The new ONZOFF controller proposed in this study requires only knee angle sensors to meet this ideal. It allows the user to control sitting down. If the patient wants to support a higher percentage of his/her body weight with their upper body then the controller reacts by decreasing the quadriceps stimulation, as it is necessary to maintain the movement as initiated.

In our experiments we tuned the ONZOFF controller only for sitting down. Even though it would be desirable to control standing up to avoid high knee angular velocities near the end of that phase that may harm the knee joint over time if used in daily standing exercises, our patients that tested the neuroprosthesis felt comfortable with the open-loop control of standing up that simply ramps up the stimulus intensity applied to the quadriceps. Perhaps this is fortunate because closed-loop standing up may be quite difficult because of the quick transition (~0.8 seconds in able bodied persons). Recently proposed closed-loop control schemes included the voluntary arm contribution within the control strategy [6], [22]. Due to their requirements for sensors, and the need for system identification to furnish parameters for the controllers, these control schemes are far from being ready for widespread use. In contrast, the controller-centred strategies described here are easy to set up and we found that after a few trials the subjects were supporting less than 50% of their body weight through their arms and once their confidence increased they interact easily with the whole system.

The open-loop systems are based on preset down ramps. A major disadvantage is the lack of control on the end velocity, which may lead to medical complications by repeatedly striking the seat. Nowadays there still are approaches which propose the use of preprogrammed stimulation pattern to accomplish a sitting-down task while an exoskeletal bracing system ensures task safety [26]. In our experiments we found that it is quite difficult to choose the right switching line for the On/Off controller so that it will cope well with the perturbation caused by the hand forces that are providing support for some of the body weight and with differences in posture when again sitting. The ONZOFF controller has certain robustness and copes better with that kind of perturbation, trying to keep the trajectory within a ZONE where the stimulation pulsewidth is slowly increased or decreased. With the ONZOFF controller, at the beginning of the sitting down motion, due to a rapid increase of knee angular velocity, the patient reacts with his hands to decrease knee angular velocity. Then as his confidence increases, the hand forces remain quite constant (around 50% of body weight – right side of figure 8) and the controller working properly ensures a slow descent. Knee end velocity for the ONZOFF controller has been found to be significantly lower than that measured with the On/Off controller; the mean was 67.6°/s with the ONZOFF controller and 106.9°/s with the On/Off controller, and the difference was statistically significant.

The ONZOFF control strategy has been designed to allow implementation in different neuroprostheses that contain an embedded controller. The advantage of this controller is the more intuitive means to tune the controller parameters, whilst not compromising the clinical applicability. By minimising the number of controller parameters and relating each one to an observed characteristic of the sit-stand-sit motion, our experience indicates that disseminating the information to clinicians, who do not necessarily have a technical background, in order for them to be able to set up a controller for each patient is relatively straightforward. The system requires only knee angle measurements as feedback and keeping it simple in respect of required number of sensors makes it particularly useful for regular use at home. We therefore believe that that this method could be widely used both in the clinical environment and at home for daily standing exercise.

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## REFERENCES

[1] "Spinal Cord Injury statistics (updated June 2009)". Available from the Foundation for Spinal Cord Injury Prevention, Care and Cure. http://www.fscip.org/facts.htm [on-line]. [Accessed 30 September 2009].

[2] Internet world stats. http://www.internetworldstats.com/europa2.htm [on-line] [Accessed 30 September 2009].

[3] Paraplegic and Quadriplegic Forum for Complete and Incomplete Quadriplegics and Paraplegics and wheelchair users Paralyzed with a Spinal Cord Injury. http://www.apparelyzed.com/forums/index.php?s=2f63834381835165c0aca8b76a0acc74&showtopic=2489 [on-line]. [Accessed 30 September 2009].

[4] P. W. Axelson, D. Gurski, and A. Lasko-Harvill, "Standing and its importance in spinal cord injury management," in Proc. RESNA 10th Ann. Conf. Rehab. Tech., San Jose, CA, pp. 477-479, 1987.

[5] M. Dolan, B. Andrews, and P. H. Veltink, "Switching curve controller for FES assisted standing up and sitting down," IEEE Trans. Rehabil. Eng., vol. 6, pp. 167-171, 1998.

[6] N. d. N. Donaldson, and C. H. Yu, "FES standing control by handle reactions of leg muscle stimulation (CHRELMS)," IEEE Trans. Rehabil. Eng., vol. 4, pp. 280-284, 1996.

[7] N. d. N. Donaldson, and C. H. Yu, "A strategy used by paraplegics to stand up using FES," IEEE Trans. Rehabil. Eng., vol. 6, pp. 162-167, 1998.

[8] D. J. Ewins, P. N. Taylor, and I. D. Swain, "The Odstock closed-loop FES standing system: experience in clinical use," in Proc. Ljubljana FES Conf., Ljubljana Slovenia, pp. 97-100, 1993.

[9] T.Fuhr and G. Schmidt, "Design of a patient mounted multi-sensor system for lower extremity neuroprostheses", in Proc. 21st IEEE EMBS Conf., Atlanta, October 16-19, pp.1-4, 1999.

[10] T. Fuhr, J. Quintern, R. Riener, and G. Schmidt, "Assisting locomotion in patients with paraplegia. Control of WALK! – a cooperative patient driven neuroprosthetic system," IEEE EMBS Magazine, vol.27, pp. 38-48, 2008.

[11] H. Gollee, K. J. Hunt, and D. E. Wood. "New results in feedback control of unsupported standing in paraplegia," IEEE Trans. Neural Syst. Rehabil. Eng., vol. 12, pp. 73-80, 2004.

[12] A. H. Vette, K. Masani, J.-Y. Kim, and M. R. Popovic, "Closed-loop control of functional electrical stimulation-assisted arm-free standing in individuals with spinal cord injury: A feasibility study," Neuromodulation, vol.12, no.1, pp.22-32, 2009. http://www.blackwell-synergy.com/loi/ner [on-line] [Accessed 10 January 2010].

[13] X. Tortolero, K. Masani, C. Maluly, and M. R. Popovic, "Body movement induced by electrical stimulation of toe muscles during standing," Artificial Organs, vol.32, no.1, pp.5-12, 2007.

[14] R. J. Jaeger, and G. M. Yarkony, "Implementing a simple FNS standing protocol in a clinical setting," in Advances in External Control of Human Extremities, Dubrovnik Yugoslavia, pp. 265-273, 1990.

[15] T. Keller, M. R. Popovic, I. P. I. Pappas, and P. Y. Muller, "Transcutaneous functional electrical stimulator "Compex Motion"," Artificial Organs, vol. 26, pp. 219-223, 2002.

[16] A. Kralj, S. Grobelnik, and L. Vodovnik, "Electrical stimulation of paraplegic patients – feasibility study," in Proc. Int. Symp. External Control of Human Extremities, Dubrovnik Yugoslavia, pp. 561-565, 1973.

[17] A. Kralj, and T. Bajd, "Functional electrical stimulation: standing and walking after spinal cord injury", CRC Press, Inc., Boca Raton, 1989.

[18] A. J. Mulder, P. H. Veltink, and H. B. K. Boom, "On/off control in FES-induced standing up: A model study and experiments," Med. Biol. Eng. Comput., vol. 30, pp. 205-212, 1992.

[19] G. F. Phillips, J. A. Adler, and S. G. J. Taylor, "A portable programmable stimulator for surface FES," in Proc. Ljubljana FES Conf., R. Jaeger and T. Bajd, Eds., pp.166-168, 1993.

[20] M. S. Poboroniuc, T. Fuhr, R. Riener, and N. d. N. Donaldson, "Closed-loop control for FES-supported standing up and sitting down," in Proc. 7th IFESS Conf., Ljubljana Slovenia, pp. 307-309, 2002.

[21] M. S. Poboroniuc, T. Fuhr, D. E. Wood, R. Riener, and N. d. N. Donaldson, "FES induced standing-up and sitting-down control strategies in paraplegia," in FESnet Conf., Glasgow UK, pp. 1-3, 2002.

[22] R. Riener, and T. Fuhr, "Patient-driven control of FES-supported standing up: a simulation study," IEEE Trans. Rehabil. Eng., vol. 6, pp. 113-124, 1998.

[23] M. E. Roebroeck, C. A. M. Doorenbosch, J. Harlaar, R. Jacobs, and G. J. Lankhorst, "Biomechanics and muscular activity during sit-to-stand transfer," Clin. Biomech., vol. 9, pp. 235-244, 1994.

[24] S. Simcox, G. Davis, A. Barriskill, and J. Middleton, "A portable, 8-channel transcutaneous stimulator for paraplegic muscle training and mobility - technical note," J. Rehabil. Res. Devel., vol. 41, pp. 41-52, 2004.

[25] D. E. Wood, V. J. Harper, F. M. D. Barr, P. N. Taylor, G. F. Phillips, and D. J. Ewins, "Experience in using knee angles as part of a closed-loop algorithm to control FES-assisted paraplegic standing," in Proc. 6th Int. Workshop on FES: Basics, Technology and Application, Vienna Austria, pp. 137-140, 1998.

[26] R. Kobetic, C.S. To, J. R. Schnellenberger, M. L. Audu, T. C. Bulea, R. Gaudio, G. Pinault, S. Tashman, R. J. Triolo, "Development of hybrid orthosis for standing, walking, and stair climbing after spinal cord injury," J. of Rehabil. Res. Devel., vol.46, no.3, pp.447-462, 2009.