APPLICATIONS OF
MODELLING AND SIMULATION
http://arqiipubl.com/ams eISSN 2600-8084 VOL 3, NO. 2, 2019, 102-110

Effect of Subject Mass with Feedback Linearisation Controller for Induced Sit-to-Stand Regulation

Mohammed Ahmed[1,2,3*], M. S. Huq[4] and B. S. K. K. Ibrahim[3,5]

[1]Department of Electrical and Electronics Engineering, Faculty of Engineering and Engineering Technology, Abubakar Tafawa Balewa University, P. M. B. 0248 Bauchi, Bauchi State, Nigeria
[2]Department of Mechatronic Engineering, Faculty of Electrical and Electronic Engineering, Universiti Tun Hussein Onn Malaysia, 86400 Parit Raja, Johor, Malaysia
[3]Advanced Mechatronic Research Group, Faculty of Electrical and Electronic Engineering, Universiti Tun Hussein Onn Malaysia, 86400 Parit Raja, Johor, Malaysia
[4]Department of Mechatronic Engineering, Faculty of Engineering Design, Institute of Technology Sligo, Ash Lane, Sligo F91 YW50, Ireland
[5]School of Mechanical, Aerospace and Automotive Engineering, Coventry University, Coventry, United Kingdom

*Corresponding author: inunugoloma@yahoo.com

Received 31 January 2019, Revised 16 April 2019, Accepted 20 May 2019.

Abstract: Presented is the application of the Feedback Linearisation Control (FLC) method for compensation of variations in human masses for the restoration of sit-to-stand movement function with the aid of Functional Electrical Stimulation. According to literature, enhancement of the control system of such arrangements, which is one of the promising techniques employed is required to attain the desired goal. Hence, improving the system by making it more accurate and robust with aim ushering such devices towards clinical acceptance. The FLC approach is employed, and effort is made to investigate the effect of global human mass distribution as obtained in the literature. The plant is modelled by using the Newton-Euler and Euler-Lagrange methods for the segment dynamics, and the muscles model is adopted from the previous works. The plant for the study is the paraplegic subject and for sit-to-stand movement revival. The control system tries to maintain the stimulation current optimum during the entire process. Results show remarkable improvements with a drastic reduction in the stimulation current, the rate of change in the stimulation current, and the tracking error. Therefore, the system would have a drastic enhancement in accuracy and delayed attainment of fatigue.

Keywords: Feedback linearisation control; Functional electrical stimulation; Induced sit-to-stand; Paraplegia; Robustness.

1. INTRODUCTION

Functional Electrical Stimulation (FES) is a method popularly employed for revival lost movement functions as results of injuries in the nervous system [1]. The fatalities of the human nervous could emanate due to occurrences of accidents, falls or diseases. Additional essential applications of the FES are therapy and rehabilitation of movements [2-4]. The technique though very useful but has not yet correctly pass clinical approval [5]. More improvement is required, and control systems have been portraying fruitful results. A significant challenge worth addressing was that of different human mass distribution across the world. The variations may require changing the controller operating variables to give appropriate levels of compensations. The control system should be robust enough to accommodate that, so that the equipment will be universal making it efficiently and effectively applicable for different subjects and hence the possibility of its users worldwide with just a little or no adjustments [6, 7]. The scope of the study was limited as follows: it was for restoration Sit-to-Stand (STS) movement in paraplegia, the significant disturbance under consideration are the changes in the subject's mass, and it is a simulation work harnessing the SIMULINK/MATLAB software.

According to the literature, the Proportional-Derivative (PD) control method was utilised by Tsukahera *et al*. [8] and Fattah *et al*. [9] for the revival of the STS movement. The Proportional Integral Derivative (PID) was employed by Yu *et al*. [10], Previdi *et al*. [11] and Pobonoruic [12] for a similar purpose. In their works, linear control techniques were utilised, and researchers are of the view that such methods cannot adequately handle such systems due to their complexities attributed to rigorous nonlinear nature [13, 14]. The Adaptive Network-based Fuzzy Inference System (ANFIS) and Genetic Algorithm (GA)-ANFIS were used by Davoodi and Andrews [15, 16] in their simulation study for the STS restoration in paraplegia. The PID-ANFIS technique was applied by Hussain *et al*. [17] and Massoud [18]. PID-Artificial Neural Networks (ANN) was used by Afzal *et al*. [19] for the scenario. These control methods are categorised as intelligent based techniques. In reality, the methods are also nonlinear, but the embedded intelligence make their representations very difficult with the aid of mathematical equations. The absence of these equations makes the methods very difficult for some essential analyses, such as evaluating the levels of stability [20]. The Model Predictive Control was harnessed by Espanjani and Towhidkhah [21] for the FES aided STS in paraplegic. Results obtained were encouraging, but the approach was quite complicated. The technique employed in this paper is designated nonlinear and has the advantage of having explicit mathematical models apart from improvement in the system performance. The Feedback Linearisation Control (FLC) scheme is worth exploring because it has been applied for other forms of FES induced movements with relative success, and it is also a nonlinear control scheme. Therefore, it is expected the dual advantages mention could be beneficial in tackling the issue of robustness while handling different masses associated with humans around the world. The intention was making the device proficient in handling such changes with minor or no need of adjusting of variables to cancel the effects.

2. METHODOLOGY

Figure 1 shows the simplified block diagram of the system under study. It has basically three components: The plant or system, the controller or control system and the disturbance (effect of human mass). The various components are the controller, stimulator, plant, desired trajectory and the output. The controller is a compensator which always tries to make adjustments to annul for the changes in error e which is the difference between the desired trajectory, θ_d and the output, θ. The controller varies the pulse width, P_w of the stimulator which adjusts the amount of the stimulation current, i which is changed based on the control signal/law, u. It is then sent to the knee muscle that produced torque used for the STS movement. The variable mass block indicates how the mass of the paraplegic patient is changed between the lowest and highest levels for the examination purpose. More descriptions about the system are discussed in the next section. The plant is the paraplegic subject whose STS movement is facilitated by the FES. The movement is initiated when the appropriate muscles in the thigh are stimulated. The mathematical model is obtained using the principles of robotics for the segment dynamics and the Ferrarin and Pedotti [22] knee joint representation for the muscle model as given in Equations (1)-(3). ω is the resting angle of the knee joint, G is the gain, δ is the time constant, β is the coefficient of dynamic friction, τ_A is the torque produced at the knee joint as a consequence of stimulating the required muscles, τ_D is the dynamic torque and τ_S the stiffness torque.

$$\tau_A = P_W G/(1 + \delta s) \tag{1}$$

$$\tau_D = \beta\dot{\theta} \tag{2}$$

$$\tau_S = (\theta + \pi/2 - \omega) \tag{3}$$

The control system regulates the amount of stimulation current to achieve the desired movement efficiently. Although the study focusses on the FLC, a gain scheduling controller based on the PID was used for comparison purpose. It is referred to as the PID in the text. The recommended settings for the movement are 20-500 µs for the pulse duration, 10-50 Hz for the frequency and 0-130 mA for the stimulation current [23, 24].

2.1 The Controller

The controller tries to make compensation using feedback via sensors. It compares the current output and the reference or desired movement position. The compensation is done by increasing or decreasing the amount of stimulation depending on the level of the system response. The equations of the control law are obtained, as [25, 26]

$$\tau = D(\theta)\ddot{\theta} + C\big(\theta,\dot{\theta}\big)\dot{\theta} + g(\theta) \tag{4}$$

The torque in the left-hand side of Equation (4) is obtained as the summation of Equations (1), (2) and (3). Equation (4) is the generalised form of the model, which is obtained for the system under study using the Euler-Lagrange and Newton-Euler methods. $D(\theta)\ddot{\theta}$ is the inertia matrix, $C\big(\theta,\dot{\theta}\big)\dot{\theta}$ is corriolis matrix and $g(\theta)$ is the gravity matrix. These parameters are functions of the various segment masses, lengths and angular orientations during the movement. The masses of the segments would change depending on the mass of the subject.

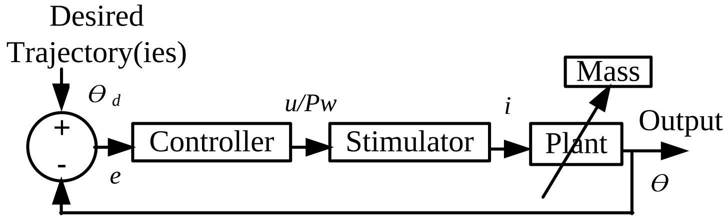

Figure 1. Block diagram of the system

Note that the human body is approximated using four segments namely the shank, thigh, trunk, and above the shoulder is categorised as the neck segment. Therefore, τ is a 4×1 matrix, D and C vectors are 4×4 matrices and g is a 4×1 matrix. The τ are torques at the ankle, knee, thigh and the neck joints. The excitation was achieved at the knee joint, hence the parameters in Equations (1), (2) and (3) are associated with it. The torque produced at the knee joint is governed by Equation (1) and it can be seen that the variable that determines the level of torque is the pulse width of the current, P_W and it becomes the control signal, u. The other parameter as given by Equations (2) and (3) are substituted into C and g matrices respectively in order to collect the like terms.

The plant model was then arranged in a canonical form, as shown in Equation (5). They are functions of the various segment masses, lengths and angular orientations as they are obtained from Equation (4). Their details are not shown due to lengthy nature.

$$\ddot{\theta} = f(\theta, \dot{\theta}) + g(\theta, \dot{\theta})u \tag{5}$$

$$u = \frac{\ddot{x} - f(x,t)}{g(x,t)} \tag{6}$$

$$u = \frac{\ddot{x}_d + ek_1 + \dot{e}k_2 - f(x,t)}{g(x,t)} \tag{7}$$

$$\ddot{e} + ek_1 + \dot{e}k_2 = 0;\ k_1 and\ k_2 \geq 0 \tag{8}$$

$f(\theta, \dot{\theta})$ and $g(\theta, \dot{\theta})$ are assumed to be determinate (non-zero) and continuous, θ is the angle, $\dot{\theta}$ is the velocity of the angle, e is the error in the angle, $\dot{e}$ is the rate of change in the error, $\ddot{e}$ is the second order rate of change in the angle, k_1 and k_2 are the controller parameters to be tuned. Equation (5) is the plant model in a canonical form and resulted to Equation (6) when the control signal u is made the subject. The control law using the FLC is given in Equation (7) by making appropriate replacement whose origin is Equation (8) and k_1 and k_2 are chosen to suit Equation (8).

2.2 Global Mass Distribution

The minimum mass considered is 45 kg, which correspond to about 53% of the mass of the paraplegic subject, which is 85 kg. The maximum mass is 107 kg, which is about 26% above the subject. Literature shows that Koreans have the least weight, which is nearly 45.7 kg. It also reveals the highest human mass in the case of the United States, which is up to an average of 89 kg [27-29].

3. RESULTS AND DISCUSSIONS

Figures 2-7 are the plots of the tracking response, the tracking errors, cumulative tracking errors, the control signals, cumulative stimulation currents and the cumulative rate of change in the stimulation currents. Even though the tracking errors cannot be seen clearly in Figure 2, but they are clearly shown in Figure 3. The root means square error (RMSE) are 269.7000 and 0.8406 with the PID and FLC, respectively, without disturbance. Hence, it indicates that an average error of around 260 times higher will result in the PID compared to the FLC. As mentioned earlier, this property is vividly portrayed in Figure 3.

Figure 3 shows the plots of the Absolute Tracking Error (ATE) and Time Absolute Tracking Error (TATE). Apart from indicating the levels of magnitudes of the tracking errors, it shows that the tracking error with the FLC approach zero while it increases with time with the PID without disturbance. The increase in the tracking error with time is an indication of instability in the case of the PID scheme. Figure 4 gives the resulted integral tracking error (ITE) and integral time tracking error (ITTE) as 80.8103 and 323.2413, and 1.7973 and 7.1901 accordingly with the PID and FLC without disturbance, respectively. Therefore, it can be seen that the PID has an estimated cumulative error of about 44 times higher than with the FLC.

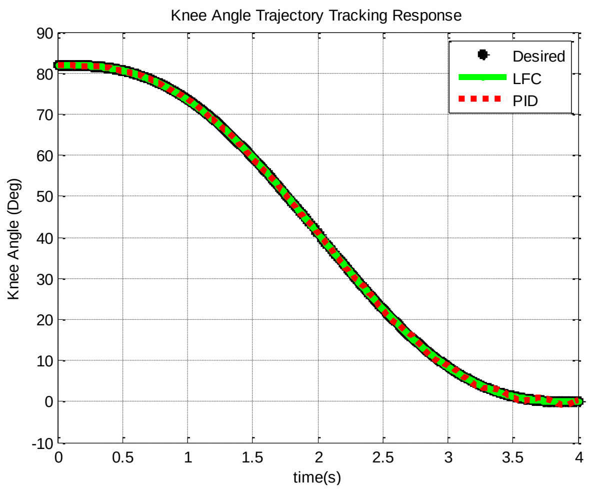

Figure 2. Tracking response without disturbance

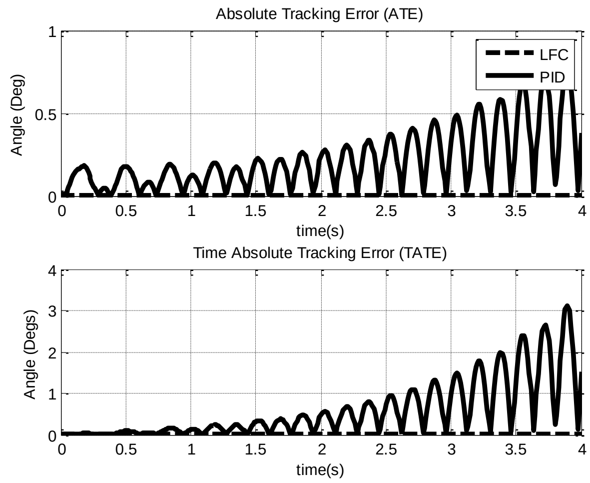

Figure 3. Tracking errors without disturbance

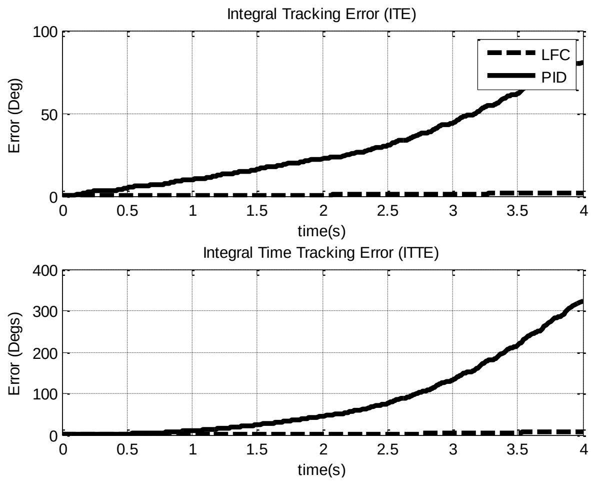

Figure 4. Integral tracking errors without disturbance

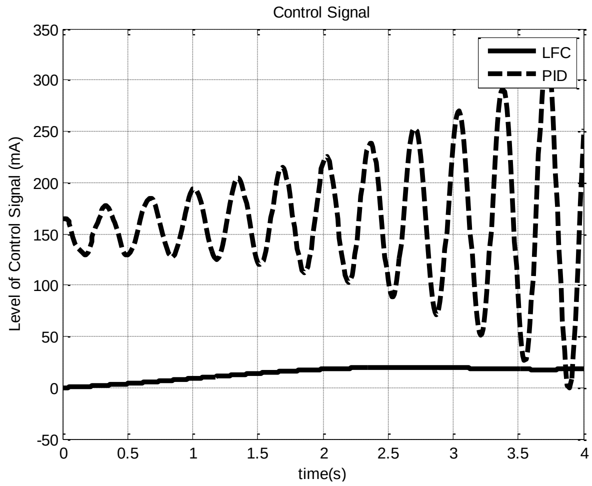

Figure 5. Control signals without disturbance

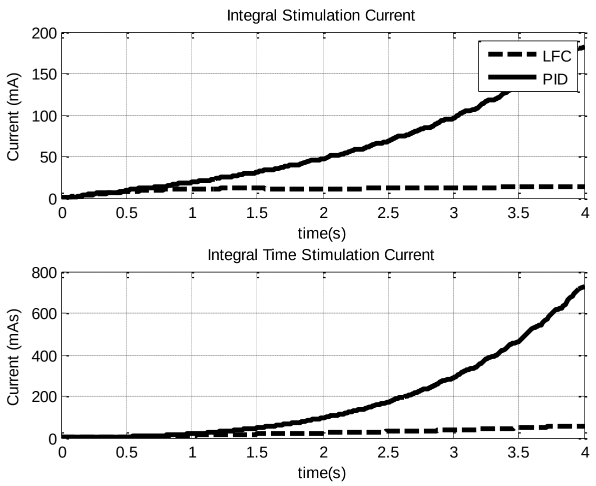

Figure 6. Integral of stimulation currents without disturbance

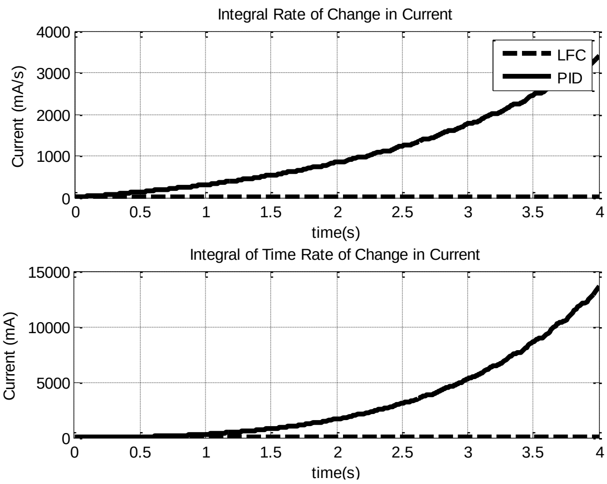

Figure 7. Integrals of the rate of change in stimulation current

As shown in Figure 5, the maximum control signals (MCS) without disturbance are 316.5532 mA and 19.8116 mA with the PID and FLC, respectively. Hence, the PID requires 16 times higher amount of stimulation current. It also implies that the peak stimulation current is below the threshold with the FLC, but it is about three times greater with the PID. Hence, this clearly portrays the unsuitability of the PID for the application.

The integrals of the stimulation current (ISC) and the integrals of the time stimulation current (ITSC) are 182.1604 mA and 728.6414 mAs, and 14.4031 mA and 57.6123 mAs with the PID and FLC respectively without controller, as shown in Figure 6. It means the total amount of stimulation current required using the PID is 12 times higher than that of the FLC and hence portrays another advantage of the FLC method. Figure 7 gives the integrals of the rate of change in stimulation current

(IRCSC) and the integrals of the rate of change in time stimulation current (IRCTSC) as 3405.1000 mA/s and 13620.0000 mA, and 22.6749 mA/s and 90.6998 mA. It indicates that the rate of change in the stimulation current is around 149 times greater than that of the FLC. This is also another desirable attribute of the FLC. High levels of the changes in the stimulation current produce earlier exhaustion of the subjects.

Figures 8-13 portray the tracking response, tracking errors, integrals of tracking errors, control signals, integrals of the stimulation currents and the integrals of the rate of change in the stimulation currents with 53% mass. Figure 8 shows the tracking response, which resulted in the RMSEs of 76.2000 and 0.0487 with the PID and FLC, respectively. It shows that an average tracking error, which is 1564 worse compared to using the FLC, resulted with the PID. Figure 9 is an illustration of the tracking errors. It can be inferred as the previous situation where the errors with the FLC diminish with time and it is the opposite in the case the PID and it entails system instability. The ITE and ITTE values are given by Figure 10, which are 37.8825 and 151.5301 respectively with the PID. They are 0.2954 and 1.1815 respectively in the case of the FLC. The results show that the total amount of the average tracking error resulted in the FLC is 127 times better than that of the PID.

Figure 11 is the MCSs with the PID and the FLC control methods. The peak values of the stimulation currents used during the transition obtained are 72.0630 mA and 10.5305 mA with the PID and FLC accordingly. The PID utilises 5.8 times higher amount of current than the FLC, the amount for both the control scheme are below the threshold. Figure 12 gives the ISCs and ITSCs for both control methods harnessed. Their quantities obtained are 54.5686 mA, 1218.2742 mAs, and 5.9543 mA, 283.8172 mAs with the PID and FLC respectively. Hence, the PID control requires 8 times the amount of the stimulation current greater than the FLC. The IRCSCs and IRCTSCs are the plots in Figure 13, which are 1353.3000 mA/s and 5413.3000 mA respectively with the PID and 12.0013 mA/s and 48.0051 mA respectively with the FLC. It implies that the cumulative amount of the rate of change in the stimulation current with the PID is about 111 times bigger than that of the FLC.

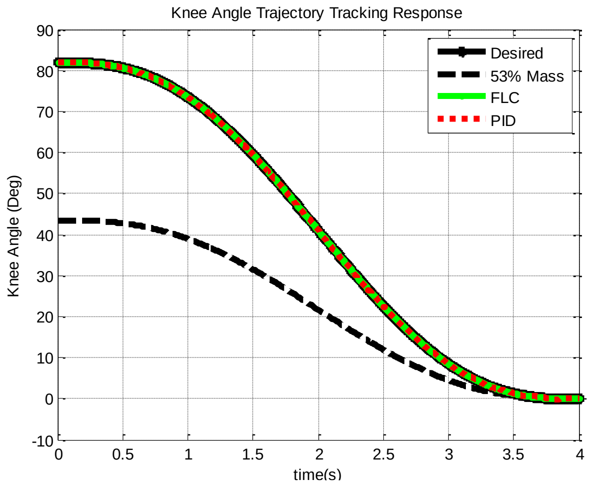

Figure 8. Tracking errors with 53% mass

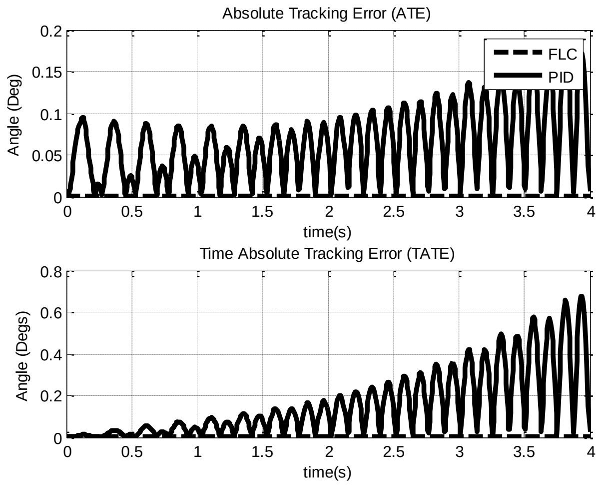

Figure 9. Tracking errors with 53% mass

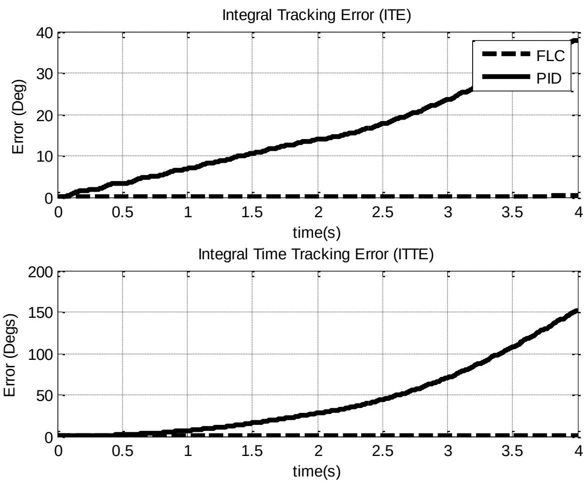

Figure 10. Integral tracking errors with 53% mass

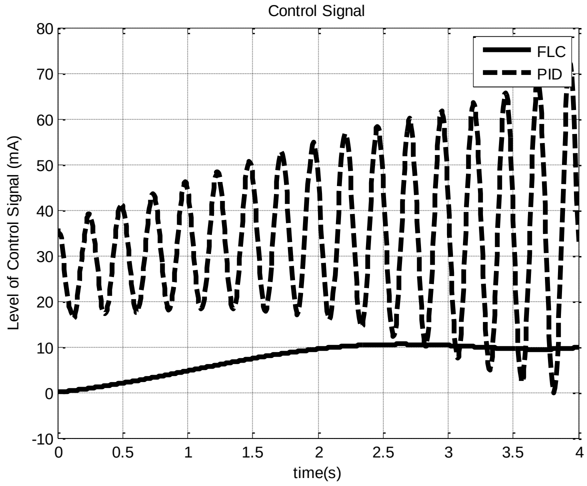

Figure 11. Control signals with 53% mass

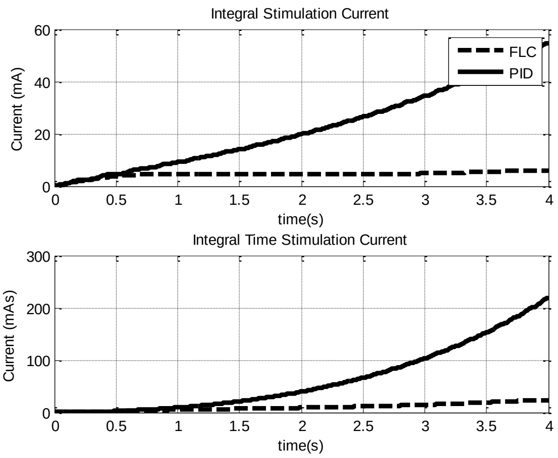

Figure 12. Integrals of stimulation currents with 53% mass

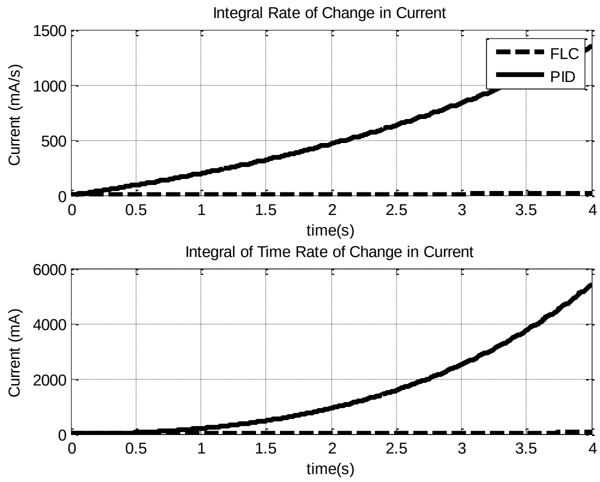

Figure 13. Integrals of the time rate of change in stimulation currents with 53% mass

Figures 14-19 are graphs showing the tracking response, tracking errors, integral tracking errors, control signals, integral stimulation currents and integral time stimulation currents, respectively with 126% mass. Figure 14 shows the disturbed response with the PID controller and the FLC. An average tracking error approximated by the RMSE is obtained as 405.1000 and 0.0487 accordingly with the PID and FLC. It shows that the PID control is associated with 340 times error in comparison to the FLC. The tracking errors, as shown in Figure 15, are similar to the previous ones. In the case of the FLC, it dies off while for the PID based gain scheduling controller rises with time. Figure 16 is the cumulative tracking errors ITE/ITTE, which in the case of the PID are 100.9881 and 403.9524 while they are 0.2954 and 1.1815 for the FLC. Hence, the total tracking error of the PID based control method was 340 times greater than that of the FLC.

As shown with MCS in Figure 17, the control signal plots are 517.2153 mA and 25.0346 mA with the gain scheduling based PID controller and the FLC, respectively. It implies that about 20 times higher amount of the maximum stimulation would be harnessed with the PID compared to the FLC. That of the PID is even by far beyond the recommended value. Figure 18 is the graph of the cumulative stimulation currents with ISC and ITSC. Their quantities are deduced from the plots as 269.6443 mA and 1078.6000 mAs, and 14.1555 mA and 56.6218 mAs for the PID and FLC schemes respectively. Hence, the PID method consumes 18 times higher amount of stimulation current than the FLC. Figure 19 is the plot of the rate of change in the stimulation currents. The IRCSC and IRTCSC are 4315.7000 mA/s and 17263.0000 mA with the PID and they are 28.5138 mA/s and 114.0553 mA with the FLC. Therefore, the amount of the rate of change in the stimulation current was 150 times worse with the PID compared to the FLC.

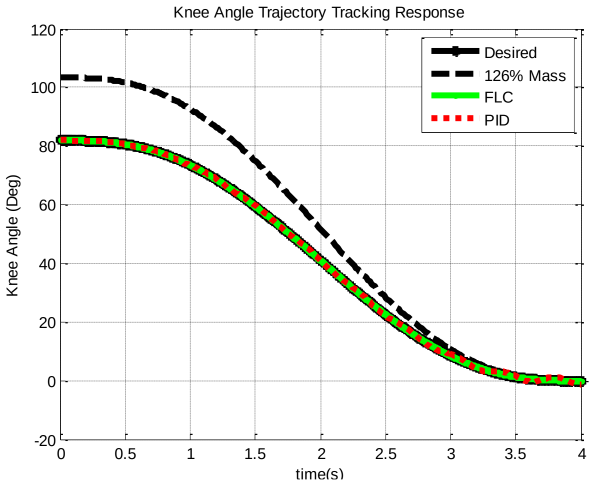

Figure 14. Tracking response with 126% mass

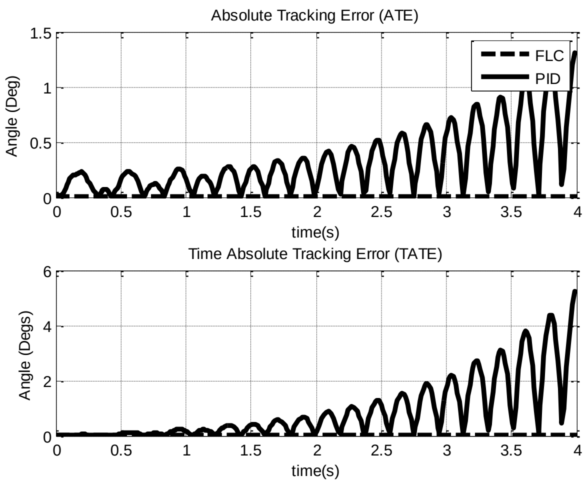

Figure 15. Tracking errors with 126% mass

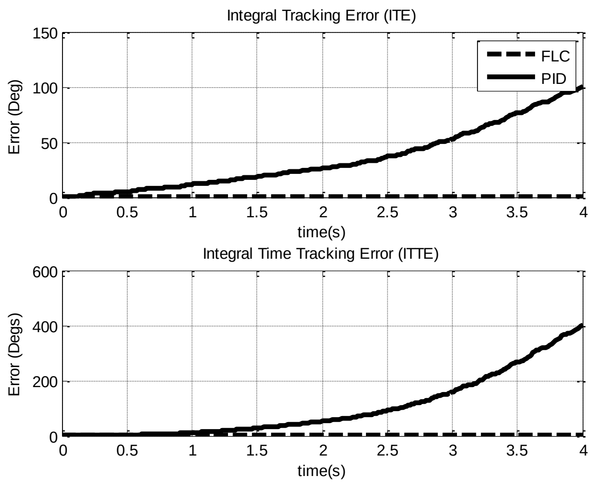

Figure 16. Integral tracking errors with 126% mass

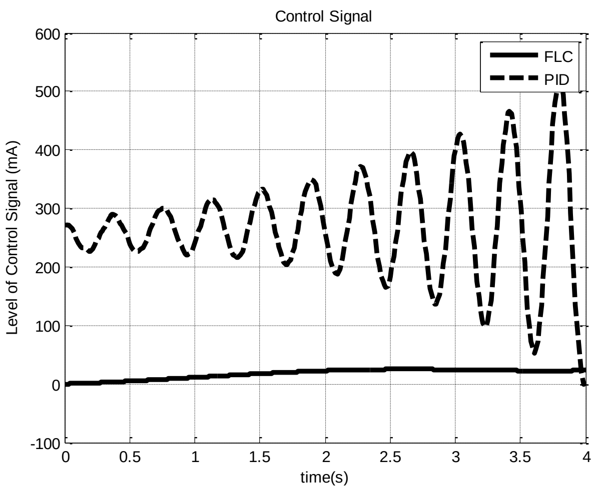

Figure 17. Control signals with 126% mass

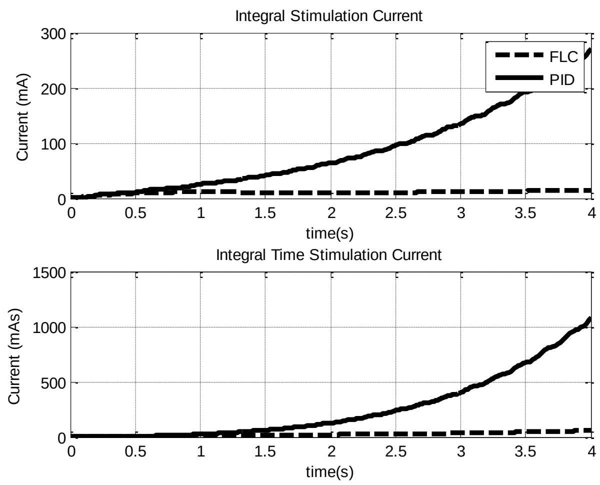

Figure 18. Integrals of Stimulation currents with 126% mass

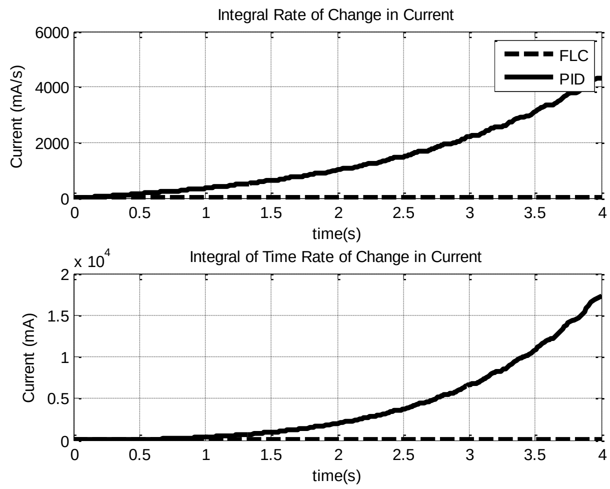

Figure 19. Integrals of the rate of change in stimulation currents with 126% mass

In terms of the recommended system specifications, the frequency is fixed, but the amount of stimulation current is varied in order to achieve the control objective. That is to make the movement as accurate as possible within recommended settings. The results without disturbance show that maximum amount the stimulation current utilised by the PID is about 16 times greater than of the FLC, and the current also exceeds the threshold. Examination of the response with a minimum amount of subject mass show that both the control schemes used amounts of currents less than the cut-off, but the PID utilised about 5.8 times higher than that of the FLC. The system with the peak amount of mass reveals that the PID consumed about 20 times that of FLC and also around four times greater than the threshold. In terms of the cumulative amounts of stimulation currents required as revealed using the ISC and ITSC, the margin is quite high in favour of the FLC. Results of other parameters obtained also revealed the remarkable margin regarding the system performance. That is in terms of tracking accuracies as portrayed by RMSE, ITE and ITTE. The possibilities of frequency changes which could be a function of the rate of change in the stimulation current have the negative effect of causing early fatiguing of the subjects. It is evaluated by harnessing the IRCSC and IRTCSC. Hence, the FLC performed remarkably well compared to the PID.

4. CONCLUSION

The application of the FLC for compensation of human mass variation in the revival of STS movement function utilising the FES was successfully presented. Results showed improvement in the system by making it more accurate and robust, which would help in making such devices to attain clinical acceptance. Literature however, indicates that the enhanced control system is required in attaining the desired goal, and the FLC was evaluated on the effect of global mass distribution. Results show remarkable improvements and hence, there was a substantial decrease in the tracking error, stimulation current and the rate of change in stimulation current, and all these are favourable for the application.

ACKNOWLEDGEMENT

Special gratitude to the Centre for Graduate Studies, Universiti Tun Hussein Onn Malaysia, Research Management Office Universiti Tun Hussein Onn Malaysia, Abubakar Tafawa Balewa University Bauchi Nigeria and Tertiary Education Fund Abuja Nigeria for their support.

REFERENCES

[1] Y. Chen, J. Hu, L. Peng and Z. Hou, The FES-assisted control for a lower limb rehabilitation robot: simulation and experiment, *Robotics and Biomimetics,* 1(2), 1-20, 2014.
[2] N. Arantes, D. Vaz, M. Mancini, M. Pereira, F. Pinto and T. Pinto, Effects of functional electrical stimulation applied to the wrist and finger muscles of hemiparetic subjects: a systematic review of the literature, *Brazilian Journal of Physical Therapy,* 11, 419-427, 2007.
[3] A. Papachristos, Functional electrical stimulation in paraplegia, in *Topics in Paraplegia,* Y. Dionyssiotis, Ed. London: IntechOpen, 2014.
[4] G. P. Braz, M. Russold and G. M. Davis, Functional electrical stimulation control of standing and stepping after spinal cord injury: A review of technical characteristics, *Neuromodulation: Technology at the Neural Interface,* 12, 180-190, 2009.
[5] Memo for Neuromuscular Electrical Stimulation (NMES) for Spinal Cord Injury (Website). http://www.cms.gov/medicare-coverage-database/details/Decision (accessed 24.11.2015).
[6] J. Villagra and C. Balaguer, A model-free approach for accurate joint motion control in humanoid locomotion, *International Journal of Humanoid Robotics,* 8, 27-46, 2011.

[7] S. H. Sadati, S. E. Naghibi, I. D. Walker, K. Althoefer and T. Nanayakkara, Control space reduction and real-time accurate modeling of continuum manipulators using ritz and ritz–galerkin methods, *IEEE Robotics and Automation Letters,* 3, 328-335, 2018.

[8] A. Tsukahara, R. Kawanishi, Y. Hasegawa and Y. Sankai, Sit-to-stand and stand-to-sit transfer support for complete paraplegic patients with robot suit HAL, *Advanced Robotics,* 24, 1615-1638, 2010.

[9] A. Fattah, M. Hajiaghamemar and A. Mokhtarian, Design of a semi-active semi-passive assistive device for sit-to-stand tasks, *Proceedings of 16th Annual (International) Conference on Mechanical Engineering (ISME2008),* Kerman, 2008, pp. 1-6.

[10] N.-Y. Yu, J.-J. J. Chen, and M. S. Ju, Closed-loop control of quadriceps/hamstring activation for FES-induced standing-up movement of paraplegics, *Journal of Musculoskeletal Research,* 5, 173-184, 2001.

[11] F. Previdi, M. Ferrarin, S. M. Savaresi, and S. Bittanti, Closed-loop control of FES supported standing up and sitting down using Virtual Reference Feedback Tuning, *Control Engineering Practice,* 13, 1173-1182, 2005.

[12] M. S. Poboroniuc, New experimental results on feedback control of FES-based standing in paraplegia, *AL.I.CUZA University Scientific Annals of Biophysics, Medical Physics and Environment Physics,* 3, 83-89, 2007.

[13] C. Lynch and M. Popovic, Closed-loop control for FES: Past work and future directions, *Proceedings of 10th Annual Conference of the International FES Society*, Montreal, 2005, pp. 2-4.

[14] C. L. Lynch and M. R. Popovic, A comparison of closed-loop control algorithms for regulating electrically stimulated knee movements in individuals with spinal cord injury, *IEEE Transactions on Neural Systems and Rehabilitation Engineering,* 20, 539-548, 2012.

[15] R. Davoodi and B. J. Andrews, Computer simulation of FES standing up in paraplegia: A self-adaptive fuzzy controller with reinforcement learning, *IEEE Transactions on Rehabilitation Engineering,* 6, 151-161, 1998.

[16] R. Davoodi and B. J. Andrews, Optimal control of FES-assisted standing up in paraplegia using genetic algorithms, *Medical Engineering & Physics,* 21, 609-617, 1999.

[17] R. Hussain, R. Massoud and M. Al-Mawaldi, ANFIS-PID control FES-supported sit-to-stand in paraplegics:(Simulation Study), *Journal of Biomedical Science and Engineering,* 7, 208-217, 2014.

[18] R. Massoud, The influence of control design on energetic cost during FES induced sit-to-stand, *Journal of Biomedical Science and Engineering,* 7, 1096-1104, 2014.

[19] T. Afzal, L. Khan and M. Tokhi, Simulation of a patient driven strategy for FES supported sit-to-stand movement, *Proceedings of 2010 International Conf. Information and Emerging Technologies (ICIET),* Karachi, 2010, pp. 1-5.

[20] M. Huq and M. Tokhi, Genetic algorithms based approach for designing spring brake orthosis–Part II: Control of FES induced movement, *Applied Bionics and Biomechanics*, 9, 317-331, 2012.

[21] R. M. Esfanjani and F. Towhidkhah, Application of nonlinear model predictive controller for FES-assisted standing up in paraplegia, *Proceedings of 27th Annual Conference on Engineering in Medicine and Biology Society IEEE-EMBS 2005,* Shanghai, 2006, pp. 6210-6213.

[22] M. Ferrarin and A. Pedotti, The relationship between electrical stimulus and joint torque: A dynamic model, *IEEE Transactions on Rehabilitation Engineering*, 8, 342-352, 2000.

[23] RehaMove, *FES Cycling with RehaMove*, ed. Germany: Hasomed, 2014.

[24] RehaMove, *Functional Electrical Stimulation, FES Applications*, ed: Hasomed, 2015.

[25] J. J. E. Slotine and W. Li, *Applied Nonlinear Control*, Englewood Cliffs, NJ: Prentice Hall, 1991.

[26] H. K. Khalil, *Nonlinear Systems*, 3rd ed., Upper Saddle River, NJ: Prentice Hall, 2002.

[27] C. D. Fryar, Q. Gu, C. L. Ogden, and K. M. Flegal, Anthropometric reference data for children and adults: United States, 2011-2014, *Vital and Health Statistics*, 3(39), 1-38, 2016.

[28] S. C. Walpole, D. Prieto-Merino, P. Edwards, J. Cleland, G. Stevens and I. Roberts, The weight of nations: an estimation of adult human biomass, *BMC Public Health*, 12(439), 1-6, 2012.

[29] A. K. Chang and J. Y. Choi, Factors influencing BMI classifications of Korean adults, *Journal of Physical Therapy Science,* 27, 1565-1570, 2015.