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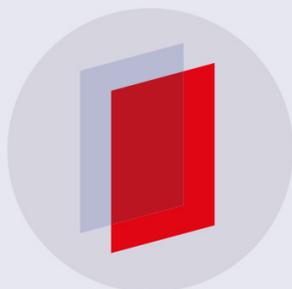
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Restoration of Movement using FES: An Introductory Study I

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Abstract. FES has been applied for movement restoration, rehabilitation and therapy in spinal cord injury and nervous system failure subjects, whose number rise worldwide every year. Despite the increase, assist devices that are expected to aid healthcare delivery are not in abundance and could be as a result of standards imposed due to the sensitive condition of the subjects. Although closed loop control systems are expected to positively aid in that regards, the delicacy of the plant was a big constraint. An existing model from the works Ferrarin and Pedotti was elaborated from the knee swinging to the sit-to-stand movements and from the two mathematical models it can be inferred that even though similar higher level of excitation would be required for sit-to-stand manoeuvre. As part of research to improve on closed approach for the FES system, an appraisal was successfully done on the neuromuscular model together with fatigue models from the works on Lynch. The remodelled fatigue models to suit the research were presented.

1. Introduction

Functional electrical stimulation (FES) is a technique used for restoration of movements in humans lost due to health problems in the nervous system. Causes of the complications could be accidents or diseases [1-6]. The technique harnesses electrical signals of suitable form which are passed to appropriate muscles to trigger the desired movement [2, 4, 7, 8]. FES are also used for rehabilitation (for movement as well as gait restoration) and therapy [9-11].

Records showed global growth in the number of the patients annually [12, 13] without corresponding increase in assistive devices which could be due to laws imposed (www.cms.gov/CAG-00153R). Earlier devices in existence employ the open loop control systems. The predefined fixed nature, dependency on the subject for initiations of signals are among the properties of such systems that could be tackled with closed loop control systems. And were identified as promising tools (has the potential for solution for part of stringent requirements and old systems) towards reaching clinical approval for lower limb FES assisted movements [14-16].



The report basically is an initial stage of research which directed towards refining of models that could be used with relative ease for closed loop control designs in restoration of motion using FES in spinal cord injury subjects (paraplegics). It also examines the torque developed during stimulation and the effect of fatigue which is one of the major disturbances affecting accomplishment of the task. The remodeled fatigue model to be adopted for the research was also presented. The manuscript has five sections: introduction, movement models, fatigue effects, discussion and conclusion. The introduction provides information regarding background of the study and the second section discusses the transfer functions for both FES induced knee swinging and sit-to-standing movements from the experimental work conducted by Ferrarin and Pedotti. Next was the fatigue effects which briefly explains the Lynch models, the proposed approximation and the effect of fatigue in performing the task. The chapter on discussion was brief examination on the FES movement models, fatigue models and disturbances effects, while the conclusion rounds up major indices of the study.

2. Movement models

The Ferrarin's model describes the dynamics of the FES-induced movements of the knee joint, obtained from data obtained from both paraplegic and healthy subjects [17]. The knee angle θ during the FES-induced free swinging movement is equivalent to the thigh angle during FES-assisted sit-to-stand movement, so it is also denoted θ . The major difference during swinging opposite repetition of movement occur due to action of gravity which may even make the dynamics overshoot its initial resting position. Of course the action of gravity could result in reverse movement but not beyond the initial resting position which is sitting for the sit-to-stand, but is not part of interest for now. The FES sit-to-stand model is hypothesized based on the same assumptions in Ferrarin and Pedotti FES knee swinging model [17], that the ankle joint was fixed. Other assumptions were that the feet remain stationary and since same quadriceps muscles are stimulated, the damping, stiffness as well as the response of muscle to stimulator signals are same.

2.1. The FES-Induced Knee Swinging Movement

Figure 1 describes arrangement used which also gave insight on the FES knee swinging dynamics. The extension of the knee is due to the stimulation and hence the torque developed by the shank ($\tau_{Shank}(t)$) is the summation of the torques developed due to the stimulation ($\tau_A(t)$), gravity ($\tau_G(t)$), damping ($\tau_D(t)$) and stiffness ($\tau_S(t)$) [18-21]. Equation (1) describes the statement.

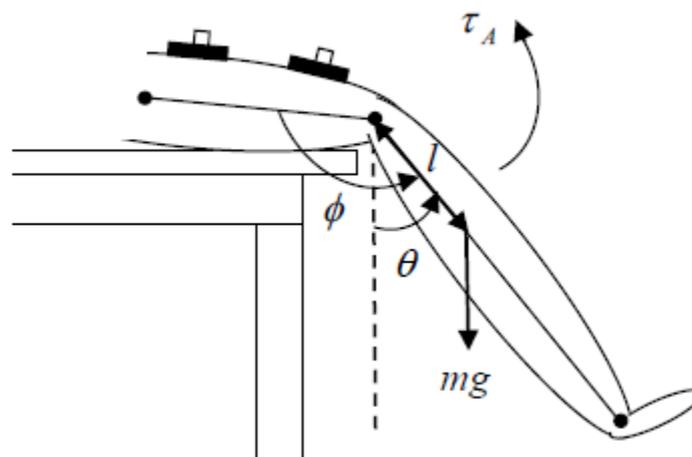


Figure 1: Description of the FES-induced knee swinging

$$\tau_{Shank}(t) = \tau_A(t) + \tau_G(t) + \tau_D(t) + \tau_S(t) \quad (1)$$

Flexion occur under the influence of gravity and therefore becomes the torque developed by the shank (τ_{Shank}) is the summation of the torques developed gravity (τ_G), damping (τ_D) and stiffness (τ_S), as illustrated by (2) [18-21].

$$\tau_{Shank}(t) = \tau_G(t) + \tau_D(t) + \tau_S(t) \quad (2)$$

$$\tau_{Shank}(t) = I \ddot{\theta} \quad (3)$$

$$\tau_A(s) = \frac{P_w(s)G}{1 + s\tau} \quad (4)$$

$$\tau_G(t) = mgl \sin \theta \quad (5)$$

$$\tau_D(t) = \beta \dot{\theta} \quad (6)$$

$$\tau_S(t) = \lambda e^{-E\left(\theta + \frac{\pi}{2}\right)} \left(\theta + \frac{\pi}{2} - \omega \right) \quad (7)$$

The parameters: θ is the knee angle, I the moment inertia of the combination of the shank and foot, $\dot{\theta}$ the angular velocity of the combination of the shank and foot, $\ddot{\theta}$ the angular acceleration of the combination of the shank and foot, θ_{TS} the angle between the thigh and shank, β the coefficient of viscous friction, g acceleration due to gravity, m the mass of the combination of the shank and foot, P_w the stimulating signal (Pulse Width Modulated) and l the distance between the knee and the centre of mass the combination of the shank and foot.

$$\text{Let } x_1 = \theta \quad (8)$$

$$x_2 = \dot{\theta} = \dot{x}_1 \quad (9)$$

$$x_2 = \ddot{\theta} \quad (10)$$

$$x_3 = \tau_A \quad (11)$$

$$x_3 = \dot{\tau}_A \quad (12)$$

$$\text{From (4); } \dot{\tau}_A = \frac{GP_w}{\tau} + \frac{\tau_A}{\tau} \quad (13)$$

Substituting (3)-(6) in (1) and making $\ddot{\theta}$ subject;

$$\ddot{\theta} = \frac{\tau_A}{I} - \frac{mgl \sin \theta}{I} - \frac{\beta \dot{\theta}}{I} - \frac{\lambda e^{-E\left(\theta + \frac{\pi}{2}\right)} \theta}{I} - \frac{\lambda e^{-E\left(\theta + \frac{\pi}{2}\right)} \left(\frac{\pi}{2} - \omega\right)}{I} \quad (14)$$

$$\text{Let } g(x_1) = -\frac{mgl \sin \theta}{I} - \frac{\lambda e^{-E\left(\theta + \frac{\pi}{2}\right)} \theta}{I} - \frac{\lambda e^{-E\left(\theta + \frac{\pi}{2}\right)} \left(\frac{\pi}{2} - \omega\right)}{I} \quad (15)$$

$$\ddot{\theta} = \frac{\tau_A}{I} + g(x_1) - \frac{\beta \dot{\theta}}{I} \quad (16)$$

$$\begin{pmatrix} \dot{x}_1 \\ \dot{x}_2 \\ \dot{x}_3 \end{pmatrix} = \begin{pmatrix} 0 & 1 & 0 \\ g(x_1) & -\frac{\beta}{I} & \frac{1}{I} \\ 0 & 0 & -\frac{1}{\tau} \end{pmatrix} \begin{pmatrix} x_1 \\ x_2 \\ x_3 \end{pmatrix} + \begin{pmatrix} 0 \\ 0 \\ \frac{G}{\tau} \end{pmatrix} P_w \quad (17)$$

$$\theta = \begin{bmatrix} 1 & 0 & 0 \end{bmatrix} \begin{pmatrix} x_1 \\ x_2 \\ x_3 \end{pmatrix} \quad (18)$$

Equations (17) and (18) forms the state space transfer function of the Ferrarin and Pedotti knee joint model. Note that; the model is nonlinear due the presence of the following terms: $\sin \theta$, $e^{-E\left(\theta+\frac{\pi}{2}\right)}$ terms, so also the τ_A (i.e. = $P_w G e^{-m}$; from (4)) even though is somehow hidden in the final model, but in reality its dynamics is nonlinear.

Linearization is achieved using the well-known and popular linearization technique which also used in control systems, achieved representing the function in the Taylor's series form as shown in equation (19). At equilibrium, x_0 , $A(x)$ equals zero. (That is $A(x_0) = 0$) and the higher order terms are neglected. Hence, everything is approximated as shown in equation (20) [22-24].

$$\dot{x} = \left| A(x) + \frac{f' A(x)}{1!} (x - x_0) + \frac{f'' A(x)}{2!} (x - x_0)^2 + \dots \right|_{x=x_0} \quad (19)$$

$$\dot{x} = \left[f' A(x)(x - x_0) \right]_{x=x_0} \quad (20)$$

The linearized transfer function becomes as shown in (21) and (22), upon substituting the parameters of the paraplegic subject three (P3) from the work of Ferrarin and Pedotti [18] as shown in table 1.

$$\begin{pmatrix} \dot{x}_1 \\ \dot{x}_2 \\ \dot{x}_3 \end{pmatrix} = \begin{pmatrix} 0 & 1 & 0 \\ -41.82 & -0.73 & 2.54 \\ 0 & 0 & -4.65 \end{pmatrix} \begin{pmatrix} x_1 \\ x_2 \\ x_3 \end{pmatrix} + \begin{pmatrix} 0 \\ 0 \\ 0.07 \end{pmatrix} P_w \quad (21)$$

$$\theta = \begin{bmatrix} 1 & 0 & 0 \end{bmatrix} \begin{pmatrix} x_1 \\ x_2 \\ x_3 \end{pmatrix} \quad (22)$$

TABLE1. Parameters of the subject [18]

Parameter	Value
Moment inertia of shank and foot I (Kg m^2)	0.394
Mass of combined shank and foot m (Kg)	4.76
Distance between the knee and the center of mass combined shank and foot l (m)	0.233
Coefficient of exponential term λ (Nmrad $^{-1}$)	15.352
Coefficient of exponential term E (rad $^{-1}$)	1.644
Resting Elastic Knee angle ω (rad)	3.896

Parameter	Value
Time constant η (s)	0.215
Static gain G (Nms ⁻¹)	0.015
Stimulation frequency (Hz)	50
Mass of subject (Kg)	85
Age of subject (Yrs)	25
Height of subject (m)	1.78
Duration with injury (Yrs)	4
Level of leison	T3/T4
Moment of inertia for combined head, neck, trunk and thigh I_M (Kgm ²)	9.344
Mass of combined head, neck, trunk and thigh M (Kg)	49.13
Length of thigh L (m)	0.4361
Coefficient of viscous friction β (Nmsrad ⁻¹)	0.289
Mass of thigh m_t (kg)	8.50
Center of mass of thigh l_t (m)	0.2472

Changing to s-domain, which is obtained using;

$$\frac{\theta(s)}{P_w(s)} = c(sI - A)^{-1}B + D \quad (23)$$

$$\frac{\theta(s)}{P_w(s)} = \begin{bmatrix} 1 & 0 & 0 \end{bmatrix} \left(s \begin{bmatrix} 1 & 0 & 0 \\ 0 & 1 & 0 \\ 0 & 0 & 1 \end{bmatrix} - \begin{bmatrix} 0 & 1 & 0 \\ -41.82 & -0.73 & 2.54 \\ 0 & 0 & 4.65 \end{bmatrix} \right)^{-1} \begin{pmatrix} 0 \\ 0 \\ 0.07 \end{pmatrix} + \begin{pmatrix} 0 \\ 0 \\ 0 \end{pmatrix} \quad (24)$$

$$\frac{\theta(s)}{P_w(s)} = \frac{0.18}{s^3 + 5.38s^2 + 45.21s + 194.46} \quad (25)$$

Equation (25) becomes the transfer function in the s-domain.

2.2. The FES-Induced Sit-to-Stand Movement

Figure 2 describe the arrangement for the FES induced sit-to-stand movement which also gave insight on its dynamics.

In the case of FES-induced sit-to-stand the extension is the most important movement which result upon stimulation of the quadriceps muscles. The extension of the knee is developed due to the stimulation but the torque is experience by the thigh (τ_{Thigh}), the ankle joint and the feet are assumed to be fixed.

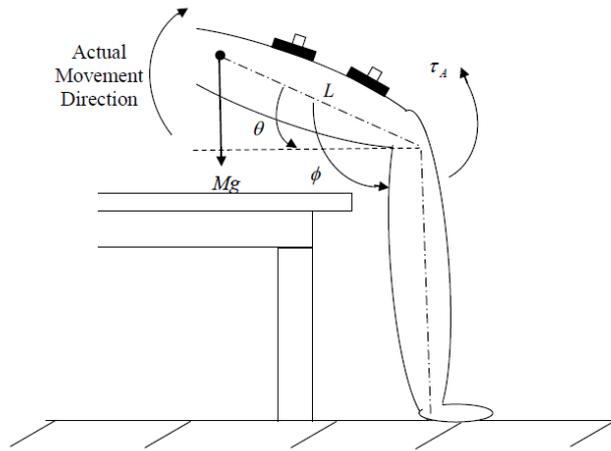


Figure 2: Description of the FES-induced sit-to-stand.

In the case of FES-induced sit-to-stand the extension is the most important movement which results upon stimulation of the quadriceps muscles. The extension of the knee is developed due to the stimulation but the torque resulted in the thigh ($\tau_{Thigh}(t)$), the ankle joint and feet are assumed to be fixed. The torque is the summation of the torques developed due to the stimulation ($\tau_A(t)$), gravity ($\tau_G(t)$), damping ($\tau_D(t)$) and stiffness ($\tau_S(t)$) [18-21]. Equation (26) is a description of the statement. The expressions for torques produced as a result of stimulation $\tau_A(s)$, damping $\tau_D(t)$ and stiffness $\tau_S(t)$ remain similar to that FES assisted knee swinging including the meanings of parameters involved and notations.

$$\tau_{Thigh}(t) = \tau_A(t) + \tau_G(t) + \tau_D(t) + \tau_S(t) \quad (26)$$

$$\tau_{Thigh}(t) = I_M \ddot{\theta} \quad (27)$$

$$\tau_A(s) = \frac{P_w(s)G}{1 + s\eta} \quad (28)$$

$$\tau_G(t) = MgL \sin(90 - \theta) + m_t g l_t \sin(90 - \theta) \quad (29)$$

$$\tau_D(t) = \beta \dot{\theta} \quad (30)$$

$$\tau_S(t) = \lambda e^{-E\left(\theta + \frac{\pi}{2}\right)} \left(\theta + \frac{\pi}{2} - \omega \right) \quad (31)$$

The parameters: θ is the knee angle during swinging movement which is equivalent to the thigh angle during sit-to-stand movement and in reality the angular movements are with the case of swinging as shown in the figure but only that the movement seen is the opposite, I_M the moment inertia of the combination of the head, neck and thigh, $\dot{\theta}$ becomes the angular velocity of the combination of the head, neck and thigh, $\ddot{\theta}$ the angular acceleration of the combination of the shank and foot, ϕ the angle between the thigh and shank, β the coefficient of viscous friction, g acceleration due to gravity, m

the mass of the combination of the shank and foot, M the mass of the combination of the head, neck, trunk and thigh, m_t the mass of the thigh, l_t is the centre of mass of the thigh, P_w the stimulating signal (Pulse Width Modulated) L is the length of the thigh. Please note that; the gravity torque component in addition to its effect on M it also act on m_t though not mentioned on the figure.

Let;

$$x_1 = \theta \quad (32)$$

$$x_2 = \dot{\theta} = \dot{x}_1 \quad (33)$$

$$x_3 = \ddot{\theta} \quad (34)$$

$$x_3 = \tau_A \quad (35)$$

$$x_3 = \tau_A \quad (36)$$

From(29)

$$\tau_A = \frac{GP_w}{\eta} + \frac{\tau_A}{\eta} \quad (37)$$

Substituting (27)-(31) in (26) and making $\ddot{\theta}$ subject;

$$\ddot{\theta} = \frac{\tau_A}{I_M} - \frac{MgL \cos \theta}{I_M} - \frac{m_t g l_t \cos \theta}{I_M} - \frac{\beta \dot{\theta}}{I_M} - \frac{\lambda e^{-E\left(\theta + \frac{\pi}{2}\right)} \theta}{I_M} - \frac{\lambda e^{-E\left(\theta + \frac{\pi}{2}\right)} \left(\frac{\pi}{2} - \omega\right)}{I_M} \quad (38)$$

$$h(x_1) = -\frac{MgL \cos \theta}{I_M} - \frac{m_t g l_t \cos \theta}{I_M} - \frac{\lambda e^{-E\left(\theta + \frac{\pi}{2}\right)} \theta}{I_M} - \frac{\lambda e^{-E\left(\theta + \frac{\pi}{2}\right)} \left(\frac{\pi}{2} - \omega\right)}{I_M} \quad (39)$$

$$\ddot{\theta} = \frac{\tau_A}{I_M} + g(x_1) - \frac{\beta \dot{\theta}}{I_M} \quad (40)$$

$$\quad (41)$$

$$\begin{pmatrix} \dot{x}_1 \\ \dot{x}_2 \\ \dot{x}_3 \end{pmatrix} = \begin{pmatrix} 0 & 1 & 0 \\ h(x_1) & -\frac{\beta}{I_M} & \frac{1}{I_M} \\ 0 & 0 & -\frac{1}{\eta} \end{pmatrix} \begin{pmatrix} x_1 \\ x_2 \\ x_3 \end{pmatrix} + \begin{pmatrix} 0 \\ 0 \\ \frac{G}{\eta} \end{pmatrix} P_w \quad (42)$$

$$\theta = [1 \quad 0 \quad 0] \begin{pmatrix} x_1 \\ x_2 \\ x_3 \end{pmatrix} \quad (42)$$

The linearized transfer function becomes as shown in (43) and (44), after substituting the parameters of the paraplegic subject three (P3) from the work of Ferrarin and Pedotti [18] as shown in table 1.

$$\begin{pmatrix} \dot{x}_1 \\ \dot{x}_2 \\ \dot{x}_3 \end{pmatrix} = \begin{pmatrix} 0 & 1 & 0 \\ -24.71 & -0.03 & 0.11 \\ 0 & 0 & -4.65 \end{pmatrix} \begin{pmatrix} x_1 \\ x_2 \\ x_3 \end{pmatrix} + \begin{pmatrix} 0 \\ 0 \\ 0.07 \end{pmatrix} P_w \quad (43)$$

$$\theta = [1 \ 0 \ 0] \begin{pmatrix} x_1 \\ x_2 \\ x_3 \end{pmatrix} \quad (44)$$

Changing to s-domain, which is obtained using;

$$\frac{\theta(s)}{P_w(s)} = c(sI - A)^{-1} B + D \quad (45)$$

$$\frac{\theta(s)}{P_w(s)} = [1 \ 0 \ 0] \left(s \begin{bmatrix} 1 & 0 & 0 \\ 0 & 1 & 0 \\ 0 & 0 & 1 \end{bmatrix} - \begin{bmatrix} 0 & 1 & 0 \\ -24.71 & -0.03 & 0.11 \\ 0 & 0 & 4.65 \end{bmatrix} \right)^{-1} \begin{pmatrix} 0 \\ 0 \\ 0.07 \end{pmatrix} + \begin{pmatrix} 0 \\ 0 \\ 0 \end{pmatrix} \quad (46)$$

$$\frac{\theta(s)}{P_w(s)} = \frac{0.0077}{s^3 - 4.62s^2 + 24.57s - 114.90} \quad (47)$$

Equation (47) becomes the transfer function in the s-domain.

It can be seen that more amount of torque would be required for moving the knee joint in order to achieve the goal which is standing. The effort from the upper limbs is used to provide the additional necessary support.

3. Effect of fatigue

Fatigue contributes to the nonlinear nature of the plant which is time variant in nature [20, 25, 26]. It is one of the disturbances intended to be suppressed as well as evaluated in the study. The model proposed by Lynch et al. [20, 25, 26] would be adopted which were obtained from experimental by the authors. The models were examples of the nature of the fatigue as shown in figure 3, figure 4 and figure 5, due to unavailability of the mathematical models for analysis remodeling was done.

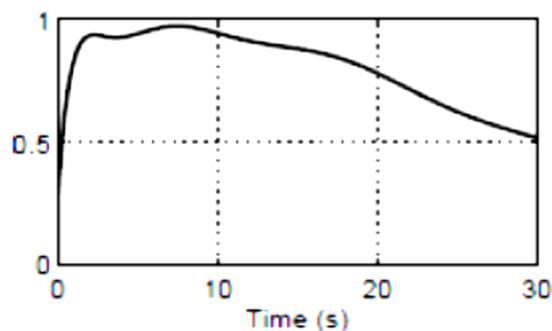


Figure 3: Mild fatigue [20].

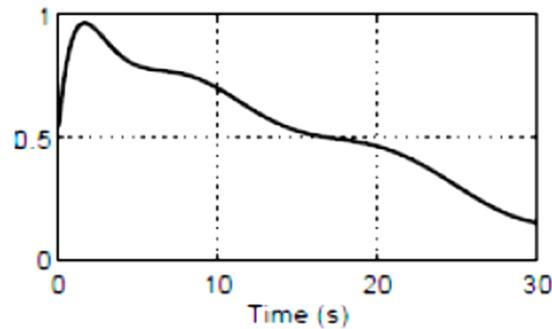


Figure 4: Moderate fatigue [20].

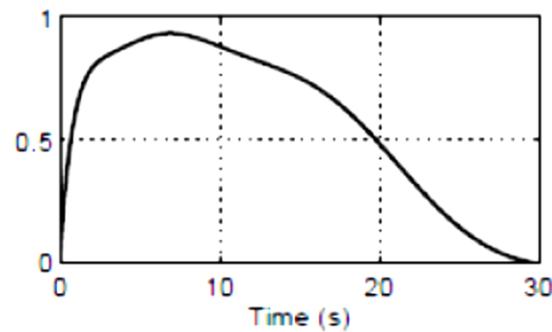


Figure 5: Severe fatigue [20].

The mild ($F_i(t)$), moderate ($F_o(t)$) and severe fatigue ($F_s(t)$) models were approximated using 6 order polynomials as given by (48), (49) and (50) with regression errors of 0.9655, 0.9951 and 0.9618 respectively.

$$F_i(t) = -0.0000001x^6 + 0.00001x^5 - 0.0004x^4 + 0.008x^3 - 0.0775x^2 + 0.3446x + 0.4125 \quad (48)$$

$$F_o(t) = -0.00000008x^6 + 0.000006x^5 - 0.0003x^4 + 0.006x^3 - 0.0597x^2 + 0.2834x + 0.4107 \quad (49)$$

$$F_s(t) = -0.00000006x^6 + 0.000007x^5 - 0.0003x^4 + 0.0058x^3 - 0.0579x^2 + 0.2219x + 0.6405 \quad (50)$$

4. discussion

As earlier mentioned the study is part of work towards developing a novel control scheme using sliding mode wavelet networks approach for FES induced movements in subjects with impaired nervous systems. A good representation (model) of the plant is required for such application. Considering the delicacy of the situation, it would be hectic to carry out measurements on each of such individuals which may even be impossible in some cases. The work tries to obtain suitable FES

induced sit-to-stand model (transfer function) for control design as well as analysis. It was based on experimental works of Ferrarin and Pedotti, where they developed the swinging model of the knee joint. An elaboration of the model was done and also extended for the sit-to-stand movement. In further studies, the model would be combined with fatigue, tremor and spasm models from the works of Lynch et al. for analysis.

Figures 6, 7, 8 and 9 portray the torques developed at frequencies of 20, 25, 33 and 50Hz respectively due to stimulation pulse widths of 50, 100, 150 and 200 μ s. It can be seen that the level of torques produced increased as the stimulation currents increase.

Figure 10, Figure 11, Figure 12 and Figure 13 show the torques developed with stimulation pulse widths of 50, 100, 150 and 200 μ s respectively at frequencies of 20, 25, 33 and 50Hz. It can be seen that the level of the torques produced increase as the stimulation frequencies increase.

Figure 14, Figure 16, Figure 18 and Figure 20 indicate the effect of attenuating disturbance at frequencies of 20, 25, 33 and 50Hz respectively on the system in open loop configuration. Figure 15, Figure 17, Figure 19 and Figure 21 express what happens when the closed loop system (step response) was employed. It can be seen that the level of error was more severe for the open loop.

Figure 22 was an indication of the response when the fatigue is the disturbance to the system. It can be seen that apart from reduction of the response, nonlinearity was introduced due to the nature of the perturbation.

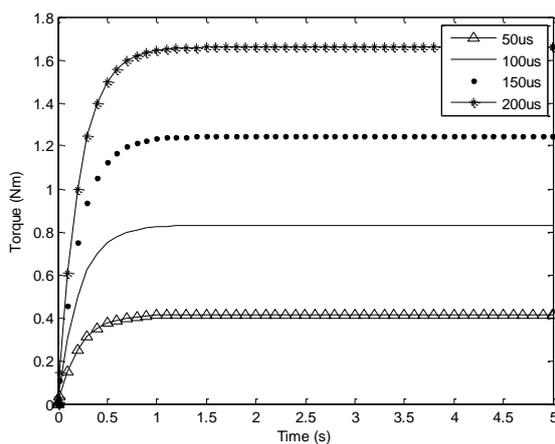


Figure 6: Torque, 20Hz and different currents.

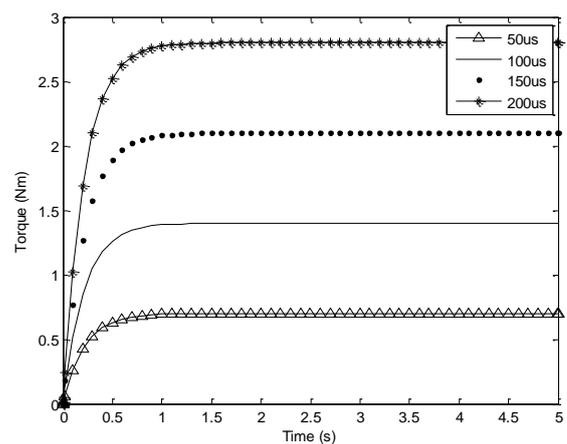


Figure 8: Torque, 33Hz and different currents.

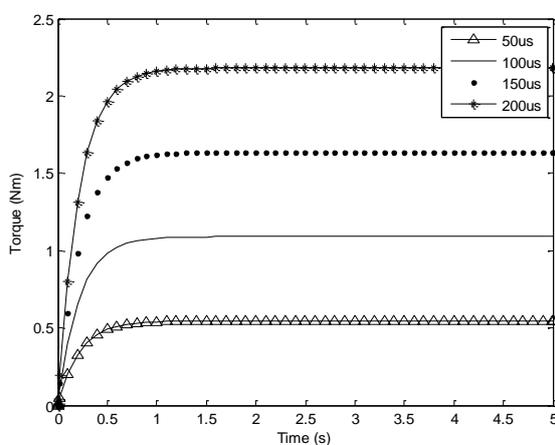


Figure 7: Torque, 25Hz and different currents.

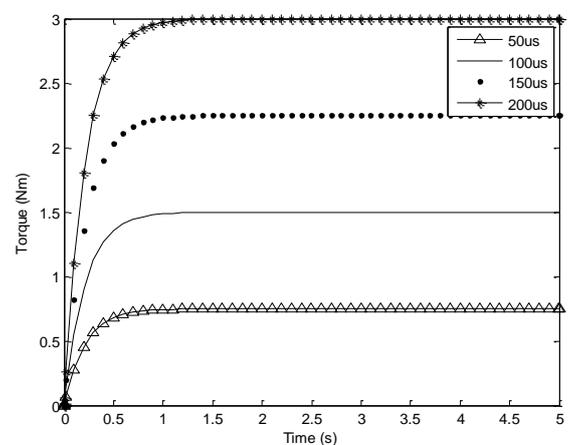


Figure 9: Torque, 50Hz and different currents.

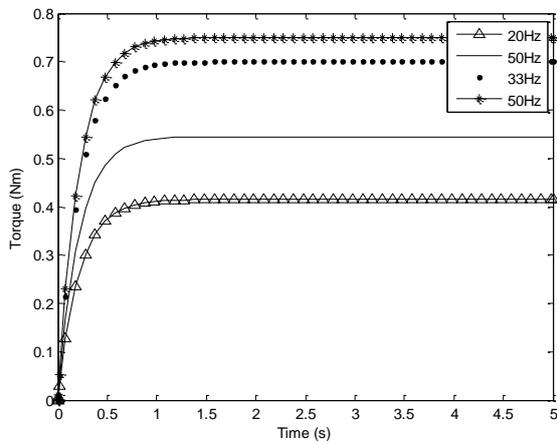


Figure 10: Torque, 50us PW and different frequencies.

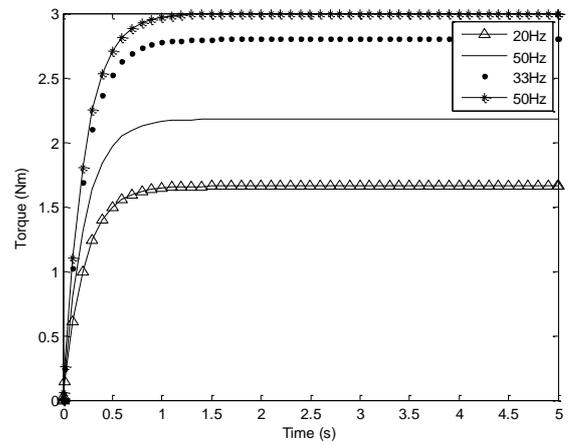


Figure 13: Torque, 200us PW and different frequencies.

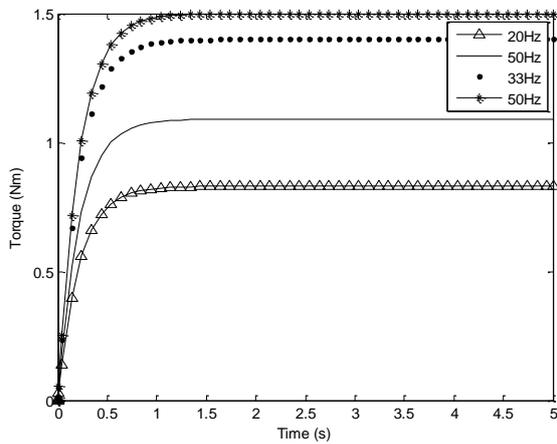


Figure 11: Torque, 100us PW and different frequencies.

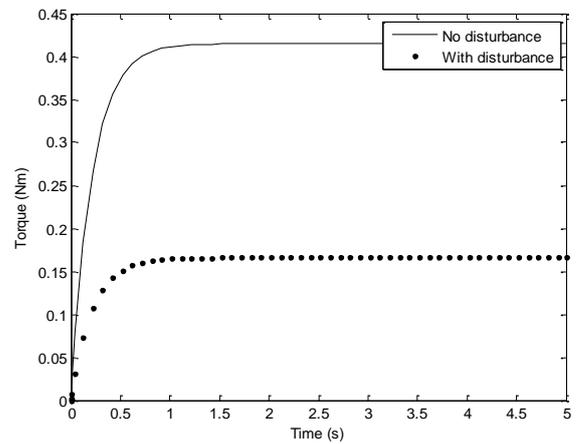


Figure 14: Effect of disturbance on torque 20Hz open loop.

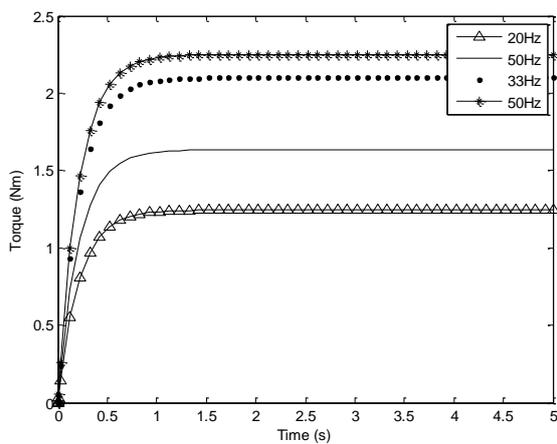


Figure 12: Torque, 150us PW and different frequencies.

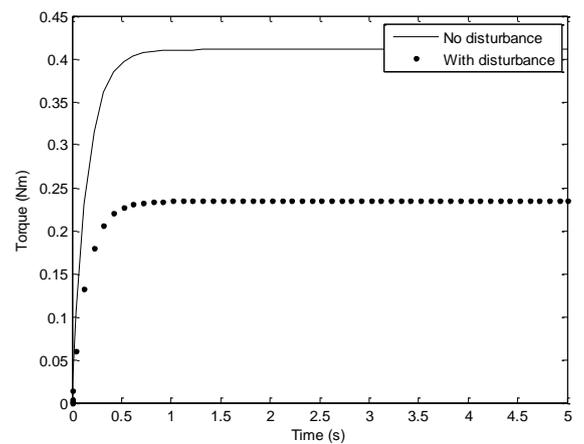


Figure 15: Effect of disturbance on torque 20Hz closed loop.

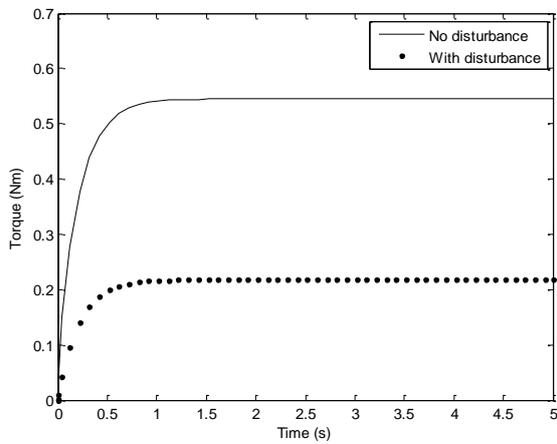


Figure 16: Effect of disturbance on torque 25Hz open loop.

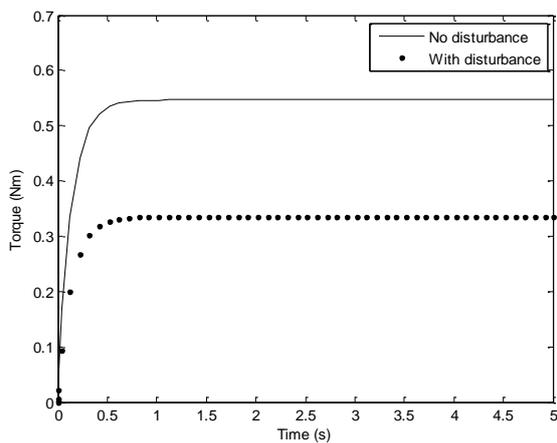


Figure 17: Effect of disturbance on torque 25Hz closed loop.

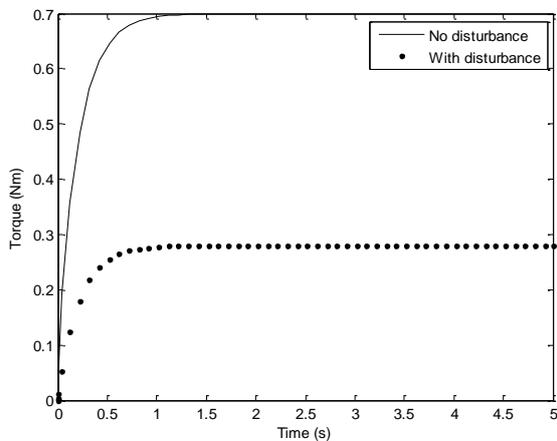


Figure 18: Effect of disturbance on torque 33Hz open loop.

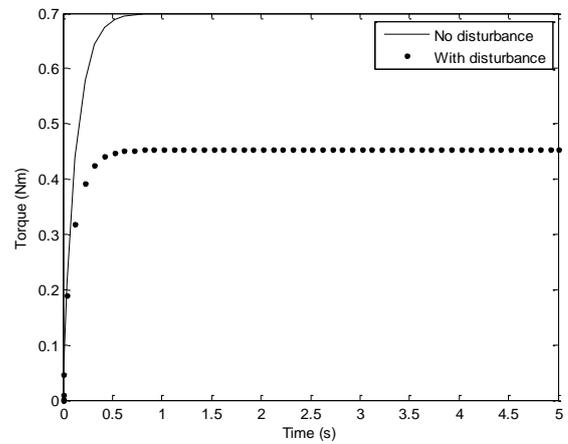


Figure 19: Effect of disturbance on torque 33Hz closed loop.

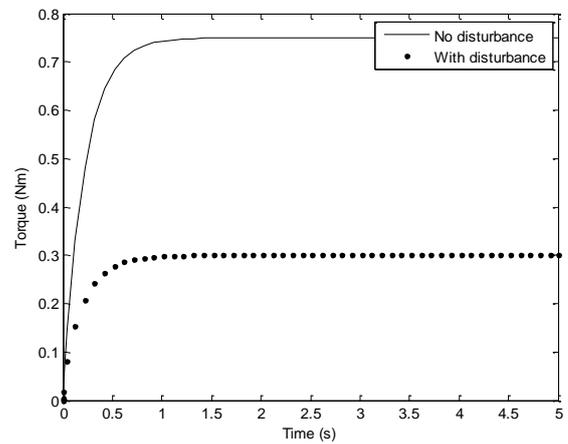


Figure 20: Effect of disturbance on torque 33Hz open loop.

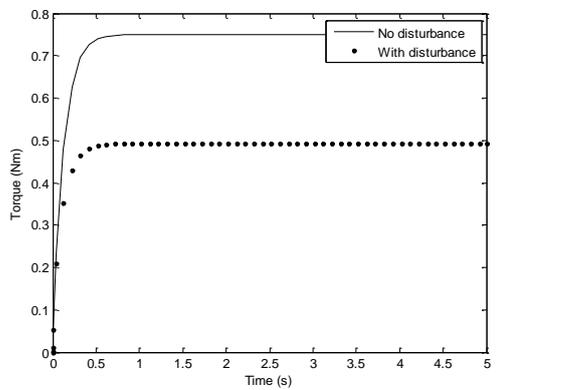


Figure 21: Effect of disturbance on torque 50Hz closed loop.

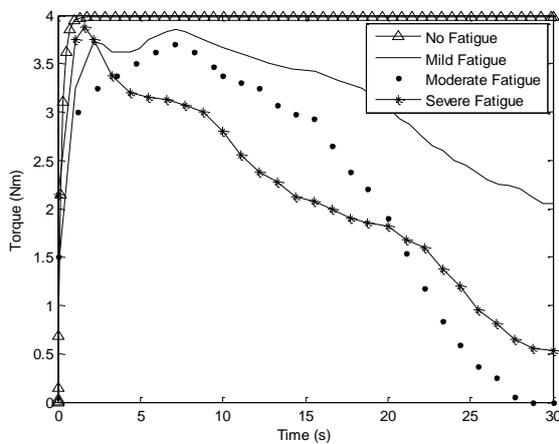


Figure 22: Effect of fatigue on torque developed.

5. Conclusion

FES is used for movement restoration, rehabilitation and therapy. Availability of abundant assist devices could improve healthcare delivery in the area which is currently lacking due to most likely standards compliance issues. Closed loop control systems could be harnessed to achieve that, but the plant is delicate and has to be carefully handled. Suitable modification of the Ferrarin and Pedotti knee swinging model was made in order to achieve the sit-to-stand movement. And from the models it can be seen that higher effort (stimulation) would be required for the sit-to-stand scenario.

The work is new in the sense that there is no known work to the best knowledge of the authors that: applied or extend the Ferrarin and Pedotti model for sit-to-stand, or portrays intention towards harnessing the model together with fatigue, tremor and spasm models proposed by Lynch, as well as the sensor error for robustness and stability analysis.

FES is a technique used for revival of movement in different forms but existing devices whose operation utilize that together with other gadgets are scarce. The effects of the stimulation current and frequency, closed loop arrangement and fatigue on the models were appraised and found to be in conformity with literature. The torque produced due to electrical stimulation was the main component in achieving the induced motion and is associated with fatigue. Fatigue causes decline in the torque produced as well as introduce nonlinearity in the behavior of the overall system dynamics.

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