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Input-Output Feedback Linearization Control (IOFLC) for muscle force control by functional electrical stimulation

Abdennacer Ben Hmed, Toufik Bakir, Anis Sakly, Stephane Binczak.

***Abstract*— Functional Electrical Stimulation (FES) is mainly used to help paralyzed individuals to restore functional movements. The efficiency of FES-systems is technically limited by the imprecise control of movements and also by the rapid appearing of the fatigue. Therefore, there is a need to include control strategies into FES-systems to modulate stimulation parameters (frequency, pulse amplitude and pulse duration). We aimed in this work to compute the required stimulation pulse duration using the Input-Output Feedback Linearization Control (IOFLC) in order to perform electrical stimulation by tracking a given desired force reference. The IOFLC approach is validated in this study through different simulation scenarios. The illustrated results showed promise in application of IOFLC into closed loop control in FES rehabilitation systems.**

***Index Terms*— Functional Electrical Stimulation, Muscle-force model, Feedback linearization, Pulse duration.**

I. Introduction

The potential objective of functional electrical stimulation (FES) is to enable paralyzed individuals to produce functional tasks. This neuroprosthesis method uses low-level electircal stimulation pulses with constant stimulation parameters (pulse amplitude, pulse duration, frequency) who builds a constant frequency train (CFT). The use of constant stimulation protocol can't take into consideration some additional problems to the FES applications such as the uncertainties and nonlinearities of muscle dynamics, the imprecise control of movements and the presence of the fatigue [1]. However, the design of an automatic stimulation strategy is necessary as it is important to adapt this typical stimulation to achieve precisely functional movements and to delay the onset of the muscle fatigue. Thus, achieving a desired muscle force response using adequate control strategies to determine the stimulation protocol could generate these precise movements.

Unlike open loop control technique that can be used to compute stimulation parameters, stimulation protocols based on feedback control is required to control the muscle force response and to obtain more precise movements. Furthermore, the use of mathematical models based on physiological steps under FES control can significantly enhance model-based controllers applied to the FES parameters. In the literature, there have been many physiological-based models designed to predict the muscle force response under a large variety of stimulation conditions. The most known of them is the model developed by Ding et al [1], [2]. This model was evaluated by Frey Law and Shields and compared to many others mathematical model [3]. The results presented in this evaluation showed that the Ding et al model provided the best fit of the force developed by the human quadriceps muscles.

For a long time, the control and the modeling of the muscle response (torque, knee angle joint, force) has received considerable attention from various research groups. In previous works, the majority of controllers are designed to adjust the stimulation intensity (pulse amplitude [4]–[7], pulse duration [8], [9]) and also to co-modulate the stimulation intensity parameters [10], [11]. In other case, a normalized input control was used to map the force-intensity relationship [12]. In this current study, we use the input-output feedback linearization control (IOFLC) applied to the Ding et al force model to control the muscle force output and we take the pulse duration as a control variable. The efficiency of the IOFLC is proved for many applications systems such as biological and medical systems [12], power electronics systems [13], [14] and chemical systems [15].

The main objective of this study focus in the controlling of the muscle force by adjusting the stimulation pulse duration. In this paper, we first describe the Ding et al force model. Then, the force-pulse duration relationship and the proposed normalized input control used to compute the pulse duration are developed. Subsequently, the IOFLC theory is recalled and its application to control the muscle force is designed and tested for a typical subject. Finally, the performance of the IOFLC is discussed relative to various simulation results.

II. Materials and Methods

A. Muscle force model

The Ding et al model considered to predict the muscle force consists of two differential equations [2]: the first one (Eq. (1)) represents the muscle activation and the second equation (Eq. (2)), the mechanical step, consists of the development of the muscle force:

$$\frac{dC_N}{dt} = \frac{1}{\tau_c}\sum_{i=1}^{n} R_i \exp(-\frac{t-t_i}{\tau_c}) - \frac{C_N}{\tau_c}, \qquad (1)$$

$$\frac{dF}{dt} = A\frac{C_N}{K_m + C_N} - \frac{F}{\tau_1 + \tau_2 \dfrac{C_N}{K_m + C_N}}. \qquad (2)$$

with,

$$R_i = \begin{cases} 1 & for \quad i = 1, \\ 1 + (R_0 - 1)\exp(-\frac{t_i - t_{i-1}}{\tau_c}) & for \quad i > 1. \end{cases}$$

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The term R_i is a scaling term that accounts for the nonlinear summation of the Ca^{2+} transient within the muscle fibers in responses to two closely spaced pulses [2], [16]. In Eqs (1) and (2), C_N and F are the two dependent state variables and the force output (F) is driven by six physiological parameters (τ_c, A, τ_1, τ_2, R_0, and K_m). More details about these parameters can be found in Table I.

The Ding et al force model was validated to predict the muscle force response from quadriceps muscles of healthy individuals [2] and of persons with spinal cord injuries (SCI) [16]. This model successfully predicted the force response to stimulation trains over a wide range of stimulation parameters (frequency, pulse duration, pulse pattern) under various physiological conditions (nonfatigued muscles, fatigued muscles).

B. Force-pulse duration relationship

To take into account the stimulation pulse duration effect, Ding et al extended their force model in [16] by adding a new component in the system (see Fig. 1 (red block)). They showed that the effect of varying the pulse duration on the force response could be accounted by modeling the parameter A in Eq. (2) as function of the pulse duration. In addition, an exponential equation was used to represent the force-pulse duration relationship and it is given by:

$$A = a^{'}\left(1 - e^{-\dfrac{pd - pd_0}{pd_t}}\right), \tag{3}$$

where $a^{'}$ is a scaling factor term for the force, pd is the stimulation pulse duration (pulse width) (μs), pd_0 is the threshold pulse duration (μs) and pd_t is the time constant controlling the rise of the A-pd relationship with increasing the pulse duration (μs).

TABLE I: Definition of the states and parameters of the muscle force model from Ding et al. [2], [16]

Symbol	Unit	description
Activation		
C_N	—	Normalized amount of Ca^{2+}-troponin complex
t_i	ms	Time of the i^{th} stimulation pulse
n	—	Total number of stimulation pulse before time t in the train
τ_c	ms	Time constant which commands the rise and the decay of C_N
R_0	—	Mathematical term characterizing the magnitude of enhancement in C_N from successive stimuli
Force generation		
F	N	Force generated by muscle
A	N/ms	Scaling factor for force
τ_1	ms	Time constant of force decline in the absence of strongly bound cross-bridges
τ_2	ms	Time constant of force decline due to friction actin and myosin resulting from the presence of strongly bound cross-bridges
K_m	—	Sensitivity of strongly bound cross-bridges to C_N

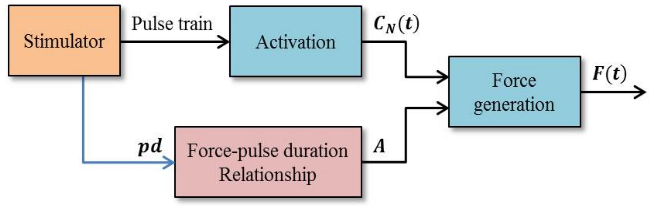


Fig. 1: Mechanism of the force generation by quadriceps muscle from FES signal : the Ding et al force model [16].

In [17], the same research group proved that the force-pulse duration relationship did not change with the frequency and fatigue. These findings shows the ability of the model to include it into a model-based controller design for adjusting the stimulation pulse duration in order to achieve desired force response during FES applications.

In our study, the additional component (Eq. (3)) is considered as the normalized stimulation input and will be associated to the IOFLC to compute the pulse duration in order to control the force output. This normalized stimulation input is given as follows:

$$u(pd) = \frac{A}{a^{'}} = 1 - e^{-\dfrac{pd - pd_0}{pd_t}}. \tag{4}$$

As the muscle cannot generate negative force and the scaling term factor A should be positive, we consider that pd shouldn't be inferior to pd_0 ($pd > (pd_{min} \simeq pd_0)$) and we saturate also the pulse duration to a maximal practical value ($pd < (pd_{max} \simeq 800\mu s)$) usually used in FES (see Fig. 2). Thus, the normalized stimulation input is equal to:

$$u = \begin{cases} 0 & if \quad pd \leqslant pd_0, \\ 1 - e^{-\dfrac{pd - pd_0}{pd_t}} & if \quad pd_0 < pd \leqslant pd_{max}, \\ 1 & if \quad pd > pd_{max}. \end{cases} \tag{5}$$

However, the stimulation pulse duration can be computed in the control session using the following transformation:

$$pd = pd_0 - pd_t ln(1 - u) \quad with \quad u \in [0, u(pd_{max})]. \tag{6}$$

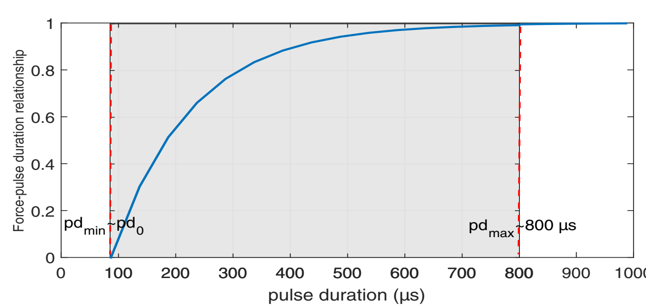


Fig. 2: The normalized force-pulse duration relationship for a typical subject. The grayed area shows the used range of pulse duration in FES applications.

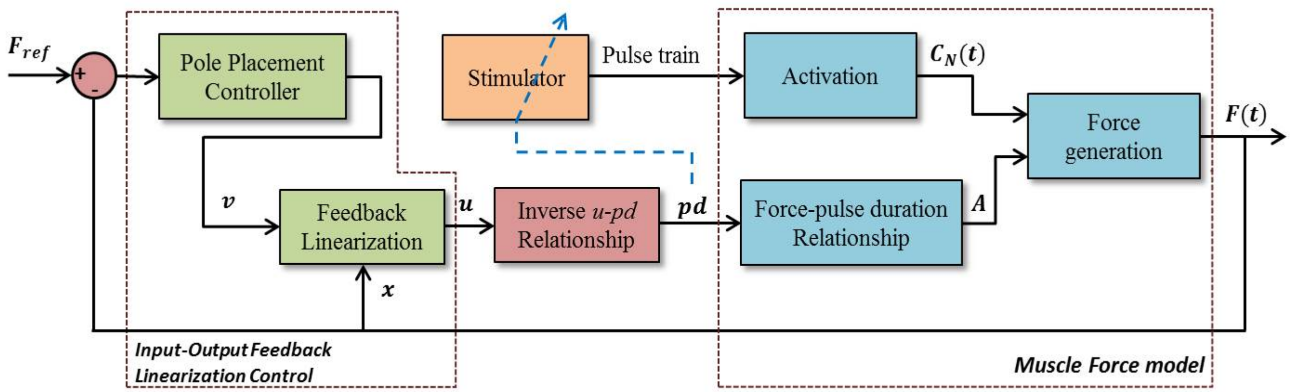


Fig. 3: Block diagram showing the input-output feedback linearization control (IOFLC) applied to the muscle force model where the input control is the pulse duration and the output is the muscle force.

C. Controller Design: IOFLC

1) Feedback linearization: The feedback linearization approach has been introduced by Isidori in 1995 [18]. This approach is essentially based on the Lie derivative and the relative degree of a nonlinear system.

Consider the following state space representation of a SISO nonlinear system:

$$\begin{cases} \dot{x} = f(x) + g(x)u, \\ y = h(x). \end{cases} \tag{7}$$

with $x \in \mathbb{R}^n$ corresponds to the state, $u \in \mathbb{R}$ the control input and $y \in \mathbb{R}$ the output, f and g are smooth vector fields on $\mathbb{R}^n$ and h is a smooth nonlinear function. Differentiating y with respect to time, we obtain:

$$\begin{aligned} \dot{y} &= \frac{\partial h}{\partial x}\left(\frac{\partial x}{\partial t}\right) = \frac{\partial h}{\partial x} f(x) + \frac{\partial h}{\partial x} g(x)u \\ &= L_f h(x) + L_g h(x)u \end{aligned} \tag{8}$$

with $L_f h(x) : \mathbb{R}^n \to \mathbb{R}$ and $L_g h(x) : \mathbb{R}^n \to \mathbb{R}$ are the Lie derivative of h with respect to f and g, respectively. Then, if the $L_g h(x) \neq 0$, the state feedback law:

$$u = \frac{1}{L_g h(x)}(-L_f h(x) + v). \tag{9}$$

If the nonlinear system has a relative degree r, the differentiation of y in Eq. (7) is continued until $L_g L_f^{r-1} h(x) \neq 0$ and is expressed as follows:

$$y^{(r)} = L_f^r h(x) + L_g L_f^{r-1} h(x)u, \tag{10}$$

and the control input is equal to:

$$u = \frac{1}{L_g L_f^{r-1} h(x)}(-L_f^r h(x) + v). \tag{11}$$

The final new input-output relation becomes:

$$y^{(r)} = v \tag{12}$$

where v is called the synthetic control where many linear controller methods can be used to design it [19]. For our study, a pole placement control is applied, this classic feedback controller takes as input the error between the desired output and the system output and as output the control v.

2) Pole placement controller: The pole placement method is a controller design method to stabilize the output at a desired reference value or to track a desired trajectory. In general case when the system has a relative degree r, v is chosen as:

$$v = y_d^{(r)} - c_{r-1}(y^{(r-1)} - y_d^{(r-1)}) - \cdots - c_1(\dot{y} - \dot{y}_d) - c_0(y - y_d). \tag{13}$$

Replacing the expression of v (Eq. (13)) in Eq. (12), yields:

$$e^{(r)} + c_{r-1}e^{(r-1)} + \cdots + c_1\dot{e} + c_0 e = 0, \tag{14}$$

with, $e = y - y_d$. In the frequency domain, the Eq. (14) yields:

$$s^r + c_{r-1}s^{r-1} + \cdots + c_1 s + c_0 = 0. \tag{15}$$

To guarantee the stability of the system, coefficients c_0, c_1, $\cdots$, c_{r-1}, should be selected so that the polynomial (Eq. (15)) is Hurwitz (i.e., the eigenvalues should have negative real part).

3) Muscle force control: In our case, the output of the model to be controlled ($y = x_2 = F$) is linked directly to the control input and its differentiation is equal to the following equation:

$$\dot{y} = \dot{x}_2 = f_2(x) + g_2(x) * u, \tag{16}$$

where $f_2(x) = -\frac{F}{\tau_1 + \tau_2 \frac{C_N}{K_m + C_N}}$ and $g_2(x) = a' \frac{C_N}{K_m + C_N}$ and u is the normalized stimulation input.

Now, according to the IOFLC approach, we obtain:

$$\dot{y} = L_f h(x) + L_g h(x)u, \tag{17}$$

where,

$$\begin{aligned} L_f h(x) &= f_2(x) = -\frac{F}{\tau_1 + \tau_2 \frac{C_N}{K_m + C_N}}, \\ L_g h(x) &= g_2(x) = a' \frac{C_N}{K_m + C_N}. \end{aligned}$$

To track a reference F_{ref}, the output v of the pole placement controller that controls the normalized stimulation input of the force model is: $v = \dot{F}_{ref} - c_0(F - F_{ref})$ as ($r = 1$). Then, the normalized control input u is computed by the Eq. (9) and after that, we use the inverse u-pd relationship (Eq. (6)) to update the pulse duration at each pulse applied to the muscle force model (see Fig. 3).

III. Results and discussion

The simulations results of this study were performed in MATLAB/SIMULINK software environment. The parameters values set in the force model such as presented in Table II are reported from [1] by fitting experimental data from quadriceps muscles of able-bodied subject. The sampling time used in these simulations was equal to 1 ms and the sampling time used for the control was equal to the inter-pulse interval (frequency train). It is assumed that the method does not affect the stability of the system.

A. Prediction ability of the muscle force model

Referred to the experimental results presented by Ding et al in previous studies, this model that included the force pulse duration relationship for human quadriceps muscles has well predicted the force responses over a wide range of stimulation pulse durations (100 μs to 800 μs)and stimulation frequencies (12.5 Hz to 80 Hz). In the figure 4, we show the predicted force responses for different constant frequency trains (12.5 Hz, 20 Hz, 33 Hz, 50 Hz, 80 Hz) when the pulse duration was fixed to 500 μs.

The prediction of the pulse duration effect is shown in Fig. 5. This figure presents the force responses for different stimulation pulse durations (100 μs to 800 μs) when the constant frequency train is with an inter-pulse interval equal to 20 ms (CFT20$\sim$50 Hz).

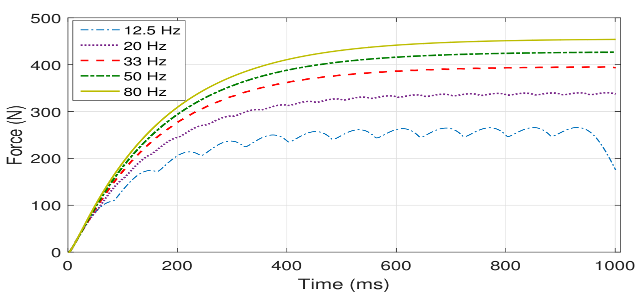


Fig. 4: Example of force responses for a typical subject to different constant frequency trains (12.5 Hz (CFT80), 20 Hz (CFT50), 33 Hz (CFT30) and 50 Hz (CFT20)) with a fixed pulse duration of 500 μs.

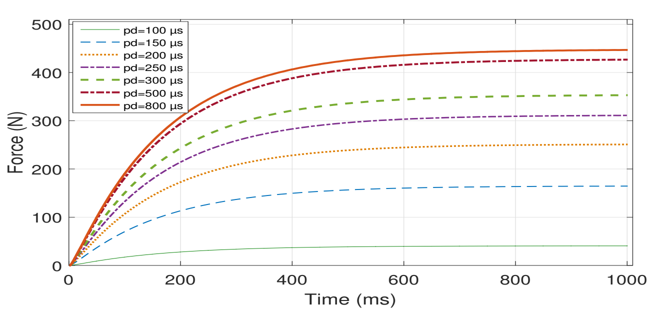


Fig. 5: Example of force responses for a typical subject to different pulse durations which are ranged between 100 μs and 800 μs of a constant frequency train with an inter-pulse interval equal to 20 ms (CFT20).

These results show that increasing stimulation parameters yields a rise of the peak force responses. In addition, the exponential dependence relation between the developed peak force level and the pulse duration justifies the use of a control strategy in order to control the muscle force by modulating the stimulation pulse duration. These simulation results, without control, prove that a large interval of forces could be reached by varying the stimulation pulse duration.

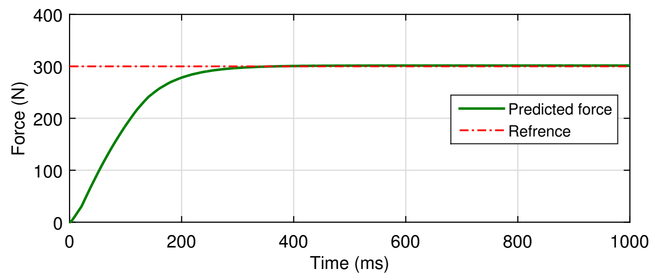
B. Results of the IOFLC

A good tracking using the IOFLC was achieved in all simulations. Firstly, in the case of tracking a fixed reference force F_{ref}=300 N (Fig. 6). The performance of the control, the normalized control input and the controlled pulse duration are respectively shown in Fig. (6a, 6b, 6c). Fig. 6c demonstrates that the controlled pulse duration is saturated by pd_{max} at start time and then it satisfies the control saturation condition ($pd_0 \leqslant pd \leqslant pd_{max}$), this result is acceptable because at start time the system needs to reach the desired reference. The IOFC method is also tested for various reference forces (180 N, 200 N, 250 N and 300 N) with a constant frequency train CFT20 (Fig. 7). The results presented in figure 7 show also that this strategy of control can track many desired forces using the same parameters control and using only one stimulation frequency train (CFT20 in this case). For all the tested force references, we can see that the force is maintained to the reached force reference and instability was never observed. Fig. 8 shows the results when the force reference is varied. In this case, the control is applied to track a force reference of 100 N, followed by a reference of 200 N, then for a reference of 150 N and at last for 230 N as reference, the switching between these references is realized every 1 s.

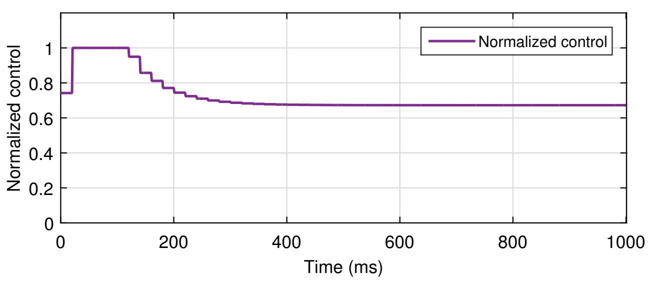
Furthermore, we show in Fig. 9 the simulation results of the IOFLC ability to modulate the pulse duration to track a desired force reference of 300 N when varying the stimulation frequency. These results show that the more the frequency of stimulation decreases (i.e., increasing the sampling time of the control) the more the performances of the control degrade like for a low pulse frequency (20 Hz [T_s=50 ms]). This inconvenient is due to the increasing for the sampling time used to calculate the control.

In order to check the robustness of the control, a disturbing noise was added to the reading of the muscle force output from the nonlinear system between 500 ms and 600 ms (see Fig. 10). The disturbance effect can be observed from the zoom displayed in Fig. 10a and it has a mean magnitude of 10 N. The results, shown in Fig. 10a, 10b and 10c, proved that the control is robust to the external disturbance rejection.

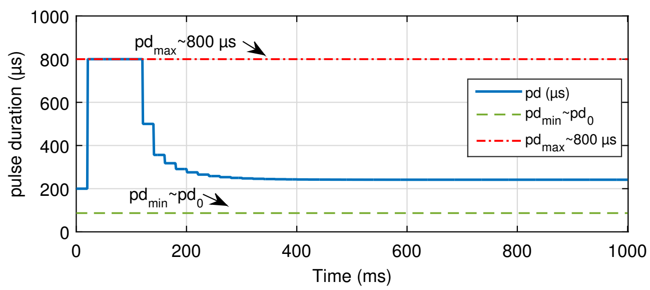
All of the simulation results show that this model can be included into a feedback control FES system to provide a more precise control of the muscle force during FES tasks. The approach that considers the additional component as a normalized input control, proposed in this study to control the muscle force, make the Ding et al model useful to design model-based controllers for controlled FES systems.



(a) Predicted Force (—), reference (-.-)

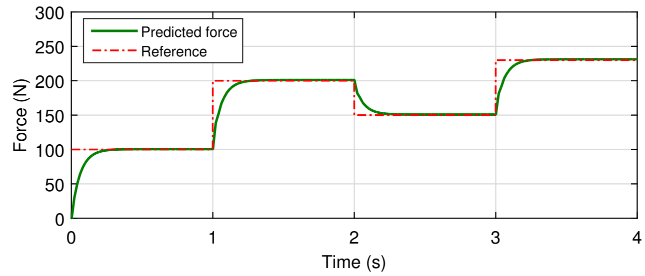


(b) Normalized control

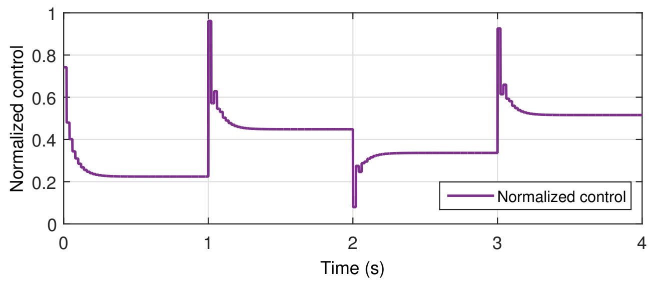


(c) Pulse duration (pd)

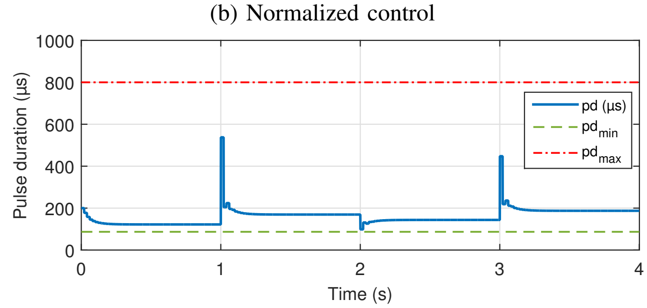
Fig. 6: Computer simulation results when the desired force is a fixed reference F_{ref} =300 N. (a) Top graph shows the predicted force (solid line) and the desired reference (dashed line). (b) The middle graph represents the corresponding normalized control calculated by the IOFLC. (c) The bottom graph represents the pulse duration computed by the normalized input control.



(a) Predicted Force (—), reference (-.-)



(b) Normalized control



(c) Pulse duration (pd)

Fig. 8: Computer simulation results when the desired force is a varying step reference. (a) Top graph shows the predicted force (solid line) and the desired reference (dashed line). (b) The middle graph represents the corresponding normalized control calculated by the IOFLC. (c) The bottom graph represents the pulse duration computed by the normalized input control.

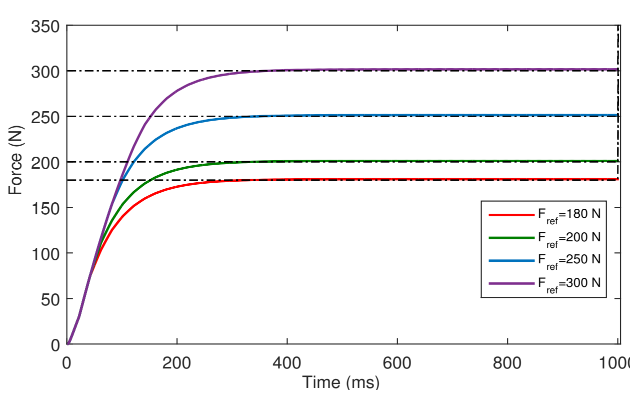


Fig. 7: Control performance of the muscle force for different desired force references. The results of four references are illustrated (180 N, 200 N, 250 N, 300 N).

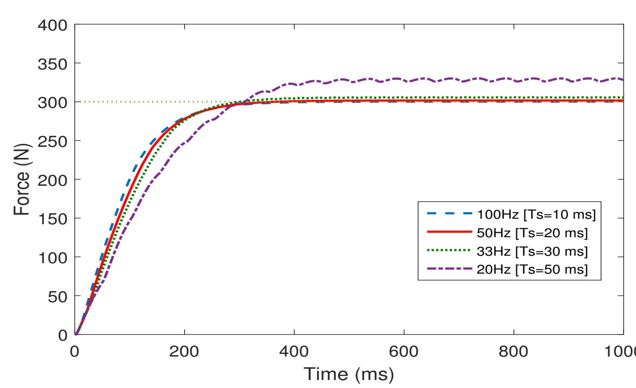
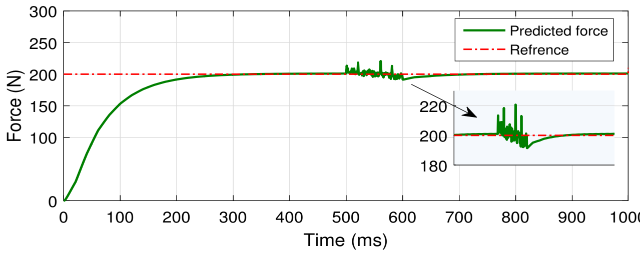
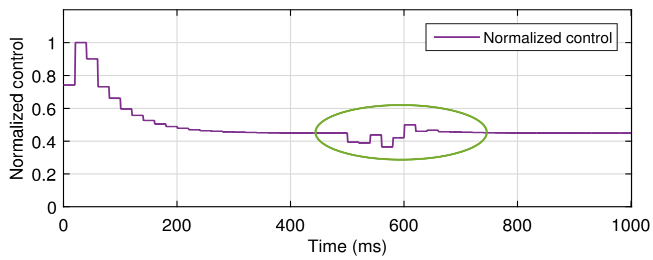


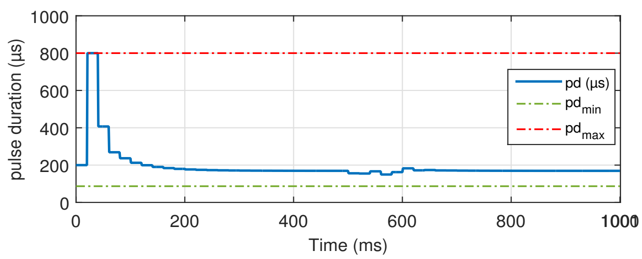
Fig. 9: Control performance of the muscle force with a desired force reference of 300 N when varying the stimulation frequency. The results of four frequencies are illustrated (20 Hz, 33 Hz, 50 Hz, 100 Hz).



(a) Predicted Force (—), reference (-.-)



(b) Normalized control



(c) Pulse duration (pd)

Fig. 10: Results from disturbance rejection when tracking a force reference F_{ref} =200 N. A random noise is applied to the output measurement approximately between 500 and 600 ms during a closed loop control session.

TABLE II: Parameters values for a typical subject used to simulate and to control the muscle force model.

τ_c	R_0	a'	τ_1	τ_2	K_m
20 ms	1.143	3.009 N/ms	50.95 ms	124.4 ms	0.103

IV. Conclusion

A novel closed loop FES control scheme based on the theory of feedback linearization was designed and tested in computer simulations in this paper for the controlling of the muscle force response through FES. The IOFL controller is applied to a physiological nonlinear force model that successfully predicts the force response. A normalized control output is designed to modulate the stimulation pulse duration. The simulation results show that the IOFLC is capable to control a produced muscle force and to achieve a good tracking performance in different cases when tracking a fixed or a varying desired reference, even in the presence of a disturbance. This study is very helpful to examine the performance of a feedback linearization into a closed loop control for FES-systems.

References

[1] J. Ding, A. S. Wexler, and S. A. Binder-Macleod, “Mathematical models for fatigue minimization during functional electrical stimulation,” *Journal of Electromyography and Kinesiology*, vol. 13, no. 6, pp. 575–588, 2003.

[2] J. Ding, A. S. Wexler, and S. A. Binder-Macleod, “A mathematical model that predicts the force–frequency relationship of human skeletal muscle,” *Muscle & nerve*, vol. 26, no. 4, pp. 477–485, 2002.

[3] L. A. F. Law and R. K. Shields, “Predicting human chronically paralyzed muscle force: a comparison of three mathematical models,” *Journal of applied physiology*, vol. 100, no. 3, pp. 1027–1036, 2006.

[4] N. Kirsch, N. Alibeji, and N. Sharma, “Nonlinear model predictive control of functional electrical stimulation,” *Control Engineering Practice*, vol. 58, pp. 319–331, 2017.

[5] S. Jezernik, R. G. Wassink, and T. Keller, “Sliding mode closed-loop control of fes controlling the shank movement,” *IEEE transactions on biomedical engineering*, vol. 51, no. 2, pp. 263–272, 2004.

[6] A. Ben Hmed, T. Bakir, A. Sakly, and S. Binczak, “Controlling muscular force by functional electrical stimulation using intelligent pid,” in *Sciences and Techniques of Automatic Control and Computer Engineering (STA), 2015 16th International Conference on*, pp. 75–79, IEEE, 2015.

[7] A. Ben Hmed, T. Bakir, Y. Garnier, S. Binczak, and A. Sakly, “A novel strategy for adjusting current pulse amplitude of fes-systems with pid based on pso algorithm method to control the muscle force,” in *Proceedings of the 14th International Conference on Informatics in Control, Automation and Robotics - Volume 1: ICINCO*, pp. 664–669, INSTICC, 2017.

[8] T. Schauer, N.-O. Negard, F. Previdi, K. Hunt, M. Fraser, E. Ferchland, and J. Raisch, “Online identification and nonlinear control of the electrically stimulated quadriceps muscle,” *Control Engineering Practice*, vol. 13, no. 9, pp. 1207–1219, 2005. Modelling and Control of Biomedical Systems.

[9] A. Ben Hmed, T. Bakir, S. Binczak, and A. Sakly, “Model free control for muscular force by functional electrical stimulation using pulse width modulation,” in *Control Engineering & Information Technology(CEIT), 2016 4th International Conference on*, IEEE, 2016.

[10] C.-C. K. Lin, W.-C. Liu, C.-C. Chan, and M.-S. Ju, “Fuzzy control with amplitude/pulse-width modulation of nerve electrical stimulation for muscle force control,” *Journal of Neural Engineering*, vol. 9, no. 2, p. 026026, 2012.

[11] N. Kirsch, N. Alibeji, B. E. Dicianno, and N. Sharma, “Switching control of functional electrical stimulation and motor assist for muscle fatigue compensation,” in *2016 American Control Conference (ACC)*, pp. 4865–4870, July 2016.

[12] S. Mohammed, P. Poignet, P. Fraisse, and D. Guiraud, “Toward lower limbs movement restoration with input-output feedback linearization and model predictive control through functional electrical stimulation,” *Control Engineering Practice*, vol. 20, no. 2, pp. 182–195, 2012.

[13] Y. Chen, G. Damm, A. Benchaib, and F. Lamnabhi-Lagarrigue, “Feedback linearization for the dc voltage control of a vsc-hvdc terminal,” in *2014 European Control Conference (ECC)*, pp. 1999–2004, June 2014.

[14] A. Ammar, A. Bourek, and A. Benakcha, “Nonlinear svm-dtc for induction motor drive using input-output feedback linearization and high order sliding mode control,” *ISA Transactions*, vol. 67, no. Supplement C, pp. 428–442, 2017.

[15] V. Alfieri, G. Conte, and C. Pedicini, “Feedback linearization control for the air amp; charging system in a diesel engine,” in *2015 European Control Conference (ECC)*, pp. 2565–2570, July 2015.

[16] J. Ding, L.-W. Chou, T. M. Kesar, S. C. Lee, T. E. Johnston, A. S. Wexler, and S. A. Binder-Macleod, “Mathematical model that predicts the force–intensity and force–frequency relationships after spinal cord injuries,” *Muscle & nerve*, vol. 36, no. 2, pp. 214–222, 2007.

[17] L.-W. Chou and S. A. Binder-Macleod, “The effects of stimulation frequency and fatigue on the force-intensity relationship for human skeletal muscle,” *Clinical Neurophysiology*, vol. 118, no. 6, pp. 1387–1396, 2007.

[18] A. Isidori, “Nonlinear control systems,” 1995.

[19] C. I. Pop and E. H. Dulf, *Robust Feedback Linearization Control for Reference Tracking and Disturbance Rejection in Nonlinear Systems*, ch. 12 Recent Advances in Robust Control - Novel Approaches and Design Methods, pp. 273–290. InTech, November 2011.