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Upper Limb Musculoskeletal Modeling for Human-Exoskeleton Interaction

Arturo González-Mendoza
UMI-LAFMIA
CINVESTAV
Mexico City, Mexico
arturo.gonzalez@cinvestav.mx

Ricardo Lopéz-Gutierrez
UMI-LAFMIA
CINVESTAV
Mexico City, Mexico

Alberto Isaac Pérez-SanPablo
Analysis movement laboratory
INR
Mexico City, Mexico

Sergio Salazar-Cruz
UMI-LAFMIA
CINVESTAV
Mexico City, Mexico

Ivette Quiñones-Uriostegui
Analysis movement laboratory
INRMexico City, Mexico

Marie-Christine Ho Ba Tho
Sorbonne University, Université de technologie de Compiègne, CNRS, UMR 7338 Biomechanics and Bioengineering, Compiègne, France

Tien-Tuan Dao
Sorbonne University, Université de technologie de Compiègne, CNRS, UMR 7338 Biomechanics and Bioengineering, Compiègne, France

***Abstract*— The integration of the human body structure and function into the design of human-exoskeleton interface is an engineering challenge. A solution to overcome this problem is the use of musculoskeletal modeling tools such as OpenSIM. The objective of this paper is to develop and use an upper limb musculoskeletal model using OpenSIM to evaluate a Proportional Derivative controller (PD) of an exoskeleton. Obtained results show that the proposed musculoskeletal model could be a reliable tool for exoskeleton design study and controller testing when joint elbow speed is not superior to 300 °/sec.**

Keywords—Human Machine Interface (HMI), musculoskeletal model, OpenSIM, surface Electromyography (sEMG), exoskeleton, controllers.

I. INTRODUCTION

One of the most common practices when designing a exoskeleton is to ignore or reduce the complexity of the dynamics, the biomechanics, and the neuromuscular controls of the human body [1]. These practices lead to poor exoskeleton designs involving technical problems like actuators and sensors [2, 3], making difficult to understand and evaluate the positive or negative effects of an exoskeleton over the musculoskeletal system [1]. The exoskeletons are emerging as a rehabilitation tool [4], whose main purpose is to help a patient to improve the health-related consequences of a pathology with abnormal movement patterns. For this reason, there arises the need of a deeper understanding of the human-exoskeleton interaction.

A potential approach that addresses the above-mentioned problems is the use of computerized musculoskeletal dynamic models. This type of models are powerful tools for studying the biomechanics of human movement. Musculoskeletal models include two main analyses: 1) inverse kinematics/dynamics [4] and 2) forward dynamics and computerized muscle control [5]. The first analysis is a descriptive approach and the second one is a predictive approach. Note that sEMG signals are commonly used to drive forward dynamic's simulation. Moreover, in [1], the authors showed that computer-based functional physical simulations are used for iterative refinement of products early in the design stage, which reduces costs and improves performance.

In addition, sEMG signals have been widely deployed for advanced HMIs. According to [6] there are three methods to define the relation between the sEMG signal and the respective joint movement. The first method consists on musculoskeletal modelling, the second deals with auto-regressive models and the last one uses artificial neural networks. OpenSIM has been successfully used to integrate sEMG signal into musculoskeletal model to simulate the respective joint movement [8]. However it suffers from poor reliability or robustness due to different reasons, such as: the type of mathematical muscle model used, the speed performance of solving dynamical equations, model parameters, or necessary software tools. In fact, the objective of this work was to develop and use an upper limb musculoskeletal model to perform dynamic movements from sEMG signals. Moreover, a PD controller was integrated into the model to evaluate its effect. The paper is organized as follows: Section II presents the methods related to the experiment protocol, the data processing, the musculoskeletal modeling and the PD controller design and implementation.

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Section III shows the results obtained from forward dynamics simulations. Section IV summarizes the work and addresses future perspectives.

II. Methods

A. Experimental setup and protocol

To carry out the data acquisition, a protocol similar to the one established in [7] was designed. The spatial configuration consists of a deflection pulley (this pulley is aligned and attached to the specific joint to be measured), which itself is connected to the pulley machine. In this way, the forearm of the subject is connected with a lever to the deflection pulley: the midpoint of the deflection pulley must coincide with the position of the joint elbow axis to be measured (see Fig. 1). Thus, the torque generated by the different external loads of the pulley machine keeps constant along the whole range of performed flexion and extension movements. In addition, the rope of the pulley machine must be connected to the deflection pulley from the bottom or from the top, allowing to invert the torque generated by the external loads. Therefore, depending on the configuration of the deflection pulley, different contributions of the activation of the arm muscles can be achieved in different movements of the joint. Subject kinematics were recorded with a 10 camera OptiTrack (NaturalPoint, Corvallis, OR, USA) system using the 27-marker upper limb base model. The sEMG of the Biceps Brachii (Long head), the long head of the triceps, the brachioradial, and the lateral head of the triceps were recorded using an 8-channel wireless Datlaog MWX8 sEMG system (Biometrics Ltd, Newport, UK). Electrodes with an electrode separation of 20mm with the electrode diameter being 4mm were placed following the SENIAM [8] and the manual of electromyography [9] (see Fig. 2.).

One healthy subject participated in the data acquisition process. The subject sat on a chair and its elbow joint was attached to the deflection pulley. The load of the joint attached to the pulley was of 2.5 kg (considered to generate high amplitude sEMG signals when making experimental testing). To stimulate different angular velocities, a metronome sound was used to give the rhythm of the arm movement (the measuring intervals depend on the velocity of the articulation movement). The TABLE I. displays the number of flexion-extension repetitions and the time of the movement.

TABLE I. FLEXO – EXTENSION TIME DURATION AND REPETITIONS. EXPERIMENT PROTOCOL.

Flexion-extension duration time and number of repetitions.	
Time [s]	***Repetitions***
6	3
3	3
1.5	5
0.9	6

B. Data Processing

To obtain the joint angle, the marker trajectory data obtained from the OptiTrack system was processed using an inverse kinematics analysis using the model described in

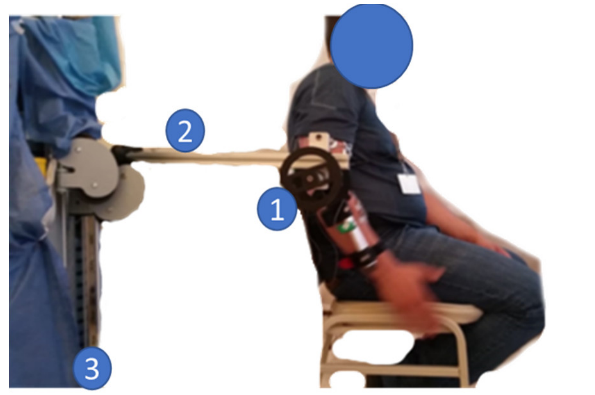

Fig. 1. Experimental Setup. 1) Deflection pulley. 2) Deflection pulley, pulley machine connection bar. 3) Pulley machine.

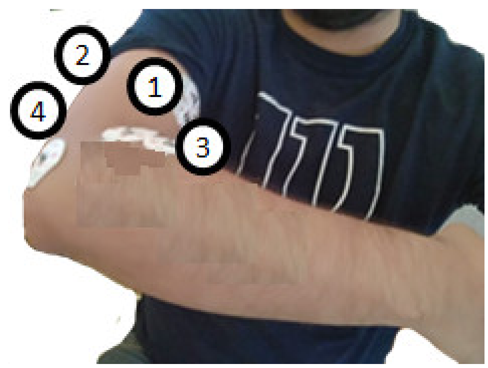

Fig. 2. Electrode Positioning. 1) Long head biceps. 2) Long head of triceps. 3) Brachioradial. 4) Lateral head of the triceps. Pulley machine.

subsection C. The sEMG data were pre-processed following the procedure stated on [10] as follows: 1) Band - pass filtering of the sEMG data, using a 4th order Butterworth filter, with the cutoff frequency of 30 and 400 Hz for removing noises; 2) full-wave rectification ; 3) 4th Order Butterworth Filter, with 10 Hz cutoff frequency to obtain the envelope of the signal; and 4) Normalization with the highest peak of the sEMG signal.

The tracking data was sampled at 100 Hz while the EMG data was sampled at 1 KHz. Therefore, the tracking data was resampled to 1 KHz. The processed sEMG data and the resampled tracking data were aligned using a cross correlation process.

C. Musculoskeletal Modeling

Three musculoskeletal models with different configuration and parameters were created as follows:

1) The first model has 25 [11] markers as the baseline upper limb model from Optitrack and has 6 degrees of freedom. This model is utilized for converse kinematics since this model has a relative position reference at the base of the column segment that allows this model to be situated in the

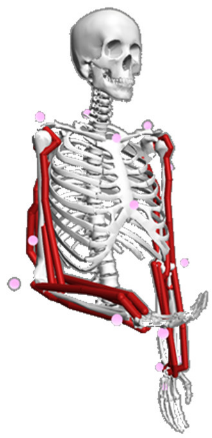

Fig. 3. Upper limb musculoskeletal model: 20 bony bodies, 10 DoF, 16 muscles, and 25 markers.

plane dependent on the standard of arrangement of markers, and furthermore doesn't have muscle elements.

2) The second model has 26 muscles and doesn't have skin-mounted markers. The reference of this model is not relative, and is referenced to a ground that simulates gravity. This model was used to run forward simulations.

3) The third model is the same as the second model, but it has a virtual actuator in the elbow (simulates elbow exoeskeleton actuator) to execute the laws of control over it.

The second and the third models are built with 20 bony segments and 16 muscles (8 muscles for left and right arm) (see Fig. 3.). The implemented muscles are: Triceps Long, Triceps Lateral, Biceps Long, Brachioradialis (BRA), Extensor Carpi Radialis Brevis (ECRB), Flexor Carpi Radialis (FCR), Flexor Carpi Ulnaris (FCU), Extensor Carpi Radialis (ECR). These muscles are involved in the flexion–extension of the elbow and wrist, pronation–supination of the elbow and, medial–lateral deviation of the wrist.

For the proposed models, according to [12] and age of the patient, parameters such as: max contraction velocity, deactivation time constant, muscle strain and ratio of maximum lengthening muscle force to isometric force, were set to 10 [m/s], 50 [ms], 0.5, 1.8 respectively. To define the inertia moments of the bones, the analytical calculations defined in [13] were established.

D. *Actuator controller testing*

To implement the controller, a program using OpenSIM – MATLAB software development kit was implemented. The flow diagram of the program is shown in Fig. 4. The program loads a file with the processed sEMG signal then, a forward dynamic process is run, for this the MATLAB function "ode23s" is run with an absolute tolerance error of 1e-3. This function envelopes a function in which the dynamics of the musculoskeletal model are integrated. Then this function gives the position of the musculoskeletal system, that after is feed into a PD controller.

In the controller, the variable, τ was the joint actuator torque, q and $\dot{q}$ were the position and velocity obtained by integration of the dynamics of the musculoskeletal model, qd, $\dot{q}d$ were desired joint position, and desired velocity respectively. PD controller output signal was:

$$\tau = k_p(qd - q) - k_d(\dot{q}d - \dot{q}) \quad (1)$$

To run the simulation, the gains k_p and k_d were set to 10. These gains were set manually considering that the time of one second to reach qd is between the range of movement of healthy subject. The simulations were run with an Intel core i7 2.8GHz clock, 16 Gb of Ram memory computer.

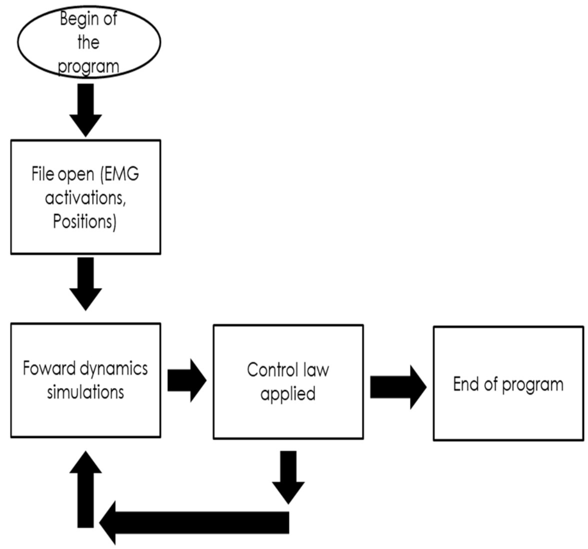

Fig. 4. Flow Diagram of the program.

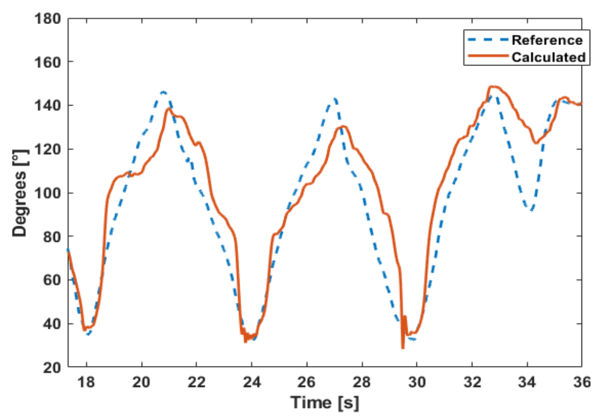

Fig. 5. Six-Seconds Flexion – Extension of elbow joint. Inverse Kinematics Vs Forward simulation outputs.

III. COMPUTATIONAL RESULTS

To evaluate the developed musculoskeletal model, four forward dynamics simulations were performed. These simulations correspond to the flexion-extension time duration and repetitions described in TABLE I. The comparisons between the outputs of the forward dynamics simulations and the experimental data are shown in Fig. 5. Fig.6. Fig.7. Fig.8. The Root Mean Square Error (RMSE) is shown in TABLE II.

TABLE II. FLEXION–EXTENSION DURATION TIME AND RMSE ERROR BETWEEN REFERENCE SIGNAL VS. CALCULATED SIGNAL.

Flexion-extension duration time and RMSE error	
Time [s]	***RMSE***
6	20.94
3	35.23
1.5	37.95
0.9	-

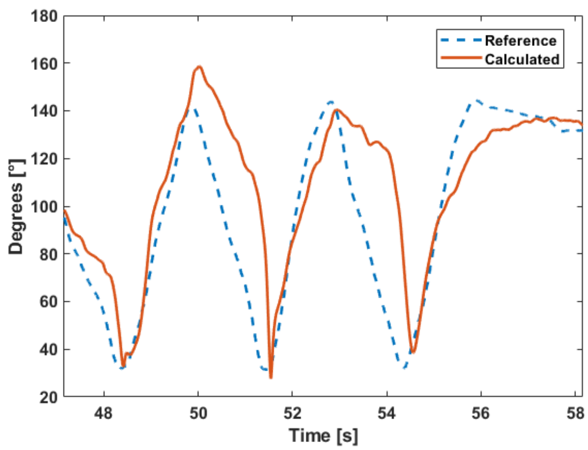

Fig. 6. 3 Seconds Flexion – Extension of elbow joint. Inverse Kinematics Vs Forward simulation outputs.

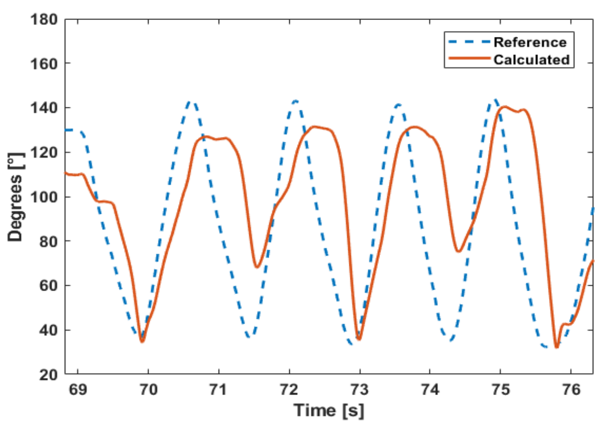

Fig. 7. 1.5 Seconds Flexion – Extension of elbow joint. Inverse Kinematics Vs Forward simulation outputs.

The Fig. 9. shows how the system drives to the specified joint elbow position by using the implemented controller integrated into the musculoskeletal model.

IV. DISCUSSION AND CONCLUSIONS

The simulations cost computational resources due to complexity of the model and data precision. The proposed model has 102 states (including joints, muscles, restrictions, and actuator) for running 100 millisecond forward simulation in 5 seconds. If a subject moves its arm with a cycle of 3 seconds, a 2-minute simulation will be necessary. As it´s shown in Fig.7, Fig.8., and Fig. 9. the shape of the simulated flexion-extension angle is similar to that given by the inverse

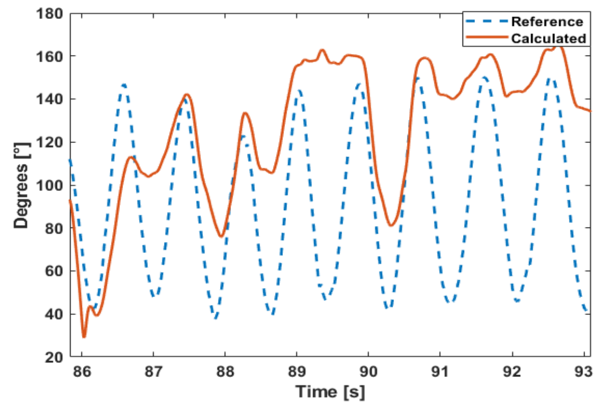

Fig. 8. 0.9 Seconds Flexion – Extension of elbow joint. Inverse Kinematics Vs Forward simulation outputs.

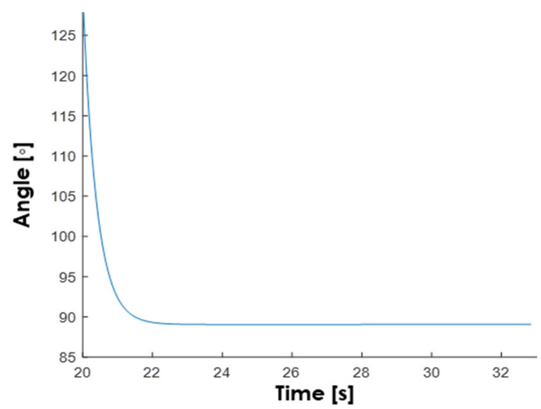

Fig. 9. Arm elbow amplitude reaching to a desired 90° position.

kinematics and considered as reference. This can be considered as an acceptable value for design applications since maximum range of movement can be expected, however these types of models might not be suitable for real time controller because of the high error in the prediction of position. Also it can be said from Fig.7, Fig. 8, Fig. 9 that the shape of the signal is lost as the speed of movement increases.

The developed model showed that its reliability for testing arm movement up to a velocity of 300 °/second (0.9 second cycle). The PD controller takes up to 1 second to get to a desired position as shown in Fig. 9 and no need of better tuning is necessary.

The developed model can be used to test controller of new hardware development of exoskeleton without physical testing on the patients. Moreover, the evaluation of muscle performance can be used for a better exoskeletons design. For future works, Simulink connection with OpenSIM and more simplified models will be investigated leading to a powerful numerical tool for real-time exoskeleton control laws testing and design.

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