Identification of human musculo-tendon subject specific dynamics using musculo-skeletal computations and non linear least square

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Abstract— **The human body dynamics is very complex because of the number of degrees of freedom and of the number of muscles. Moreover the behavior of muscles is non-linear and subject specific. This paper presents an original method to estimate the subject specific dynamic parameters of the human muscles. A dynamic model of muscle, commonly used by the biomechanics community, is first presented. Then the force-activity-length-velocity relation is given. An application with four muscles, to the flexion/extension of the elbow joint is proposed. The human arm dynamics is analyzed in a motion capture studio for specific movements that excite the muscle dynamics. The acquisition of movements allows to compute the inverse kinematics and the inverse dynamics using a musculo-skeletal model of the human body, and finally the muscle force is estimated (input of the dynamic model of the muscle). The activity (also input of the dynamic model of the muscle) is measured using electromyography (EMGs) for the superficial muscles. The subject specific parameters of those muscles are then estimated by the non-linear least square method with Newton-Gauss algorithm. Experimental results obtain for valid subject are given.**

Index Terms— **musculo-tendon dynamics, inverse dynamics, muscculo-skeletal human model, electromyogram, motion capture**

I. Introduction

Robots get closer to human due to the developments of artificial organs such as active polymer gels that are used for artificial muscles [1], the miniaturization of components and the enhancements of computation power. Although the human body is very complex and the ability of movement is very wide which make its comprehension definitely difficult. Nevertheless, for a clear understanding of the human dynamics and to generate smooth movements it is very important to have a good knowledge of the human anatomy as well of its dynamics. Though actuators of the human body are the muscles, their dynamics is very important in the study of the human body movements and thus requires the best understanding. However the human muscular system is complex. This complexity remains in the biologic and chemical nature of muscle fiber, the high number of muscles and degrees of freedom and also because of the differences between each of the human beings: size, mass, strength, capacity... which are closely related to their history: doing sports, having had injuries, metabolism...

In dynamics modelling it is important that the model used gives a realistic description of the behavior and that this model is consistent with the subject. For this reason the subject specific musculo-tendon's dynamics must be estimated. Unlike usual works on the human body where the model is only primarily scaled to the subject (size and mass) [2], the proposed method focuses in identifying in-vivo the subject specific muscle dynamics parameters. The model used to describe the muscle dynamics is based on the empirical Hill-Stroeve model [3], known to be the most accurate macro-model that describe the complex musculo-tendon dynamics in relation to the micro-structure of the muscle. A motion capture studio and a musculo-skeletal model of the human body [4] are used to compute the inverse dynamics and inverse kinematics of the human body from optical markers position and give the joint angle and the joint torque, as well as an optimize solution of the muscle force for each muscle. They are used as input of the identification model. The paper is organized as follow: in the second section the musculo-tendon dynamics and the Hill-Stroeve model are presented, while the third section gives the dynamic modelling of the flexion/extension of the elbow joint. Then the experimental set up is described in the fourth section. Finally, in the fifth section the identification method using non-linear least square method based on the Newton-Gauss algorithm is presented and experimental results of the identification of the muscle dynamics are shown.

II. Musculo-tendon complex dynamics

Each muskulo-skeletal muscle of the human body can be described by a musculo-tendon complex of length l_{mt} using the Hill-Stoeve model [3]. This musculo-tendon complex is composed of a tendon and a muscle (Fig.1). The tendon is passive and does not generate movement. The muscle is an active contractile element that generates contractions force controlled by the neural excitation $u(t)$. According to the excitation, the desired movement and the external forces and interactions the muscle lengthens ($\dot{l}_m > 0$) or shortens ($\dot{l}_m < 0$). Both cases are assumed to be iso-volumic [5].
The musculo-tendon dynamics depends on muscle activity $a(t)$, and length and velocity of the muscle respectively l_m, $\dot{l}_m$, and the length and velocity of the tendon l_t and $\dot{l}_t$.

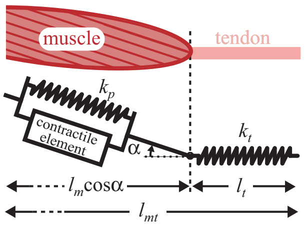

Fig. 1. Modelling of the musculo-tendon complex with pennation angle α, passive stiffness k_p and tendon stiffness k_t

In the literature several models give the muscle activity $a(t)$ with respect to the muscle activation $u(t)$ using first order differential equation such as [6] and [7] or second order differential equation such as [2] and [3]. The latest, given by (1) is the easiest to use and gives good, smooth results with only three time constants.

$$\dot{e} = (u - e)/\tau_{ne}$$
$$\dot{a} = (e - a)/\tau \text{ where } \tau = \begin{cases} \tau_{ac} & e \geq a \\ \tau_{deac} & e < a \end{cases} \tag{1}$$

where e is an intermediate variable, τ_{ne} is the excitation time constant, τ_{ac} and τ_{deac} the activation and deactivation time constants that are quite all the same for every one since it is based on the kinetics of the chemical reaction of calcium in the muscle. Usually $\tau_{act} = 15\,ms$ and $\tau_{deact} = 50\,ms$, however for old people the deactivation time increases to $60ms$ according to [8].
The model given hereafter is a simplification of the one described in [6]: the pennation angle α is neglected for the considered muscles are fusiform or pennation angle is very small and the passive stiffness is not considered. Parameters for the tendon are given in [9]. The force $F_m(t)$ developed by the muscle is function of muscle activity $a(t)$, the muscle length $l_m(t)$, the contraction velocity of the muscle $\dot{l_m}(t)$ and the maximal isometric force F_{max} at full activation $(a(t) = 1)$ which is the maximal force that can be developed by the muscle for an isometric contraction.

$$F_m(t, l_m, \dot{l}_m, x_l, x_v) = a(t) f_l(l_m, x_l) f_v(\dot{l}_m, x_v) F_{max} \tag{2}$$

where f_l is the force-length relation and f_v is the force-velocity relation shown Fig.2 and modelled by:

$$f_l(l_m, x_l) = exp\left(-\left(\frac{l_m - l_m^0}{l_m^{sh}}\right)^2\right) \tag{3}$$

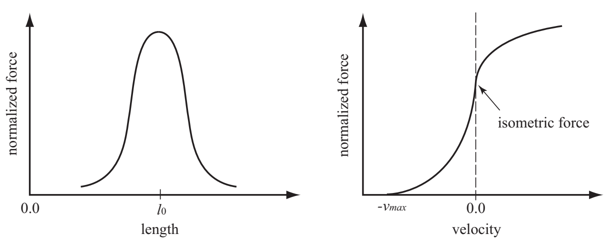

Fig. 2. Force-length relation f_l (right), force-velocity relation f_v (left)

$$f_v(\dot{l_m}, x_v) = \begin{cases} 0 & \text{if } \dot{l_m} \leq -v_{max} \\ \frac{V_{sh}\left(v_{max} + \dot{l_m}\right)}{V_{sh} v_{max} - \dot{l_m}} & \text{if } -v_{max} \leq \dot{l_m} \leq 0 \\ \frac{V_{sh} V_{shl} v_{max} + V_{ml} \dot{l_m}}{V_{sh} V_{shl} v_{max} + \dot{l_m}} & \text{if } \dot{l_m} \geq 0 \end{cases} \tag{4}$$

where v_{max} is the maximum contraction velocity, l_m^0 the optimal length of the muscle to be estimated, V_{sh}, and l_m^{sh} are the other subject specific dynamic parameters to be estimated. x_l is 1×2 vector of force-length subject specific parameters such as $x_l = [l_m^0 \; l_m^{sh}]$ and x_v is 1×3 vector of force-velocity subject specific parameters such as $x_v = [V_{sh} \; V_{shl} \; V_{ml}]$.

Finally, considering the mass of the muscle M_m and its viscosity B_m, and applying the fundamental dynamic equation to the muscle the following relation giving the muscle length is obtained:

$$M_m \ddot{l_m} = F_t - F_m - B_m \dot{l_m} \tag{5}$$

where F_t is the tendon force modelled by an elastic force of stiffness k_t (Fig.1):

$$F_t = k_t l_t \tag{6}$$

with l_t the length of the tendon computed by:

$$l_t = l_{mt} - l_m \tag{7}$$

III. Dynamic modelling of the extension/flexion of the elbow joint

The elbow joint is seldom considered in biomechanics studies but it presents the great advantage over the knee joint to have a very small number of muscles involved and a rather simple anatomy.
The human elbow joint has two main rotational degrees of freedom (Fig.3) that allow the hand to move in a wide operational space: pronation/supination movements and flexion/extension movements. The varus/valgus degree of freedom can be considered as fixed since it is mainly shoulder rotation. Flexion and extension (F/E) of the elbow joint are the only movements considered.
F/E of the elbow joint only involves 4 muscles (Fig.3):

- the Biceps, the Brachialis and the Brachioradialis for flexion,
- the Triceps for extension.

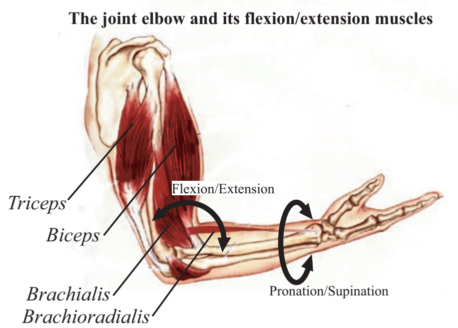

Fig. 3. The elbow joint with its four contraction and extension muscles

A first approach was to consider only the two main antagonist muscles: the Biceps and the Triceps, but the first results have shown that it was impossible to restrict the arm to such a model as it is impossible to avoid the contribution of a muscle involved in a specific movement. The four muscles must be considered. However Biceps and Brachialis are assumed to work together and to have the same activation pattern since it is not possible to measure the activity of the Brachialis with surface electrodes. Similar approximations to estimate the activity of deep muscles are commonly used [2].

The complete joint dynamics of F/E of the elbow joint is then described by (8).

$$J\ddot{q} = T = T_{Br} + T_{Tr} + T_{BB} + T_{ext} - B_l\dot{q} \qquad (8)$$

where J is the inertia of the forearm and the hand, q is the elbow joint angle, $\dot{q}$ and $\ddot{q}$ its first and second derivatives, T is the joint torque, $T_i = F_{ti}r_i$ is the torque due to the musculo-tendon complex $i = Br$ for Brachioradialis, $i = Tr$ for Triceps, $i = BB$ for Brachialis and Biceps and $T_{BB} = T_{Biceps} + T_{Brachialis}$, F_{ti} is the force applied by tendon i, r_i is the moment arm of musculo-tendon i on the moving part, T_{ext} the external torque due to external forces and gravity, and B_l the viscosity of the joint. Moment arm r_i and musculo-tendon complex length l_{mti} are computed using geometric considerations (9).

$$l_{mti} = \sqrt{L_{1i}^2 + L_{2i}^2 - 2L_{1i}L_{2i}\cos q} \qquad (9)$$

This model has been used to build a Matlab-Simulink simulator to study the musculo-tendon behavior and the identifiability of the parameters. Some results are given in Fig.4 for a flexion-extension movement of the elbow joint with step neural input.

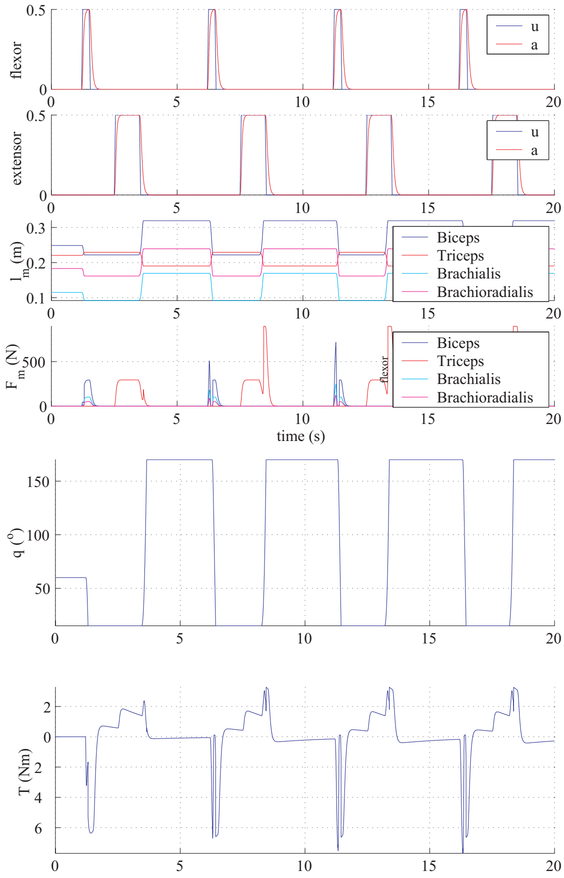

Fig. 4. Simulation of the flexion extension of the elbow joint

IV. Experimental Setup

The experimental set up has been design in order to limit stress on subjects, to have an easy to use system and easily repeatable movements. It consist in:

- a set of EMGs (ElectroMyoGraphy) electrodes to record the muscle activation and
- a set of optical markers to record the joint position with a motion capture system.

A. EMGs recording - Muscle activity

The muscle activation $u(t)$ is recorded using an EMGs system. This system uses surface electrodes that are pasted on the skin above the considered muscles (Fig.5). It can only give the neural input of the superficial muscles: Triceps, Biceps and Brachioradialis. As mentioned above the neural input of the Brachialis is supposed to be the same as the neural input of the Biceps. To obtain good measurements the skin must be prepared: washed with alcohol and scrubbed with sand paper. The temperature and humidity of the room are better to be controlled too. Despite those precautions there is some noise in the EMGs measurements due ti the measurement process itself. But it can be efficiently removed using a low pass forward and reverse second order Butterworth filter with 6 Hz cut off frequency, applied after the full wave rectification of the EMGs such as described in [10].

B. Motion capture system - Joint angle, joint torque and tendon force

The in-house motion capture system is used [11]. The whole system is capable of capturing the marker's position at 30 fps along with the data EMGs at 1 KHz.

Five markers are necessary to record the elbow joint movements accurately therefore three markers are added on the trunk (both hip and opposite shoulder) to insure the global posture as shown Fig.5 and Fig.7. The inverse kinematics and the inverse dynamics models are computed by a musculo-skeletal model of the human body shown, using those measurements [12]. From the markers position input it gives the joint position q with the inverse kinematics and the joint torque T using the inverse dynamics. Finally, $\hat{F_{ti}}$ is computed as the optimized solution for the tendon forces computed from inverse dynamics data [4].

C. Excitation of the muscle dynamics for the estimation

The movements for the estimation are chosen in order to excite the dynamic parameters that have to be estimated. They also must verify the conditions that ensure the good estimation of the muscle forces $\hat{F}_m$ from the estimated tendon force $\hat{F_{ti}}$ such as:

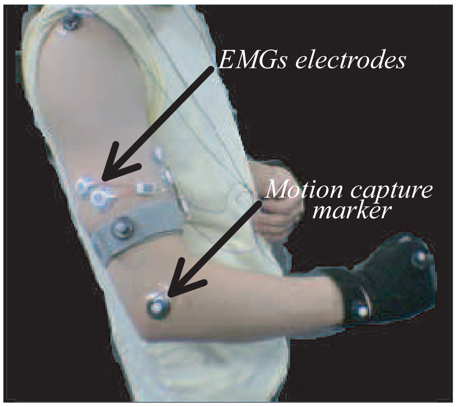

Fig. 5. The experimental equipment: optical markers and EMGs surface electrodes

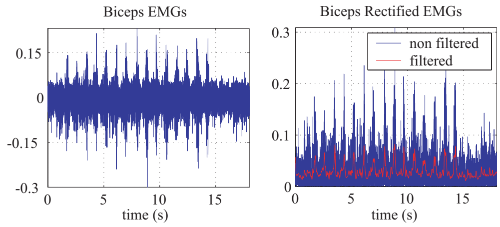

Fig. 6. EMGs, rectified EMGs and filtered rectified EMGs of the Biceps and the Triceps for a flexion/extension of the elbow

- the tendon must have the smallest solicitation so that $\hat{F}_{ti} \approx \hat{F}_{mi}$,
- co-contraction of antagonist muscles must be limited to ensure that the optimal solution found by the optimization procedure is realistic (if co-contraction occurs infinite number of solutions is found)

To ensure this latest condition one solution is to make experiments with movements that have been previously learned since it has been shown in [13] that execution of learned movements limit the co-contraction of antagonist muscles. Precise movements (grasping and posing tasks for example) also require a control pattern with co-contraction of the muscles, consequently those movements are avoided, as well as isometric contractions since they correspond to a contraction of the muscle with no movement and no relevant dynamics for the motion capture.
Finally the movements chosen for the identification are simple flexion-extension of the elbow on the horizontal plane to limit the effect of gravity.

V. Identification of Musculo-tendon dynamics

The parameters to be estimated are the subject specific constant parameters given in (3) and (4), that define the force-length relation and the force-velocity relation. Tough, there are 5 parameters for each muscle to be estimated: $\hat{l_m^0}$, $\hat{l_m^{sh}}$, $\hat{V_{sh}}$, $\hat{V_{shl}}$, $\hat{V_{ml}}$ as F_{max} and v_{max} can be found in the literature.

A. Identification method

Each muscle is considered separately (Brachioradialis, Triceps, Biceps grouped with Brachialis). According to the expression of the muscle force given above in (4) the criteria must be separated in three cases. Cases when the muscle is not activated ($a_i = 0$) are excluded as they mean that the muscle is therefore not lengthening nor shortening and the muscle force is 0. Cases when $\dot{l_m} \leq -v_{max}$ are removed too, as they also mean that muscle force is 0. Thus, the two following cases remain:

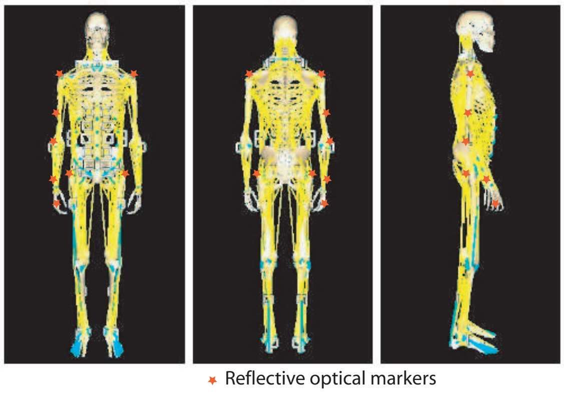

Fig. 7. Musculo-skeletal human model

$$f_v(\dot{l_m}) = \begin{cases} \text{case 1:} \ -v_{max} \leq \dot{l_m} \leq 0: & \frac{\hat{V_{sh}}(v_{max}+\dot{l_m})}{\hat{V_{sh}}v_{max}-\dot{l_m}} \\ \text{case 2:} \ \dot{l_m} \geq 0: & \frac{\hat{V_{sh}}\hat{V_{shl}}v_{max}+\hat{V_{ml}}\dot{l_m}}{\hat{V_{sh}}\hat{V_{shl}}v_{max}+\dot{l_m}} \end{cases} \tag{10}$$

The system is sampled along a movement and leads to an over determinate system for each muscle. Several methods to solve this problem have been tested: simplex method [14], simulated annealing and non-linear least square. For similar results the most efficient method is chosen: non-linear least square method implemented with a Newton-Gauss algorithm [15] with the following criteria:

$$\min_i C(x) = \min_i \frac{1}{2}\sum_i(\hat{F}_t(t_{ji}) - F_m(t_{ji}, x))^2 \tag{11}$$

where j is the case $j = 1$ or $j = 2$, t_j is the sampled time for case j, $\hat{F}_t$ is the tendon force estimated by the optimization routine of the inverse dynamic model of the human body that is assumed to be the muscle force (see section IV-C), F_m the muscle force given by (2) and x the vector of n_p parameters to be estimated. In case 1: $x = [\hat{l_m^0}\hat{l_m^{sh}}\hat{V_{sh}}]$, in case 2: $x = [\hat{l_m^0}\hat{l_m^{sh}}\hat{V_{sh}}\hat{V_{shl}}\hat{V_{ml}}]$.

The Newton-Gauss algorithm consists in minimizing the criteria C given by equation (12) by:

$$\min_{x \in \mathcal{R}^{n_p}} \parallel J(x)d - C(x) \parallel^2 \tag{12}$$

with J the Jacobian, d the direction of the search such as, at computation step k:
$J(x_k)d_k = -C(x_k)$ and $x_{k+1} = x_k + d_k$.

Preliminary works have consisted in studying the identifiability of the parameters and the validity of the method with simulated data set generated by the Matlab-Simulink model (Section III).

B. Experimental results

The muscle activation $u(t)$ for the Triceps, the Biceps (assumed to be the same for the Brachialis) and the Brachio-radialis is measured, rectified and filtered to give the muscle activity $a(t)$. The joint angle q is computed from the motion capture data using the inverse kinematics, the joint torque T and the musculo-tendon $\hat{F}_t$ and F_m forces for each of the muscles are also computed with the inverse dynamics as described above. The movements for the estimation are the one described in section IV.C. Only one record of movement has been available and consequently used for the estimation and validation. Thus cross validation is not yet possible. Results are given for the Triceps, the Brachioradialis, the Brachialis. The latest is more powerful and less subject to fatigue. The maximal isometric force F_{max} is given in the literature for the upper limb for a maximal activation [3]. The initial values x_{init} of vector x for each muscle are the values that can be found in [3]. Results are given in tables I and II according to the cases, for the Triceps, the Brachioradialis and the Brachialis (grouped with the Biceps).

TABLE I

ESTIMATED PARAMETERS WITH C_1

muscle		$\hat{l}_{m0i}$	$\hat{l}_{mshi}$	$\hat{V}_{shi}$
Triceps	x_{init}	0.216	0.018	0.3
	$\hat{x}$	0.226	0.018	0.97
Brachioradialis	x_{init}	0.212	0.035	0.3
	$\hat{x}$	0.188	0.037	0.316
Brachialis	x_{init}	0.142	0.035	0.3
	$\hat{x}$	0.124	0.021	0.1

TABLE II

ESTIMATED PARAMETERS WITH C_2

muscle		$\hat{l}_{m0i}$	$\hat{l}_{mshi}$	$\hat{V}_{shi}\hat{V}_{shli}$	$\hat{V}_{mli}$
Triceps	x_{init}	0.216	0.018	0.12	1.3
	$\hat{x}$	0.214	0.010	0.080	0.257
Brachioradialis	x_{init}	0.212	0.035	0.12	1.3
	$\hat{x}$	0.171	0.042	-1.29	1.014
Brachialis	x_{init}	0.142	0.035	0.12	1.3
	$\hat{x}$	0.125	0.024	0.002	0.147

C. Interpretation and discussion

The results given in Table I and Table II show that the estimation can be considered as successful. The Newton-Gauss non-linear least square algorithm converges for each muscles with few computation steps. Figures 8, 9 and 10 are direct validation of the results. The muscle force computed with the estimated parameters (red dashed line (top)) and the one computed by optimization of the inverse dynamics (blue line) are much more alike than the one computed with the parameters found in the literature (black dashed line (bottom)). However the movement can be separated in three sequences for the weak muscles: Triceps and Brachioradialis (Fig.8 and 9). First a sequence of warm up, then a sequence of regular behavior and finally a sequence of fatigue of the muscle. During the warm up and the

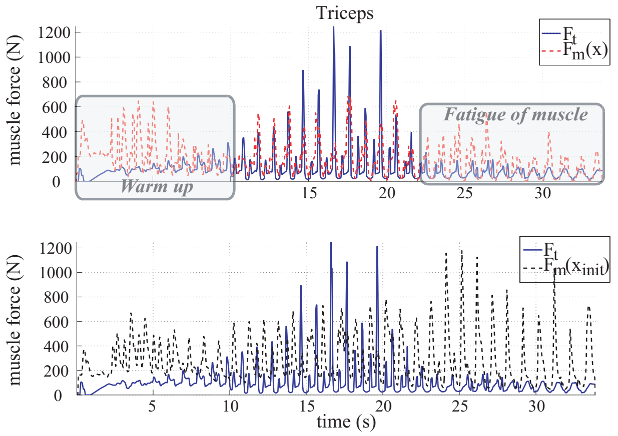

Fig. 8. Comparison with the muscle force computed with musculo-skeletal model and with the muscle model, for the Triceps

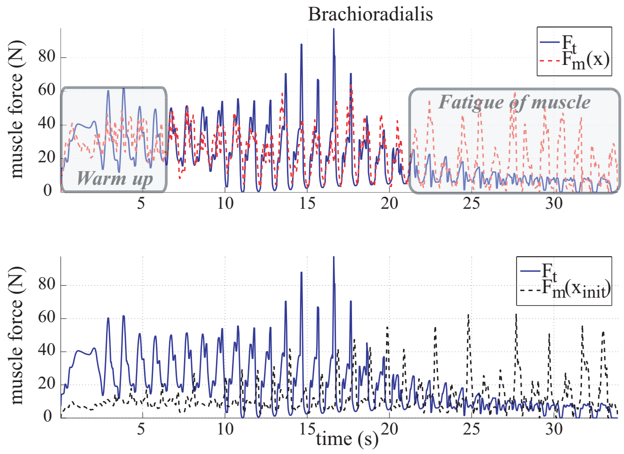

Fig. 9. Comparison with the muscle force computed with musculo-skeletal model and with the muscle model, for the Brachioradialis

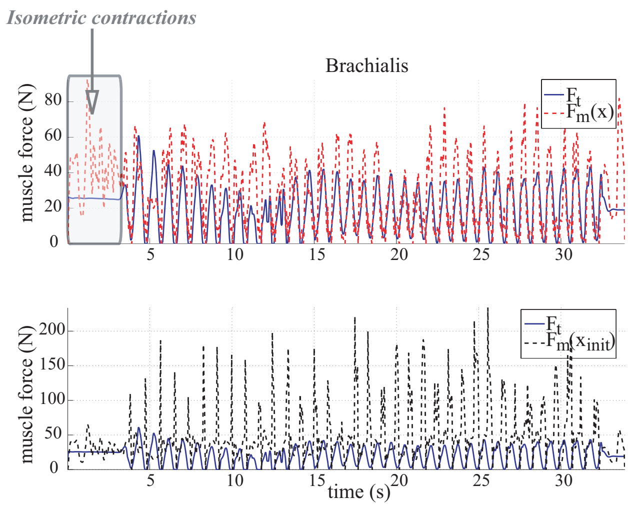

Fig. 10. Comparison with the muscle force computed with musculo-skeletal model and with the muscle model, for the Brachialis

fatigue the behavior of the muscle is different than the one described by the Hill-Stroeve model given by (2). The warm up behavior is not clear, though the fatigue shows that even with full contraction of the muscle the force generated is lower than the one expected due to the change of chemical reactions in the muscle. It is then important to note that the model used is limited to the description of muscle behavior for warmed up and no fatigue muscles. The Brachialis records Fig.10 show a phase of isometric contractions before the subject begins the requested movements. As mentioned above, the computation of the tendon force cannot take it into account since there is no corresponding movement.

Parameters $\hat{l}_{m0i}$ and $\hat{l}_{mshi}$ are estimated with both criteria. The same values for each of the cases are found which shows the correctness of the model (ie. in the value of v_{maxi}.

Despite, those results need improvements. The model is rather simple and does not take into account the full dynamics of the system: passive force, pennation angle, variations of optimal muscle length l_m^0 according to the level of activity [2]... This step is only treating the problem of muscle parameters, although the tendon slack length l_{t0} is very influent in the musculo-tendon dynamics. Conditions and assumptions of the tests: $F_{ti} \approx F_{mi}$ are restrictive and difficult to check and need to be enhanced.

VI. Conclusion

These preliminary experimental results have shown that with a simple model of musculo-tendon complex the identification of the subject specific parameters can be performed. Contrarily to usual literature in this field the Hill-based model parameters are estimated, and not only some global parameters with little physiological meaning. The proposed method is based on the measurement of the muscle dynamics using EMGs data and motion capture system with simple tasks repeatable by everyone: flexion-extension of the elbow joint in the horizontal plan. Estimation is carried on for each muscle involved. The system obtained is a multi-variable non-linear over determinate system. It is solved by linear least square method with Newton-Gauss algorithm. The results have clearly shown that the model is not enough precise to describe the fatigue of weak muscles. This latest occurs very soon when then subject is asked to contract firmly and repeatedly those muscles. The isometric force should be estimated too.

Once those subject specific parameters are estimated, it will be possible to add the muscle model to the musculo-skeletal dynamic model used for the computation of the inverse kinematics and inverse dynamics and to improve the results of the computed musculo-tendon force, to take into account the cases of co-contractions and isometric contractions. Such results also allow to improve the knowledge of the human dynamics and of the muscle constraints for example during some specific movements. It is also very promising to apply such a method to people who have suffered injury to check their recovery, or to people with muscle disease. Further works concern improvements in the model to fit more easily with the subject and to be more physiologically correct. Cross validations consisting in computing the muscle force with the muscle model and comparing it with the muscle force estimated by the inverse dynamics are planned. Therefore such results can be used by medical doctors studying sport science, rehabilitation, muscle diseases as they allow to understand, precisely simulate and control the muscle dynamics not from averaged data measured on a population of chosen subjects but with subject specific parameters.

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References

[1] M. Yamakita, N. Kamamichi, Y. Kaneda, K. Asaka, and Z.W. Luo, "Development of an artificial muscle linear actuator using ionic polymer-metal composites," *Journal of advanced robotics*, vol. 18, no. 4, pp. 383–399, 2004.

[2] D.G. Lloyd and T.F. Bessier, "An emg-driven musculoskeletal model to estimate muscle forces and knee joint moment in vivo," *Journal of Biomechanics*, , no. 36, pp. 765–776, 2003.

[3] S. Stroeve, "Learning combined feedback and feedward control of a musculoskeletal system," *Biol. Cybern.*, vol. 75, pp. 73–83, 1996.

[4] K. Yamane, Y. Fujita, and Y. Nakamura, "Estimation of physically and physiologically valid somatosensory information," in *Proc. of the Int. Conf. on Robotics and Automation*, 2005.

[5] F.E. Zajac, W.S. Levine, J. Chapelier, and M.R. Zomlefer, "Neuromuscular and musculoskeletal control models for the human leg," in *Proc. of the 1983 American Control Conf.*, June 1983, pp. 229–234.

[6] S. Stroeve, "Impedance characteristics of a neuromusculoskeletal model of the human arm i. posture control," *Biol. Cybern.*, vol. 81, pp. 475–494, 1999.

[7] D.G. Thelen, F.C. Anderson, and S.L. Delp, "Generating dynamic simulations of movement using computed muscle control," *Journal of Biomechanics*, in press.

[8] D.G. Thelen, "Adjustment on muscle mechanics model parameters to simulate dynamic contractions in older adults," *Trans. on ASME*, vol. 125, pp. 70–77, February 2003.

[9] M.E. Zobitz, A.M. Halder, and K.-N. An, "Structural properties of the subscapularis, supraspinatus andinfraspinatus tendons of the rotator cuff," in *Bioengineering Conference ASME*, 2001, vol. 50, pp. 365–366, only for the tendon stiffness coefficients.

[10] M. Hirashima, H. Kadota, S. Sakurai, K. Kudo, and T. Ohtsuki, "Sequential muscle activity and its funcional role in the upper extrimity and trunk during overarm throwing," *Journal of Sport Science*, vol. 20, pp. 301–310, 2002.

[11] K. Yamane, T. Kuroda, and Y. Nakamura, "High-precision and high-speed motion capture by combining heterogeneous cameras," in *Proceedings of IEEE/RSJ International Conference on Intelligent Robots and Systems*, Sendai, Japan, September–October 2004, pp. 279–286.

[12] K. Yamane and Y. Nakamura, "Natural motion animation through constraining and deconstraining at will," *IEEE Transactions on Visualization and Computer Graphics*, vol. 9, no. 3, pp. 352–360, 2003.

[13] R. Osu, D.W Franklin, H. Kato, H. Gomi, K. Domen, T. Yoshioka, and M. Kawato, "Short- and long-term changes in co-contraction associated with motor learning as revealed from surface emg," *Journal of Neurophysiology*, vol. 88, pp. 991–1004, 2002.

[14] G Venture, K Yamane, and Y Nakamura, "Identifying musculo-tendon parameters of human body based on the musculo-skeletal dynamics computation and hill-stroeve muscle model," in *IEEE-RAS Int. Conf. on Humanoid Robots*, Tsukuba, Japan, December 2005.

[15] J.E. Jr. Dennis, "Nonlinear least-squares," *State of the Art in Numerical Analysis*, vol. ed. D. Jacobs, Academic Press, pp. 269–312, 1977.