Physiological Muscle Forces, Activation and Displacement Prediction During Free Movement in the Hand and Forearm

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***Abstract* — This paper deals with development of a highly realistic human hand and forearm model. The model contains 38 muscles and 24 degrees of freedom representing the joints of the system. The adopted model has to be as close as possible to the reality of the human being hand, to address several features linked to manipulation tasks, grasping objects and daily routine movements like shaving, writing, etc. In addition, thanks to this model, a better comprehension of the biomechanical and neuromuscular behavior of the system is aimed. This will allow having a tool for the simulation of repairing surgery, acts such as tendon transfer. In this paper, we focus on the muscle forces determination for a given task. An optimization technique is used to resolve the redundant problem over the 24 joints of the system. Also, a muscle model is used and is integrated in the optimization technique, in order to determine measurable values, like the activation and the displacement of each muscle in the system. The calculation is made during a real-time simulation. The forces and the activations are compared to the electromyography (EMG) signals measured on a subject in vivo, by surface electrodes. A very close similarity between the theoretical and the experimental results is found.**

1. Introduction

Proper understanding of the human hand motion is a very challenging task. Knowledge of muscle forces and their action on the body is fundamental for improving the diagnosis and treatment of persons with movement disabilities. The work presented in this paper is a part of the SHARMES simulator, see Fig. 1. The main objectives of the simulator consist on producing a highly realistic simulation platform. This includes not only realistic movement but also a complete dynamic and biomechanics high levels of realism. One of the major challenges of the SHARMES simulator is to carry out all simulations in real time. This condition plays a great role in chirurgical applications (repairing, tendon transfer …) and virtual reality (haptic devices …). Indeed, carrying out a simulation at the same rate of the human being real motion deeply increases the realism of the simulation of the physical phenomenon.

At first only the muscle forces are estimated during free hand movement. Inverse dynamics can be used to estimate the external load applied to a joint. However, the contribution from muscles to generate this load is far more difficult to determine. The human musculo-skeletal system is normally redundant in that more muscles cross a joint than there are degrees of freedom at the joint. One solution to this problem is the use of EMG data in conjunction with an appropriate muscle model to estimate the forces produced in each muscle (Lloyd G.D. et al., 2003). Unfortunately, this method cannot be applied for this work, since the hand contains 38 muscles and 20 degrees of freedom, compared with the legs that have about 13 muscles and 3 degrees of freedom. Also the muscles of the hand are much smaller than the muscles of the legs, hence, are less accessible, which make it more difficult to measure the EMG signals of certain muscles using surface electrodes. Indeed, the muscles of the hand are crossed with each others in the forearm, this phenomena known as Cross-Talk makes it very difficult to differentiate between the measured EMG signals for each muscle in the system during a given movement. However, some EMG muscles in the hand can be measured, and will be used later for validation purpose. Another method for estimating muscle forces is the optimization technique.

Many studies were done in this subject, for instance, the one presented by (Chao E. Y., 1978) predicts the forces in the normal and abnormal hand during isometric hand function, mainly during the pinch position. Another, presented by (Zatsiorsky M. V. et al., 2002), studies the force and torque production in static multifinger prehension.(Chao E. Y. et al., 1976) developed a three-dimensional force analysis of finger joints in selected isometric hand functions for the pinch task. Nevertheless, all these studies were done in static mode where no movement of the joints is produced. In this paper the dynamic movement of the system joints is taken into account. The computation of the forces is made, on line, during the movement of the hand.

Some of the resulted muscle forces will be compared with EMG recordings, (Pedotti A. et al., 1978) reported a nonlinear optimization technique which computed muscle forces that agreed with the EMG recordings during walking. However, only a qualitative comparison could be made between the estimated muscle forces and the EMG data measured with surface electrodes.

Another approach for the validation of the individual muscle forces obtained by optimization involves modeling the muscles. In most of the studies that apply muscle model in their computation (Crownninshield R.D. et al., 1981; Zajac F.E., 1989) used the EMG recordings as an input to their system from which they calculated the forces in the legs. Unfortunately, as said before, this method is not doable in the hand's muscles calculation, since not all muscles activities are accessible by surface electrodes. Muscle model is, instead, used in order to integrate variables like the activation and the deformation of the muscle in the optimization algorithm which will lead to a theoretical estimation of the activation of each muscle.

Some of these activations will be compared with the measurable EMG recordings, the method of validation will be given later (see text for details). This will give us a quantitative comparison with the experimental values of the activation.

This work will provide a good validation of the entire model used in the SHARMES simulator, thus proving a good adequacy in the simulation of the free movement of the hand and the forearm.

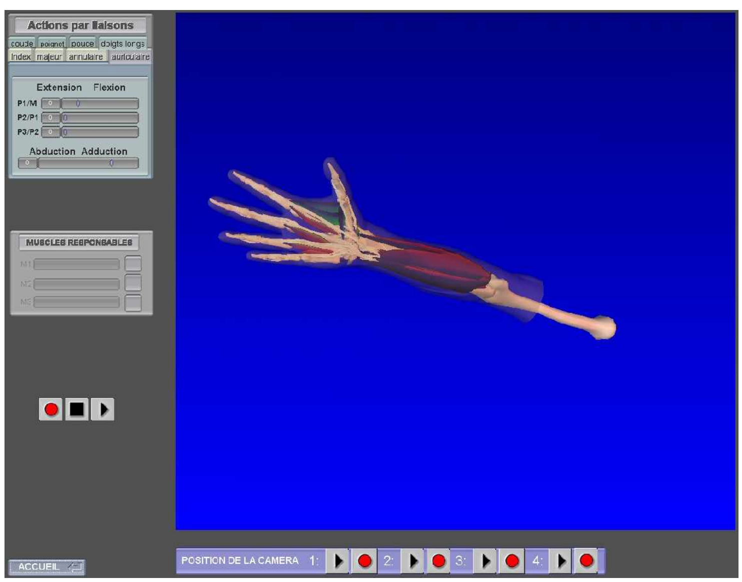

Figure 1. SHARMES simulator

2. Anatomical model

All the muscles of the hand and the forearm are taken into consideration, for the exception of the lumbrical muscles. These muscles were not modeled due to their complex behavior. The entire system comprises 38 muscles acting on four independent parts: the forearm, the wrist, the thumb and the four fingers. The four fingers are related by the common extensor (EDC), the tendon of this muscle branches into four tendons on the level of the wrist joint, to attach to the middle and distal phalanx of each finger. Muscles of the hand set in motion numerous hand joints. The joint geometry defines the degrees of freedom (DoF) of the HHF. There are two DoF in the forearm (flexion/extension and pronation/supination), two DoF in the wrist joint (flexion/extension and abduction/adduction), two DoF in the carpo-metacarpal joint of the thumb, one DoF in the metacarpo-phalangeal joints of each finger (flexion/extension and abduction/adduction), and one DoF in each interphalangeal joint (flexion/extension), which makes a system of 24 DoF. Figure 2 shows the osseous members of the system and the correspondent joints; figure 3 shows the degrees of freedom for each joint.

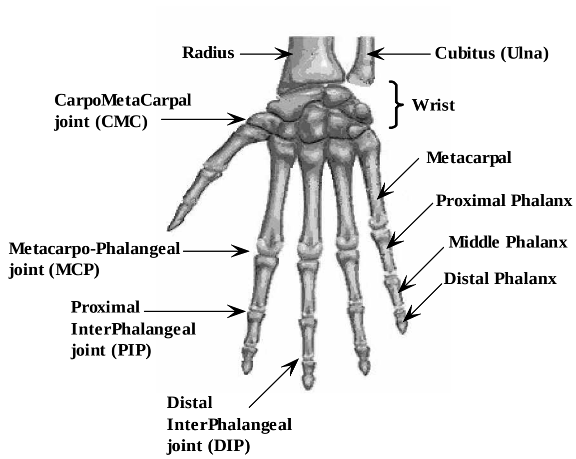

Figure 2. Osseous members and joints of the HHF

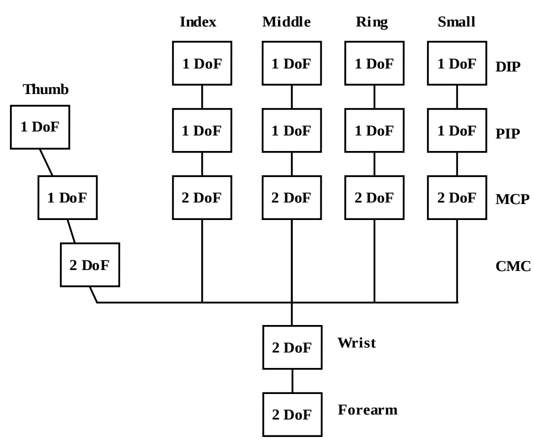

Figure 3. Degrees of freedom (24)

While some studies (Challis H.J. et al., 1994) developed a procedure which permits the estimation of some muscle model parameters directly on the subject. Other studies examining the muscle forces in human movement base their muscle model parameters on mean morphological data gathered from cadavers (Pandy M.G. et al., 1991). In our study all the muscle model parameters (insertion points, muscle rest length, cross-sectional area …) were acquired from a cadaver. The anatomical complexity of the muscles and the skeletal structure makes it imperative to make some mechanical simplifications. Following the work done by (Pedotti A. et al., 1978) we modeled the muscles by lines joining the point of origin to the point of insertion of the muscle. When the insertion or the origin covers a large surface, the approximate midpoint of the surface has been used. The overall length of the muscle includes the length of the tendon as well. The biological tissues, by which the extrinsic flexors pass in, are modeled as pulleys, and the force of action is taken to be the middle of the bone that contains them, refer to (Chalfoun J. et al., 2004) for more details.

3. Muscle forces prediction

3.1. Force-torque relation

Knowing the anatomical details of the points of insertions and origin of the muscles and their lengths, the torque arms of the muscles at the different joints have been computed so that the torques can be related numerically to the values of the forces.

These forces are related to the joints torques by the following relation:

$$[T_{lt}] \cdot [F_t] = [\tau] \quad (1)$$

Where $[F_t]$ is the vector of the tendons forces (45 columns), $[\tau]$ is the vector of the joints torques (24 columns) and $[T_{lt}]$ is the matrix (24 x 38) representing the relation between these two vectors. For details about the dynamical calculation of the joint torques and the calculation of the matrix $[T_{lt}]$ refer to (Chalfoun J. & al., 2004).

3.2. Optimization and objective function

Due to the redundancy at the joints, an optimization technique is used to calculate one solution to this infinite problem by minimizing an objective function. The objective function is considered to reflect a mechanism via which the human body recruits muscles to produce a joint moment. When such a technique was first used (Seireg A. et al., 1973), the objective function minimized the sum of the muscle forces and the weighted sum of the moments acting at the joint. Later on, several studies were made using different optimization criterion; ones such a physiologically based criterion as the maximum endurance of musculoskeletal function (Crownninshield R.D. et al., 1981). Although this criterion was the closest one to a physiological meaning, the authors did not find results that matched with the EMG recordings. Another researches (Pedotti A. et al., 1978; Challis H.J., 1997) investigated a number of objective functions: two relating to muscle stress and two relating to relative muscle forces, both of the authors found that the objective function used in equation 2 coordinated the most the EMG recordings. Therefore in this paper the objective function to minimize is the square relative sum of the muscle forces:

$$f = \sum_{i=1}^{n} \left(\frac{F_i}{(F_{\max})_i} \right)^2 \quad (2)$$

Normalization greatly reduces the magnitude of the objective function and avoids some numerical problems in large scale optimization. The equality constraint was that the sum of the product of the muscle forces and their moment arms should equal the muscle moment coming from inverse dynamic (equation 1). The inequality constraints are the limit of the muscle force between zero and the maximal muscle force that occurs at the optimal muscle fiber length and at maximum activity. The optimization method used is the Lagrange multipliers with penalty on the inequality constraints. For more details on the method used and the equations refer to (Chalfoun J. et al., 2004).

The model was tested on several system movements, namely, shaving, writing … however, the movements that were used in order to verify the accuracy of the solution were basic movements like opening and closing the hand and the pinch movement. These movements were chosen by a surgeon for a particular reason; during the opening and closing of the hand the EDC muscle is activated in order to extend the long four fingers, but in the pinch movement the Extensor Indicis (EI) of the index is activated and he’s the muscle responsible for the extension of that finger, while the EDC is normally inactive during the pinch movement.

These two movements are depicted in the following figures.

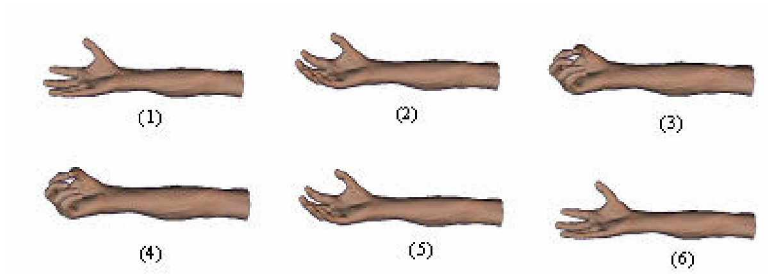

Figure 4. Opening-Closing hand movement

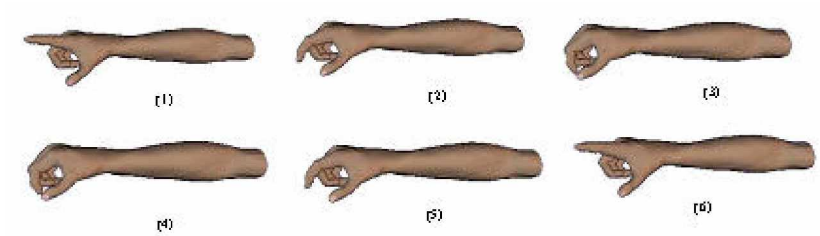

Figure 5. Pinch Movement

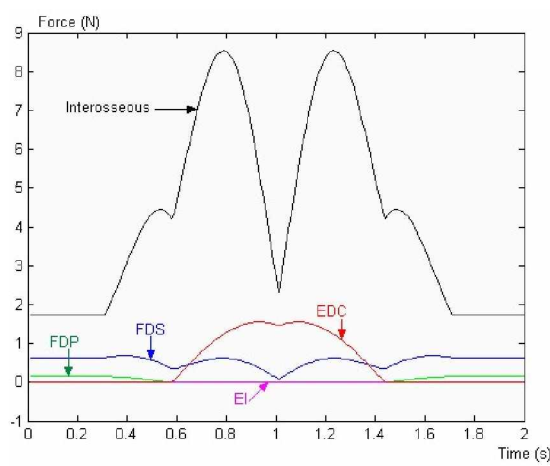

Figure 6. Index muscle forces during opening closing

The movements are divided into four phases: in the first period the entire HHF system is maintained stable for 0.3 sec. The second phase starts when the hand begins to close on itself following the sequence 1 - 2 - 3 in Fig. 4, this movement takes 0.7 sec to be accomplished. In the third phase, which has also a duration of 0.7 sec, the hand starts opening back, by following the order 4-5-6 (see fig. 4), to reach the fourth and final period where the hand is maintained in the same position (isometric) for 0.3 seconds. The time-step of the calculation is set at $\Delta t = 0.01$ sec . The forces values are observed and interpreted on the following graphical plots.

As expected the results found for the index finger match the expectations of the surgeon. However till this level only a qualitative comparison between the theoretical results and the experimental ones can be made. Therefore as said in the beginning of the paper, a muscle model is chosen in order to integrate measurable values like the activation and the displacement of the muscles so that a quantitative comparison could be made for some of the muscles.

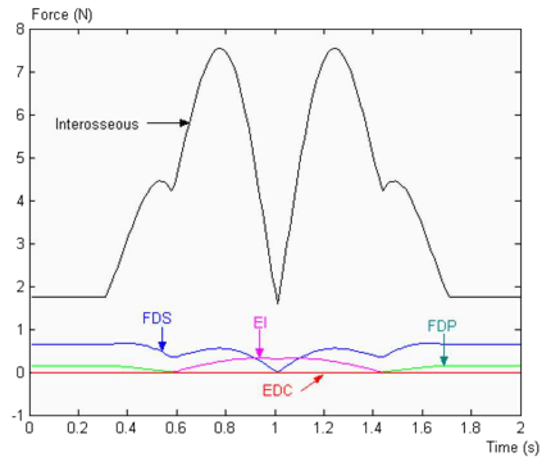

Figure 7. Index muscle forces during the pinch movement

4. Muscle model (Hill type muscle model)

Hill type muscle model is used to integrate in the optimization algorithm the activation and the displacement of each muscle as variables to optimize. The muscle tendon unit was modeled as a contractile element (E_c) in series with a passive element (E_s) and in parallel with a passive element (E_p) (Zajac F.E., 1989). The series element was neglected in our study. Thus, the total muscle force F^M is the sum of a passive force F^{PE} and an active force F^{CE}, see figure 8.

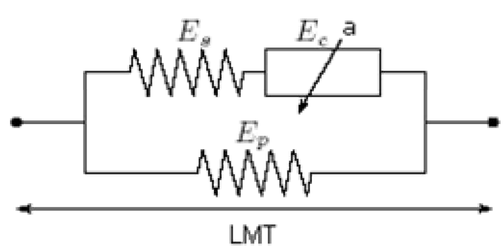

Figure 8. Hill type muscle model

Since the study in this paper is a quasi-dynamic one, the velocity of the contraction was not taken into account. Only a force-length relationship was established. We considered that all muscle fibers are oriented in the direction of the tendon. Thus, all muscles are considered not to be pennated ($\alpha = 0$), for the pennation angle is a function of the length and thus introducing this variable into the algorithm will complicate the calculation. The forces were normalized to the maximum isometric force for each muscle. The general form of the equation for the force produced by the muscle-tendon unit is given by (Audu M.L. et al., 1985):

$$F^M = F^T = F^{\max}\left[a(t)\,f(l) + f_p(l)\right] \quad (4)$$

The force-length relationship in the contractile element is given by (Walter M., 1999):

$$\begin{aligned} f(\varepsilon) &= e^{-\left[\frac{(\varepsilon+1)^{0.96343\times(1-\frac{1}{ia})}-1}{0.35327\times(1-ia)}\right]^2} && ia < 1 \\ f(\varepsilon) &= e^{-[2.727277\times\ln(\varepsilon+1)]^2} && ia = 1 \end{aligned} \quad (5)$$

Where $\varepsilon = (l - l_0)/l_0$ is the tensile strain of the muscle, l is the muscle fiber length at time *t*, l_0 is the optimal fiber length at which maximum muscle force could be obtained, *ia* is a muscle constant determined from (Walter M., 1999).

(Huijing P.A., 1996) has shown that optimal fiber lengths increase as activation. This coupling between activation and optimal fiber length was incorporated into our muscle model using the same equation used by (Lloyd G.D. et al., 2003):

$$l_0(t) = l_0(\gamma(1 - a(t)) + 1) \quad (6)$$

Where γ is the percentage change in optimal fiber length, the value of this factor was fixed by (Lloyd G.D. et al., 2003) to be $\gamma = 0.15$, *a(t)* the activation at time *t*, l_0 is the optimal fiber length at maximum activation, $l_0(t)$ is the optimal fiber length at time *t* and activation *a(t)*.

The parallel passive elastic muscle force $f_p(\varepsilon)$ was obtained from an exponential relationship used by (Walter M., 1999), and has the following form:

$$f_p(\varepsilon) = b_1 \times e^{b_2.\varepsilon} - b_1 \quad (7)$$

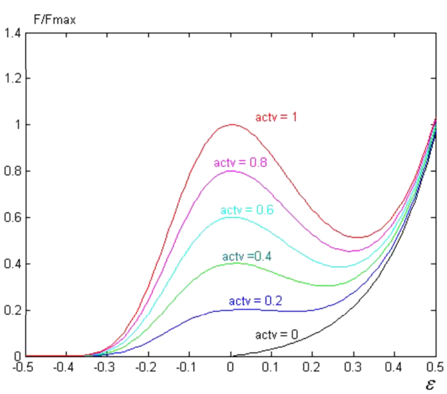

Figure 9. Normilized force produced by muscle

Where b_1 and b_2 are muscle constants determined by (Hill A.V., 1938), and taken to be 0.03 and 7 respectively. It has been proven (Garner A.B. et al., 2003) that the effective operating range of muscle begins at roughly $0.5\,l_0$ and ends at $1.5\,l_0$; muscle cannot generate active force beyond these length. This will give us a displacement range of $-0.5 < \varepsilon < 0.5$. In addition, when muscle is stretched to lengths greater than $1.2\, l_0$, it generates a significant amount of passive force. The total muscle force is represented in figure 9 for different activation values.

After integration of the muscle model into our system, the problem will yield as follows: minimize the objective function $C = \sum_{i=1}^{m} \left(\frac{F_i(\varepsilon_i, a_i)}{F_{i\max}} \right)^2$ constrained by 24 equality equations $[T_{lt}][F(\varepsilon, a)] = [\tau]$ and 76 inequalities $-0.5 \le \varepsilon \le 0.5$ et $0 \le a \le 1$. This problem was solved using the LaGrange multipliers and the nonlinear set of equations was solved using a relaxed Newton method.

5- The simulation's results

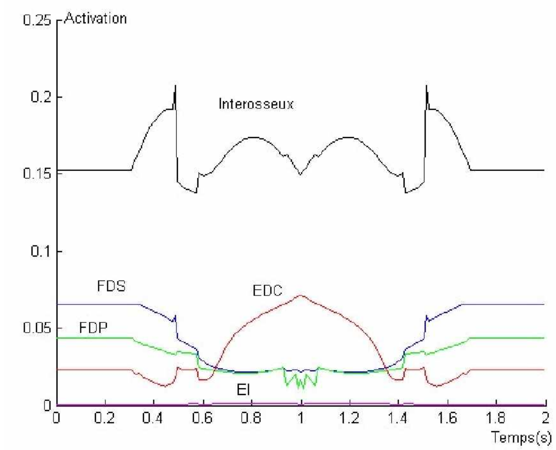

Figure 10. Activation of the index muscles in the first movement

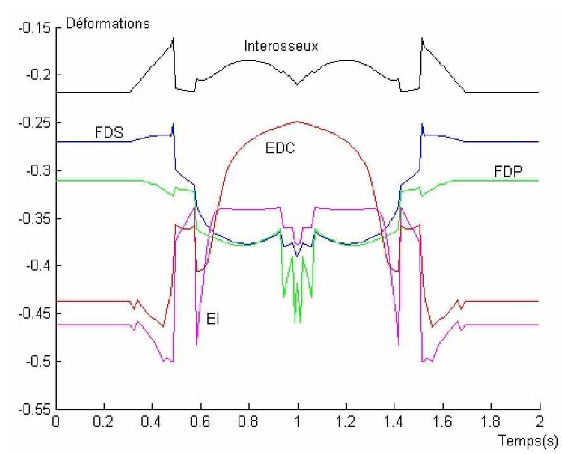

Figure 11. Deformation of the index muscles in the first movement

The solution of such problem will give the activation and the tensile strain of each muscle. The forces will be calculated using equations 5 and 7. The exact same forces were found after the integration of the muscle model, which is predictable, since the same objective function was used.

However, besides the knowledge of the muscle forces we now possess two additional muscles parameters, the activation and the displacement. Hence, we can now validate the results found with the optimization technique by comparing the activation with the EMG recordings using two surface electrodes.

In following figures are depicted the deformation and the activation of the muscles of the index finger. The focus on the results of the index finger in this paper is due to the fact that this finger possesses two extensor muscles, one proper to it and the other is the common extensor finger.

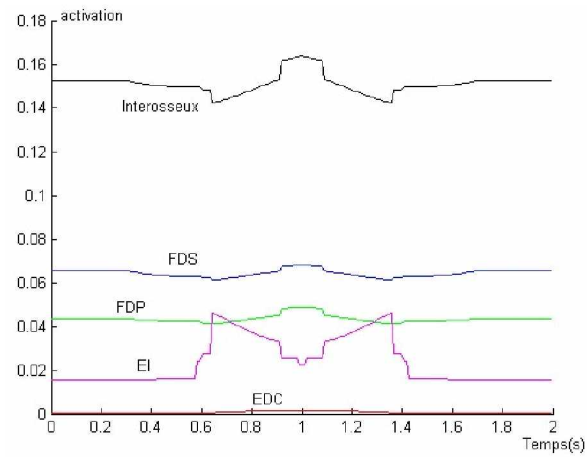

Figure 12. Activation of the index muscles in the Pinch movement

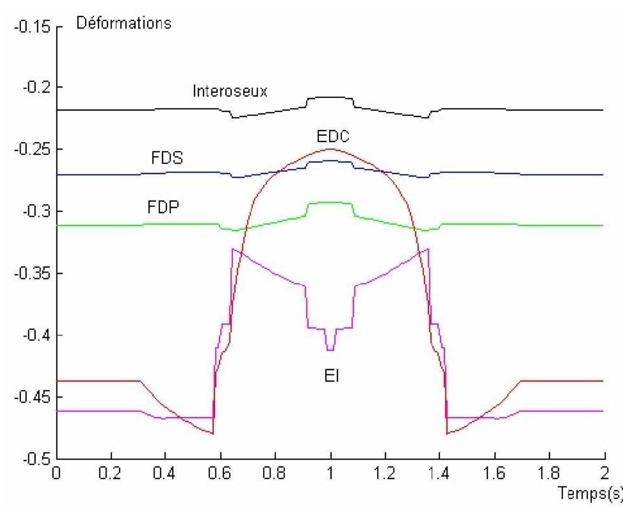

Figure 13. Deformation of the index muscles in the Pinch movement

As we can notice in figure 10 the EI muscle isn't activated while the EDC is. Inversely in figure 12 the EDC is barely activated while the EI.

6. Validation

As said in the beginning, the muscles of the hand are crossed with each others in the forearm, which makes it very difficult to measure the EMG recordings for all the muscles in the system during a random movement. Therefore, we chose specific movements in order to isolate some muscles and consequently be able to measure their EMG recordings. We used the same basic movements as before (opening-closing and the pinch) since these two movements highlight some of the muscles in the system like the EDC, the EI and the FDS muscles.

6.1. EMG-activation model

The function used to account for the linear or non-linear EMG to force relationship is that used by (Lloyd G.D. et al., 2003):

$$a_j(t) = \frac{e^{Au_j(t)} - 1}{e^A - 1} \quad (3)$$

Where $a_j(t)$ is the activation of the muscle *j*, $u_j(t)$ the post-processed EMG of the muscle *j* at time *t*, *A* the non-linear shape factor, constrained to $-3 < A < 0$, with 0 being the linear relationship. (Lloyd G.D. et al., 2003) has done a research on six different subjects with different *A* factor by subject. The mean value found is -1.0105, hence we chose a value of -1 for the non-linear shape factor.

6.2. Discussion

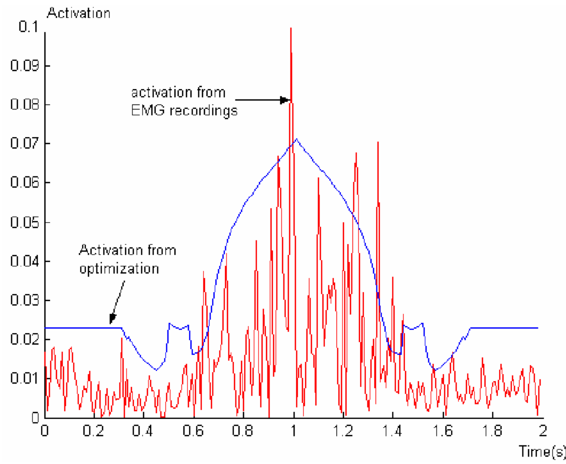

Figure 14. Activation compoaraison for the EDC muscle

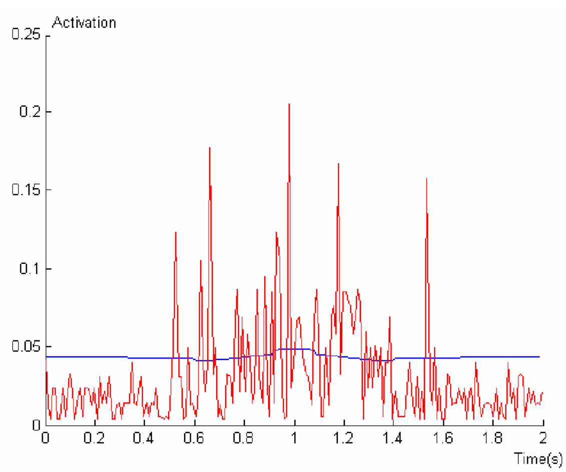

Figure 15. Activation compoaraison for the FDS muscle

The results obtained for the closing and opening of the hand for the EDC muscle matched with the experimental activation values obtained from EMG recordings (figure 10). As we can see, a very high similarity between the theoretical and the experimental data was found. However, this is not generalized for the entire system, for as we can see in figure 11, there was not a high similarity between the activation of the FDS and the EMG recordings.

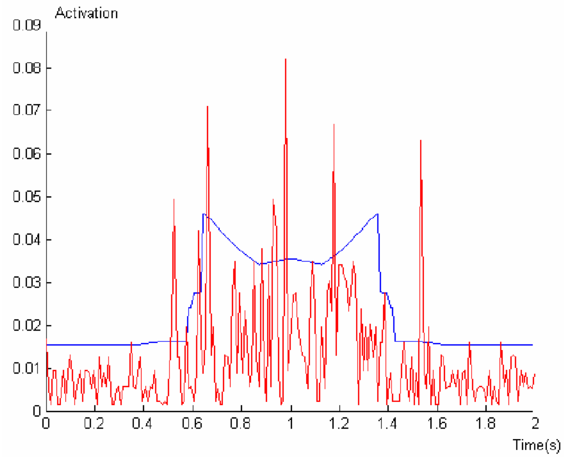

Figure 16. Activation compoaraison for the EI muscle

For this muscle the activation found was different from the recordings, but is also close to its value, meaning that the solution didn't divert. This difference is due to several reasons, for instance, the simplifications made on the muscle model (pennation angle…) and the fact that we didn't take into account the velocity of the contraction. Also, the EMG recordings for the FDS are less precise than those of the EDC since more than one flexor muscle is concerned during the movement, which makes it more difficult to isolate the signal of that muscle, while the EDC is the only major muscle concerned in the extension of the hand. The EI muscle however (figure 12), in the pinch movement, was accessible by surface electrodes since the pinch movement isolates that muscle in extension, and the theoretical values coordinate greatly with the experimental ones. This kind of validation procedure was made on several other muscles and also, a high similarity between the theoretical and the experimental activation was found for some muscles, while for the others there was a slight disagreement between the theoretical and experimental activation.

7. Conclusion

The presented work concerned the muscle forces prediction in the hand and the forearm, to produce highly realistic simulation. The model contained 38 muscles and 24 degrees of freedom representing the joints of the system. The forward and inverse problems were formulated. An optimization technique was adopted to predict the muscle forces in the system. The calculation of these forces was made according to the real-time requirements. The model was illustrated through a basic movement of the hand. Accurate results were found. However, a muscle force is not a measurable value, and due to the fact that not all muscles are accessible by surface electrodes to measure the EMG data; a muscle model was used to integrate in the optimization technique measurable variables like the activation and the displacement of each muscle in the system. The forces and the activations are compared to the EMG recordings measured on a subject by surface electrodes. A very close similarity between the theoretical and the experimental results was found for some muscles while for others there was a slight disagreement, this is due to some modification taken in order to simplify the complexity of the anatomical model of the system. Apparently these simplifications modified somewhat the results and they are to be accounted for in future studies.

This could help increasing the precision of results. Especially, when it comes to medical applications where every detail is necessary for any kind of surgery or hand replacement. This will allow having a simulator capable of reproducing the same hand movement of the human being. This study has many applications other than in medicine, and could interest a lot researchers and companies, namely a pen company could be interested in the writing simulation to know the global effort produced by the subject while writing with a new designed pen. Hence, the new design could be chosen according to the results obtained by simulation, which minimizes the global effort produced.

References

Audu M.L. and Davy D.T. (1985). "The influence of muscle model complexity in musculoskeletal motion modeling." J. Biomechanical Engineering **107**: 147-157.

Chalfoun J., Renault M., Younes R. and Ouezdou F.B. (2004). *Muscle Forces Prediction of the Human Hand and Forearm System in Highly Realistic Simulation*. Iros, Japan.

Challis H.J. (1997). "Producing physiologically realistic individual muscle force estimations by imposing constraints when using optimization techniques." Med. Eng. Phys. **19**(3): 253-261.

Challis H.J. and Kerwin G. D. (1994). "Determining individual muscle forces during maximal activity: Model developpement, parameter determination, and validation." Human Movement Science **13**: 29-61.

Chao E. Y. and Opgrande J.D. (1976). "Three-dimentional force analysis of finger joints inselected isometric hand functions." J. Biomechanics **9**: 387-396.

Chao E. Y., K.N. An (1978). "Determination of internal forces in human hand." journal of the engineering mechanics division **Vol. 104, No. EM1**.

Crownninshield R.D. and Brand R.A. (1981). "A physiologically based criterion of muscle force prediction in locomotion." J. Biomechanics **14**(11): 793-801.

Garner A.B. and Pandy M.G. (2003). "Estimation of musculotendon properties in the human upper limb." Annals of Biomedical Engineering **31**: 207-220.

Hill A.V. (1938). the heat of shortening and the dynamic constants of muscle. Proceedings of the Royal Society of London, Serie B.

Huijing P.A. (1996). "Important experimental factors for skeletal muscle modelling: non-linear changes of muscle length force characteristics as a function of degree of activity." European journal of morphology **34**: 47-54.

Lloyd G.D. and Besier F.T. (2003). "An EMG-driven mussculoskeletal model to estimate muscle forces and knee joint moments in vivo." J. Biomechanics **36**: 765-776.

Pandy M.G. and Zajac F.E. (1991). "Optimal muscular coordination strategies for jumping." J. Biomechanics **24**: 1-10.

Pedotti A., Krishnan V.V. and Stark L. (1978). "Optimization of muscle-force sequencing in human locomotion." Mathematical Biosciences **38**: 57-76.

Seireg A. and Arvikar J. (1973). "A mathematical model for evaluation of forces in the lower extremities of the musculo-skeletal system." J. Biomechanics **6**: 313-326.

Walter M. (1999). 3D modeling of the human upper limb including the biomechanics of joints, muscles and soft tissues. Paris, Ecole polytechnique**:** 182.

Zajac F.E. (1989). "Determining muscle's force and action in multi-articular movement." Exercise and Sport Siences Reviews **17**: 187-230.

Zajac F.E. (1989). "Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control." Critical Reviews in Biomedical Engineering **17**: 359-411.

Zatsiorsky M. V., Robert w. G. and Latash M. L. (2002). "Force and torque production in static multiphinger prehension: biomechanics and control. I. Biomechanics." Biol. Cybern **87**: 50-57.