CHARACTERISATION OF MOMENT-ANGLE RELATIONSHIPS IN FES-ACTIVATED MUSCLES FOR THE RECOVERY OF SIT-TO-STAND

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Abstract – *A method for the identification of dynamic characteristics of the lower leg muscles in paralysed individuals was developed. The set-up was optimised to obtain the joint moments at different joint angles for different muscle groups during the sit-to-stand (STS) movement.*
The subject seated on a custom system, which allowed him to statically reproduce the STS movement positions, while surface FES was used to elicit isometric contractions in quadriceps, gluteus and hamstrings. From the ground reaction force, acquired by a force platform, the active torque-output was computed by modelling the biomechanical system.
The knee joint moment-angle relationships obtained were validated by measuring the forces evoked using a strain gauge based bench.
With the above set-up, the active torque contributions of both mono- and bi-articular muscles, for different combinations of them, was quantified avoiding fatigue effects and in a safe way. This characterisation procedure let torques and activation sequences to be estimated and to be used e.g. for improving FES control strategies.

Keywords: parameter identification, muscle model, FES, paraplegia, human lower limb, sit-to-stand

1. Introduction

The modelling of electrically stimulated muscles in paralysed subject is an active topic of research because resulting models can be used to simulate the effects of FES and to design neuroprostheses controllers [1-4]. Due to the non-linear and time-varying characteristics of stimulated muscles, many research groups developed complex model structures, which are generally based on understanding the phenomenological behaviour of the muscular system (a comprehensive review of models appears in [11]). Different non-linear models of muscle dynamics represent the muscle output as the product of three uncloupled factors: activation, moment-angle and moment-angular velocity (Fig.1).

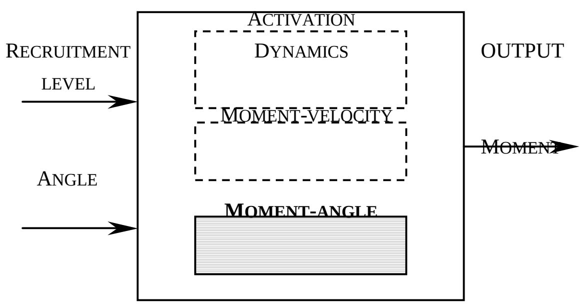


Fig.1 Block diagram of an uncoupled model. The output is the product of three factors: activation dynamics, moment-velocity and moment-angle properties.

Alternative coupled model embeds the moment-angular velocity within the activation dynamics [5-8 ,11]. In any case, it is still present the issue of relating these models to a specific physical system. In order to ensure that a model is appropriate over a wide range of applications, their parameter identification procedures in human beings are usually based on experiments that require long stimulation and data acquisition times [9-12].

As a matter of fact, a different approach regarding modelling, and consequently their parameter identification, is based on considering the muscle behaviour while a specific motor task is performed. In effect, in this case the identification problem can be dealt by an experimental system that maximises the outcomes obtained minimising the amount and duration of experiments.

In this paper, a method for the identification of the moment-angle relationships (the gray block in fig.1) is described that allows the estimation of these relationships for the knee and hip joints by stimulation of quadriceps, gluteus and hamstrings under isometric conditions. The whole procedure was optimised for FES assisted standing up in paraplegic subjects.

2. Methods

The experimental set-up is based on a specially designed system and a force platform to measure the ground reaction force due to the muscle contractions. The system consists of a saddle -where the subject was seated- that is fixed at a moving frame. The frame allows the saddle to move in the sagittal plane on a circular path around the pivot axis.

Such a system allowed the body placement in order to reproduce, in quasi-static conditions, the different joint angular configurations composing the STS movement. Fig.2 illustrates five positions within the range of movement.

Fig 2. The transfer transitions from sitting to standing reproduced by the system used. At each position the pivoting arm was blocked to assure the quasi static conditions, thus each acquisition was performed.

Among the experiments the knee angles - and consequently the hip angles as well - were imposed by fixing the saddle at different heights. In such a way all the STS range was included by knee joint angle intervals of 15°. The respective hip angles were imposed by maintaining the trunk always vertical. A typical angular hip-knee space during standing-up acquired during the standing-up is shown in fig.3.

Joint torques were due to muscular isometric contractions elicited by functional electrical stimulation. Quadriceps, gluteus and hamstrings muscle groups were considered.

An eight-channel computer-controlled stimulator delivered the stimulation pattern, which consist of a train of monopolar current pulses at a frequency of 25 Hz. The maximum pulse duration was of 350 µsec. In the case of co-activation of quadriceps and hamstrings, the pulse durations were modulated according at the envelopes displayed in fig.4(a). The muscle stimulation was provided via surface electrodes that were initially set according to literature. Thus, these positions were modified until sharp contractions were obtained. After electrodes positioning, calibrations of the stimulation current for each muscle were performed to achieve maximal isometric torques. In the healthy subjects, the amplitude was increased until the stimulation started to feel uncomfortable, generally between 45 and 55 mA. Therefore, the warming up times were reduced because of these electrode setting trials.

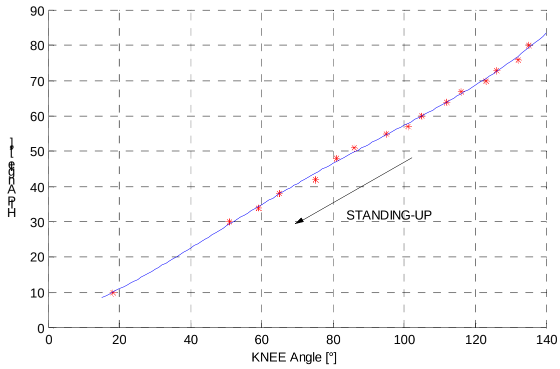


Fig.3 The hip joint angle as a function of the knee angle during the STS transfer by using the experimental set-up described above. The points are the angular positions among the trials performed.

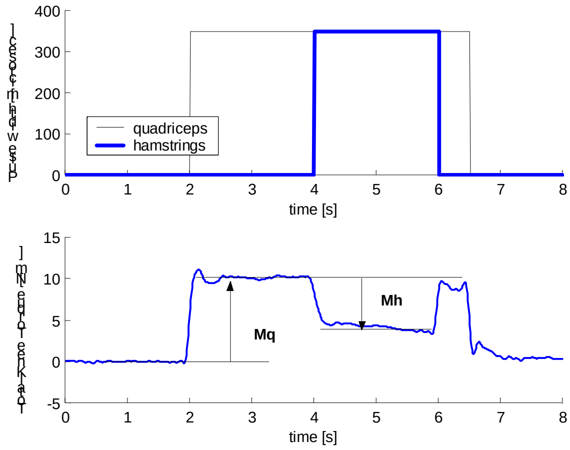


Fig.4 Stimulation pattern: two channels provided the pulse width envelope to quadriceps and hamstrings as shown in (a). A typical acquired moment curve: both the quadriceps moment M_q and the net hamstring moment M_h are displayed in (b)

The forefeet laid on the force platform (Kistler) embedded in the floor and neither support frame nor orthoses were used (Fig.5).

The active torque-output of each joint was obtained from the ground reaction force by means of a planar model of the biomechanical system.

In the model it was assumed that the lower limb joints determined a plane and thus the torques orthogonal to this plane were computed at both the knee and hip joints. Leg geometry and anthropometric parameters were measured, and inter-segmental angles and thigh, shank and foot masses were computed; being in static conditions no inertial effects were considered. A kinetic acquisition system (Elite, BTS) was used for the validation of model and computed angles. In several measurement positions, a bench based on a strain gauge sensor was used to validate the whole procedure.

At each joint, moment-angle relationships were obtained for different combinations of muscular groups' contractions. The curves were normalised by a scaling factor, which takes into account the possible intra-session variability. Thus, the normalised moment curves were fitted by a model that was of the following form:

$$M = exp\left[-\left(\left(\left(\theta/\theta o\right)^{\beta}-1\right)/\delta\right)^{2}\right] \quad (1)$$

Where **M** is the normalised active moment as a function of joint angle $\boldsymbol{\theta}$, and the parameters $\boldsymbol{\theta_o}$, $\boldsymbol{\beta}$ and $\boldsymbol{\delta}$ are the optimal angle, the skewness constant and the width constant of the active moment curve, respectively.
A non-linear least square algorithm was used to identify the optimal parameters $\boldsymbol{\theta_o}$, $\boldsymbol{\beta}$ and $\boldsymbol{\delta}$ to fit this form to each muscle tested.

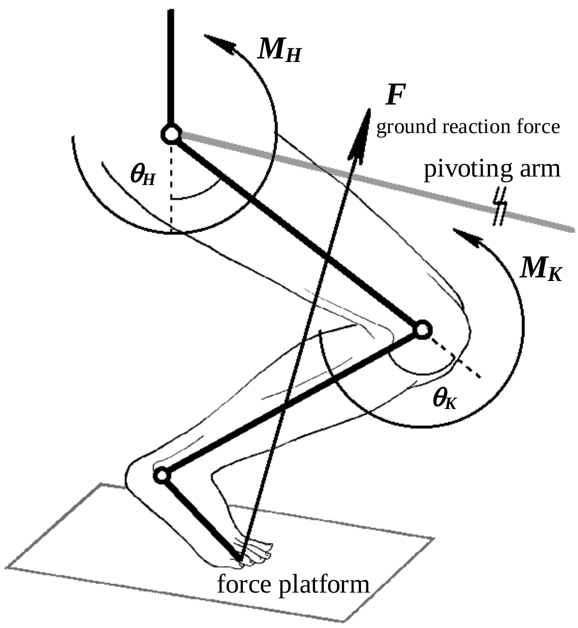


Fig.5. The pivoting arm support the trunk allowing the foot to rest in the force platform that acquire the ground reaction forces before and during the stimulation of the leg muscles. As well angles (θ_H and θ_K) and active moments (M_H and M_K) convention are shown.

3. Results

Results are presented for the identified moment-angle relationships at both knee and hip joints. The figures display typical results from specific trials in one healthy experimental subject.

A typical measured active moment due to quadriceps and hamstrings contractions is illustrated in Fig. 4(b). The net hamstrings contribute M_h to the joint torques was obtained by considering the difference between the phase in which both muscles are contracted and the phase in which the only quadriceps was stimulated M_q.

The results of the parameter identification for one typical trial are tabulated in Table I and II and shown graphically in Fig.7 and Fig.8 for knee and hip joint, respectively. The standard error SE decreased from a peak of over 5 % of normalised moment to as low as 0.3%.

Fig.6 shows a quadriceps normalised moment-angle curve, measured using this set-up, at knee joints. For validation, these curves were compared with resulting curve from data acquired using the force sensor. The differences, due to difficulties in reproducing the identical positions with each set-up, were less than 5% of maximum values.

Fig.7 illustrates, for the three considered muscles, the knee joint moment normalised with respect to the maximum value vs. the knee flexion angle. As it was expected, being the gluteus a mono-articular muscle, it was found that its contribution to the knee moment was negligible.

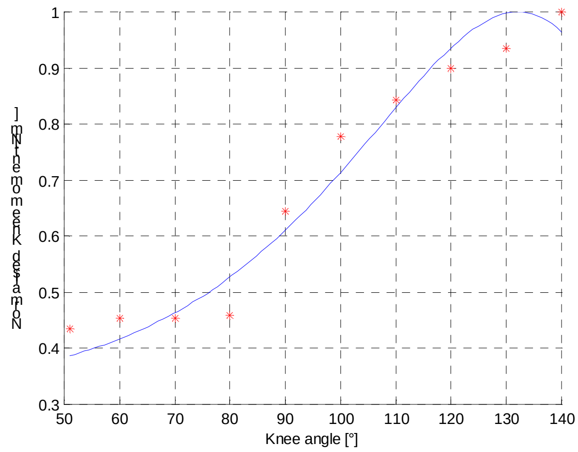


Fig.6 The knee joint moments for the quadriceps muscles vs. the knee flexion angle. The points are the acquired data and the curve displayed here was obtained by least square error algorithm. The parameters are tabulated in Table 1.

TABLE I KNEE JOINT MOMENT PARAMETERS

KNEE JOINT	θ_o	β	δ	SE
QUADRICEPS	132.2019	2.9381	0.9639	0.0027
HAMSTRINGS	102.0902	1.8910	0.5987	0.0093
GLUTEUS	133.3424	10.8527	0.9062	0.0562

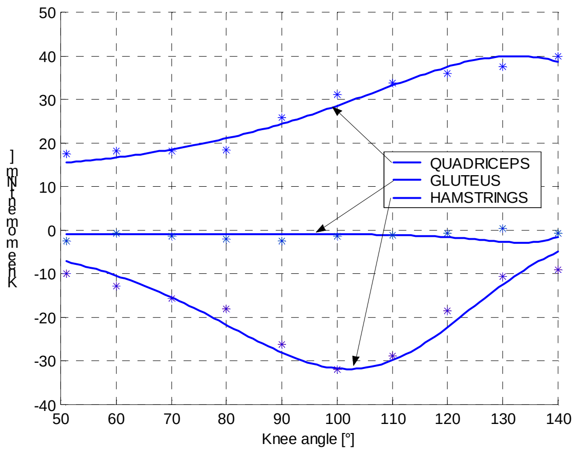
Typical data for one healthy subject, 36 yr, 56kg, 1.63m.

TABLE II HIP JOINT MOMENT PARAMETERS

HIP JOINT	θ_o	β	δ	SE
QUADRICEPS	118.0949	0.0164	0.0074	0.0057
HAMSTRINGS	105.6932	1.6206	0.4481	0.0242
GLUTEUS	128.2553	4.1611	0.7692	0.0278

Typical data for one healthy subject, 36 yr, 56kg, 1.63m.

Fig.7 The knee joint moments for the three considered muscles



vs. the knee flexion angle. The curves displayed here were not normalised and the relative ratio among curves may differ for diverse activation levels.

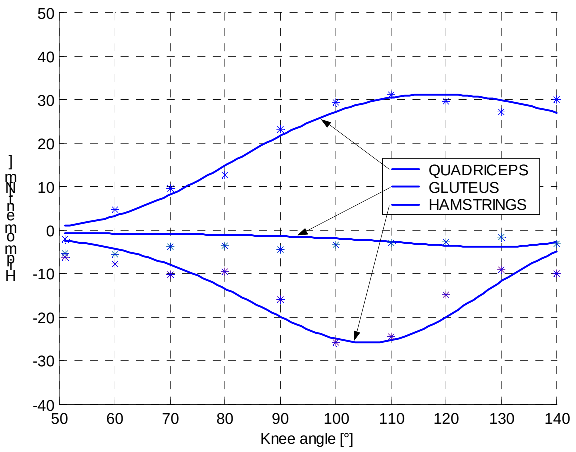


Fig.8 The hip joint moments for the three considered muscles as well vs. the knee flexion angle, because the hip angle was determined for each knee angle. Not normalised curves.

4. Conclusions

The identification procedures presented have outcomes that are broadly comparable, in the overlapped measured range, to those of equivalent procedures with force sensor. Using the set-up proposed here, the parameter identification of different muscles' groups of the leg was possible.

For certain positions, the quadriceps contraction introduce the presence of abduction that it was possible to measure by this procedure. Incorporation of quadriceps and hamstrings co-contractions reduces the knee joint abduction and increases the experiment accuracy reducing the experiments.

Regarding knee joint, it was found that the gluteus contribution was negligible.

When the *in vivo* characterisation of different muscles is the objective, the flexibility afforded to this set-up results in a wide range of parameters identification e.g. moment-angular velocity relationships or stiffness.

Comparing the results to those of the kinetic system, it is clear that leg geometry alone is sufficient for most purposes for predicting moments in an isometric loading regime.

This report was concerned with developing groundwork, not with statistical averages. Moreover, experiments carried out on normal, healthy subjects evidenced small variation of parameter estimates.

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