Medical Engineering & Physics 22 (2000) 97–107

Medical Engineering & Physics

www.elsevier.com/locate/medengphy

Variation of recruitment nonlinearity and dynamic response of ankle plantarflexors

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Received 4 October 1999; received in revised form 7 February 2000; accepted 29 February 2000

Abstract

The ankle plantarflexor muscles of paraplegics may be trained to provide balance without support from the hands (in the laboratory environment) with the controller based on a two-block Hammerstein muscle model. This paper presents data on the variations of the recruitment curve block and linear dynamics block with electrode position, among various individuals and with fatigue. The tests were conducted in six groups: 'a' tests of a neurologically-intact subject were repeated on one day several times to record the effect of muscle fatigue; 'b' the same individual kept electrodes attached for a week and the muscle was identified every day; 'c' the same subject attached electrodes at marked positions every day for a week prior to identification; 'd' another normal attached electrodes at notionally the same positions over a period of one week; 'e' three normals and 'f' two paraplegics. Measurements were made with the Wobbler apparatus, in which the subject is supported upright in a standing posture. When comparing tests of fresh muscle every day, little difference was found between the nonlinear recruitment curves and linear dynamics of groups 'b' and 'c'. In fatigued muscle the dynamics were slower. When the electrode position was not carefully reproduced, and over a longer period, significant differences in nonlinearity appear in the curve shapes (group 'd') and a similar amount of variation occurs between normals, between paraplegics, and from normals to paraplegics. The paraplegic curves show wider deadbands. The effect of prolonged stimulation on normals is slight but on paraplegics it is significant.

Keywords: Electrical stimulation; Muscle; Model identification; Model variation; Paraplegia

1. Introduction and background

Muscle models play an important role in the analysis of motor control and the design of motor system neuroprostheses. Over the years, many models have been proposed to predict muscle force, ranging from simple linear models with constant coefficients [1,2] to higher order nonlinear models which incorporate the effects of a large number of biophysical and mechanical variables [3–7]. The task of modeling requires both the determination of a particular model structure that defines the complexity for the model, and the assignment of the model parameters that relate the model to a specific physical system.

Simple second-order linear models (single input, single output) fit experimental data quite well if the experimental conditions are constrained; for example, for isometric force modulation by variation of stimulus rate [8], for isometric force modulation by varying the fraction of muscle stimulated (recruitment) [9], for force modulation by variation of length during constant stimulation [10]. The underlying nonlinear behavior of muscle is revealed only when the conditions are changed; changes in isometric length, stimulus rate, or amplitude of disturbance require changes in model parameters. The linear models do not have sufficient generality to capture muscle behavior under a wide range of operating conditions.

The muscles possess nonlinear as well as time-varying properties. The nonlinear models of electrically stimu-

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1350-4533/00/$ - see front matter
PII: S1350-4533(00)00016-3

lated muscle have been reviewed by Durfee [3]. By considering two-block models with a static nonlinear element and a linear dynamic element, the nonlinearity might be placed before or after the dynamic block resulting in so-called Hammerstein [11] or Wiener forms [12]. Researchers in the field of functional electrical stimulation have usually adopted the Hammerstein structure [9,13–16]. This is due to the correspondence with biophysics, with nerve fiber recruitment included in the first block, then being followed with muscle contraction dynamics block. The static nonlinearity describes the recruitment modulation which involves varying the number of muscle fibers activated, by varying the amplitude (current or voltage height) or the duration (width) of stimulus pulses. Pulse width modulation is the preferred method for recruitment modulation because of lessened charge transfer per stimulus pulse at any given force [17], which decreases possible electrode corrosion and tissue damage [18]. The modulation by temporal summation (stimulus period modulation, or, inversely, pulse frequency modulation), achieved by varying the time interval between the start of successive pulses, is not used for recruitment purposes [19]. The nonlinear steady-state force versus pulse-width recruitment characteristic of the electrode–muscle system is usually identified simultaneously with the input–output muscle response dynamics, using the Hammerstein-type model [20]. The shape of the nonlinearity depends upon many factors, including electrode placement and location, physiological properties of the muscle, its innervation and electrode properties. The slope for this relationship depends on muscle length and decreases as length increases [19]. In addition, these nonlinearities are time varying, subject to fatigue and other factors [20,21]. The intensity-dependence of the muscle dynamics has been known for a long period of time [22,25]. The intensity-dependent dynamics agrees with Heinneman's principle for normal neuromuscular systems, where smaller motor nerve fibers innervate slow motor units and vice versa [23]. We should also be aware that during excitation with short rectangular stimulation pulse currents, shorter pulses activate first the large nerve fibers, i.e. in nonphysiological order (reverse of their diameters) [24]. As expected, with an increase in stimulus intensity the dynamics became slower [25].

Although controllers based on the two-block representation have been found to perform acceptably, they may fail if large internal disturbances, such as changing recruitment gain, or external disturbances, such as changing loads, are introduced into the system. Since the controller parameters remain fixed in the face of changing system parameters, failure could result from insufficient controller robustness, resulting in poor disturbance rejection or unacceptable reference tracking. The schemes where controller parameters are allowed to adapt to changing plant parameters [26–29] or control based on Hill-type models yield better performance and robustness. Hill-based muscle models attempt to describe muscle behavior under general conditions by incorporating separate terms: activation, length-tension, force-velocity, and series elasticity terms as uncoupled elements [30–34]. In reality the element values depend on operating conditions as well as on mutual coupling; thus leading to the mixed dependence of the three factors and finally to the coupled models. It may easily happen that identification of numerous parameters in such complex models becomes impossible in a reasonable experimental time. The two-block Hammerstein models are, despite numerous unavoidable simplifications, convenient for conditions where the fatigue and identification time are important and in practical real-time implementation due to low computational complexity.

During our experiments of hand free standing in the Wobbler apparatus [25,35–38], the ankle plantaflexor muscles were used for balancing the body. The body was considered as a single inverted pendulum, allowed to move in the sagittal plane and leaned forward from the vertical. Muscles were stimulated via surface electrodes and always identified before the experiment to obtain the parameters in a Hammerstein model. These data were used in moment and position linear controller designs. In an experiment series it was noticed that the numerical values of controller polynomials from the old designs, calculated several days ago, differ less than expected from the new ones. The self-adhesive electrodes were always placed approximately on the same place on the calf. Several questions arise: how often is muscle identification in the Hammerstein muscle model necessary for one person using a closed-loop system with surface electrodes: every day, after the electrode change or only occasionally? How large are the differences among several intact persons and how do paraplegic muscles compare to the intact? In this paper these questions are evaluated with two objectives: (i) to examine recruitment nonlinearity variations among six groups defined later in the text and (ii) to check dynamic and static properties of the linear part of the model.

In the following Methods section we begin describing the identification equipment, identification methods of Hammerstein muscle model parameters with recruitment nonlinearity and linear transfer function, measures of model performance, subjects and test groups. The twitch response method [20] was used for acquisition of the nonlinear recruitment curve. Thereafter, three signals were used to access the linear dynamic block: twitch impulses, excitation at various frequencies and the pseudo-random binary sequence (PRBS) activation signal. The recruitment nonlinearities are presented here in a graphical form, while the properties of the linear part are shown as a gain parameter for various excitation signals and as normalized time-responses with a step excitation function. The pole-zero configurations for all excitation signals were studied and are shown for illustration. Cross validation of linear models between the model obtained with one excitation and validated subsequently with all the other identification responses was also done during the study. For a given object there are only small model parameter variations for nonlinear and linear parts. Noticeable deviations appear among various intact persons and significant differences are apparent when compared to paraplegic muscle responses. As shown earlier by Hunt et al. [25], excitation with PRBS signals gives a better model.

2. Methods

2.1. Identification equipment

We studied the ankle plantarflexor muscles separately for left and right legs. All tests were made in isometric conditions in the Wobbler apparatus [35], which had shoes fixed in two boxes engaged in a stiff horizontal position on a measurement axis. Two torque cells allowed acquisition of left and right ankle moments separately. The person stood in an upward posture while all joints, except the ankle, were made rigid by strapping the person into a special brace. The body was then a single inverted pendulum, which was tightened with eight light ropes from left/right hip and left/right shoulder forward and backward to the ceiling frame, as is shown in Fig. 1. The ropes were slack enough to prevent movements in four directions: left or right in the coronal, and forward or backward movement in the sagittal plane, while simultaneously still allowing the legs to carry the full body weight. Muscle identification in such an upward posture with the legs loaded in one of the most natural positions is a benefit of this arrangement. Body (and shank) are stabilized with the ropes, while feet are fixed in the boxes, making the measurement environment isometric. During the experiments both arms were placed on the chest and treated as a part of the body torso. The moment resolution was 0.1 Nm, measured via a 12-bit A/D converter and with custom software on a PC computer, which also carried out stimulator control.

2.2. Identification methods

Electrodes were Axelgard, 50 mm diameter,[1] placed on the midlines of the soleus and gastrocnemious muscles. Two channels with monophasic, charge balanced constant current pulses from the Stanmore stimulator [39] were used for left and right plantarflexor stimulation. One side was stimulated at a time. Four PC

[1] Axelgard Manufacturing Company, Fallbrook, CA.

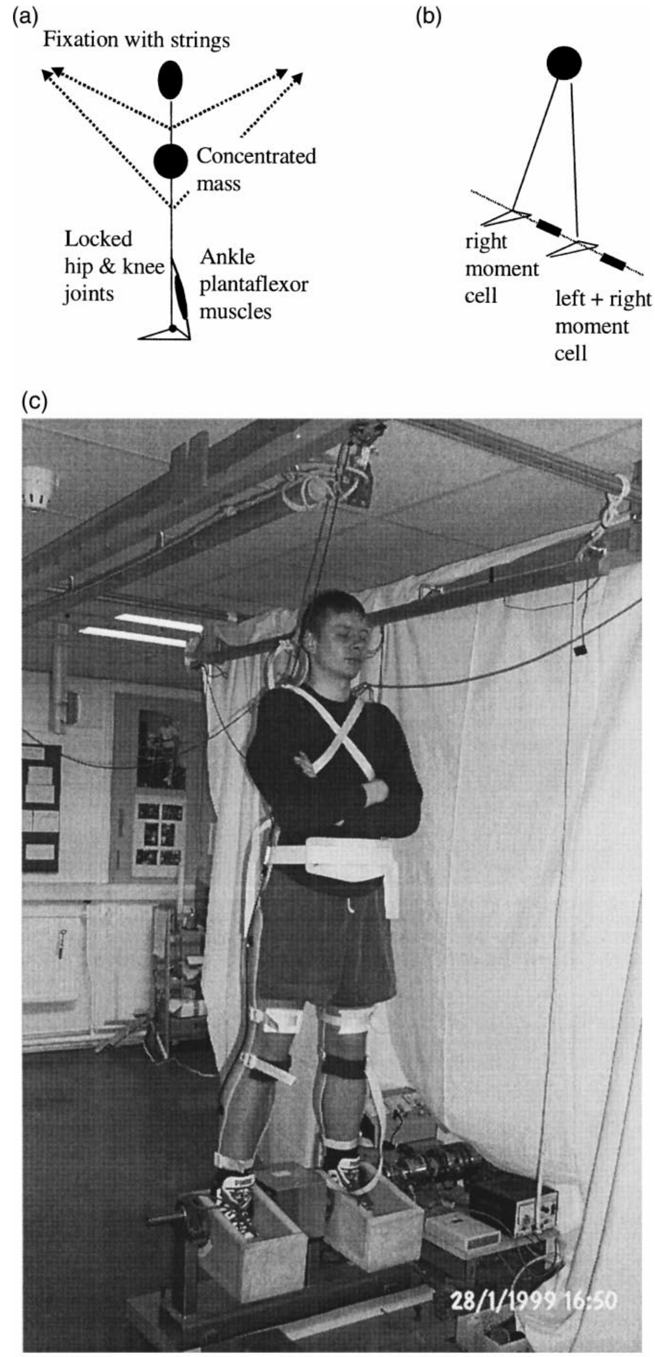

Fig. 1. (a) Ankle plantarflexor tests, body as a single inverted pendulum; (side) sagittal plane view in (b) of Wobbler apparatus. Ropes were not strained exactly forward/backward, they were inclined slightly outward to also maintain stability in the coronal plane. (c) A subject in the Wobbler apparatus during the test.

programs used during identification were labelled as Test A, Test B, Test G and Test E [35].

- **Test A** stands for stimulation pulse amplitude adjustment. Pulse width was varied from 0 to 500 μs in 2 μs steps as a 3 s ramp and 20 Hz frequency with amplitude fixed at 10 mA marks, while the moment response was acquired. According to the response and the subject's sensation, at key hit the procedure repeated at 10 mA increased amplitude. The current amplitude was fixed as maximal according to: (i) subject sensation, tolerance to stimulation, (ii) measured moment value. Experience showed that 50 Nm to 75 Nm isometric moment in one leg, continuous stimulation, could always be achieved with comfortable sensation. As can be verified later in Fig. 6, this procedure can not guarantee total excitation of muscle, but brings response to the upper, saturated region of recruitment curve. Maximal current value was recorded and then fixed for subsequent tests. Pulse-width 0 to 500 μs corresponds to muscle activation from 0 to 1000 mAct (we defined 1000 mAct = 1 Act = 100%, full activation).

- **Test B** is twitch response excitation. Five sets of ten pulses [50, 100, …, 500] μs, altogether 50, were randomized and each one delivered every 1.2 s. Moment values were sampled at 200 Hz. The recruitment nonlinearity is comprised of ten points, each representing the peak of an average of five moment responses at the observed pulse-width (Fig. 2). These data were also used for linear block identification, usually at the middle 250 μs pulse duration.
- **Test G** is sinusoidal stimulation modulation at predetermined frequencies and with 200 Hz moment sampling. Muscle activation is varied for three cycles at [0.3, 0.5, 0.8, 1.2, 2, 3.2] Hz with mean amplitude 500 ±400 mAct; all cycles together lasted for 28 s. During Test G the sinusoidal input was first passed through the inverse recruitment nonlinearity and then to the stimulator at 20 Hz. The inverse recruitment curve was calculated from piecewise linear approximation between the 10 data points taken via the twitch method Test B. The frequency response was used to identify the linear transfer function of muscle (Fig. 3a).
- **Test E** is excitation with a PRBS sequence. The PRBS signal is read from a file and dispatched to the stimulator at 20 Hz. The moment response is acquired at 20 Hz sampling frequency and used for identifying the linear transfer function of muscle (Fig. 3b).

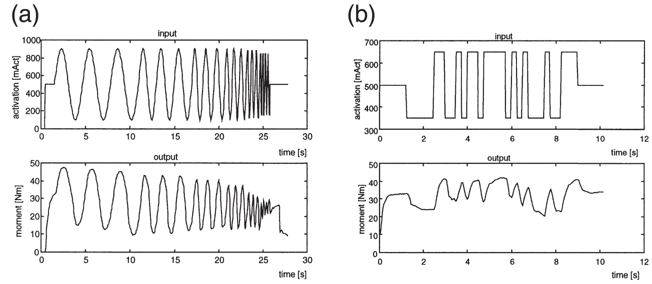

Fig. 3. Excitation signals and typical response for Test G (a) and Test E (b).

2.3. Hammerstein form muscle model

In Hammerstein form (Fig. 4) the muscle consists of the static recruitment nonlinearity f_r, in Fig. 2, followed by a linear transfer function; in this case assumed to be the ARX (auto regressive with exogenous input) model type:

$$A_m(q^{-1})m(t)=q^{-k}B_m(q^{-1})a(t)+d(t) \tag{1}$$

The variables in the muscle model are: $p(t)$, stimulation pulse width (at constant amplitude); $a(t)$, muscle activation level; $d'(t)$, disturbance signal ($d(t)$=zero-mean white noise); and $m(t)$, muscle moment.

The integer $k \geq 1$ is a discrete input–output transport time delay, which was found by Bawa and Stein [40] to be 15 ms to 20 ms for soleus muscle and confirmed as less than 50 ms for all plantarflexors [25]. At 20 Hz sampling frequency a good linear model structure was found to be $(na, nb, k) = (2, 1, 1)$ [25], with polynomials A_m and B_m, and q^{-1} as delay operator:

$$A_m(q^{-1})=1+\alpha_1 q^{-1}+\ldots+\alpha_{na}q^{-na}=1+\alpha_1 q^{-1}+\alpha_2 q^{-2} \tag{2}$$

$$B_m(q^{-1})=b_0+b_1 q^{-1}+\ldots+b_{nb}q^{-nb}=b_0+b_1 q^{-1} \tag{3}$$

$$\frac{q^{-k}B_m(q^{-1})}{A_m(q^{-1})}=\frac{q^{-1}(b_0+b_1 q^{-1})}{1+\alpha_1 q^{-1}+\alpha_2 q^{-2}} \tag{4}$$

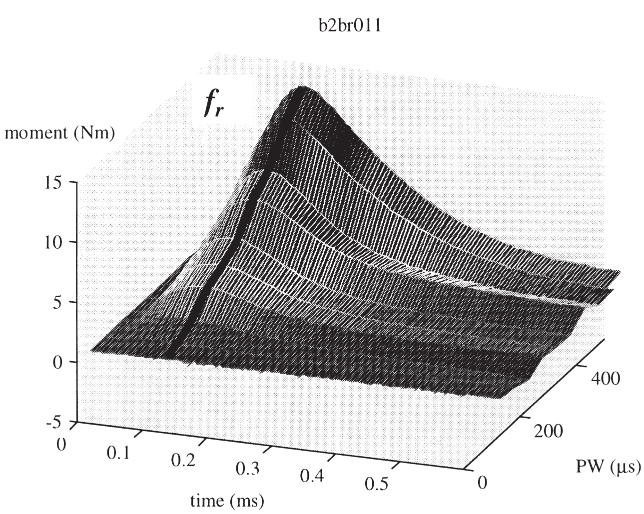

Fig. 2. Average of twitch response data for fresh intact muscle.

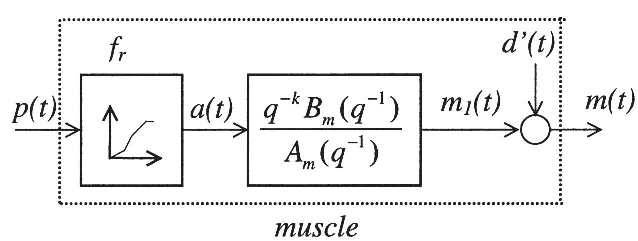

Fig. 4. Hammerstein model of muscle, $d'(t)=d(t)/A_m(q^{-1})$.

$$m(t)=\frac{q^{-1}(b_0+b_1q^{-1})}{1+\alpha_1q^{-1}+\alpha_2q^{-2}}a(t)+\frac{1}{1+\alpha_1q^{-1}+\alpha_2q^{-2}}d(t) \qquad (5)$$

In a previous study [25] we showed that there is some improvement in performance measure for a model having (na, nb, k) = (3, 2, 1) and one step ahead measure, while error increased for $na=4$ and ∞ step ahead predictions. The second order model was thus used from there on. The analytical solution for the optimal (α_1, α_2, b_0, b_1) parameter estimate [41] was found by using procedures from the System Identification Toolbox in the Matlab program package. A measure of model fidelity is provided by the least-squares, n-step ahead criterion (with $n \geq 1$).

2.4. Recruitment nonlinearity

The recruitment nonlinearity is acquired with muscle twitch responses from Test B. The peak value of the average of five same-width pulses gave one point on the recruitment curve f_r and, altogether, the recruitment nonlinearity f_r. The mesh ridge in Fig. 2 represents the f_r curve for an intact person. The mesh shape for fresh paraplegic muscle is similar, while Fig. 5 shows for comparison the worst observed case for fatigued paraplegic muscle.

2.5. Transfer function estimation

Parametric identification methods, described in detail in Hunt et al. [25] and briefly above in the section on *Hammerstein form muscle model*, are used to determine the parameters of the linear transfer function for all three excitation signals in Tests B, G and E, thus generating three model descriptions of the same system. The preceding Methods section gives insight into some specialities associated to each particular excitation.

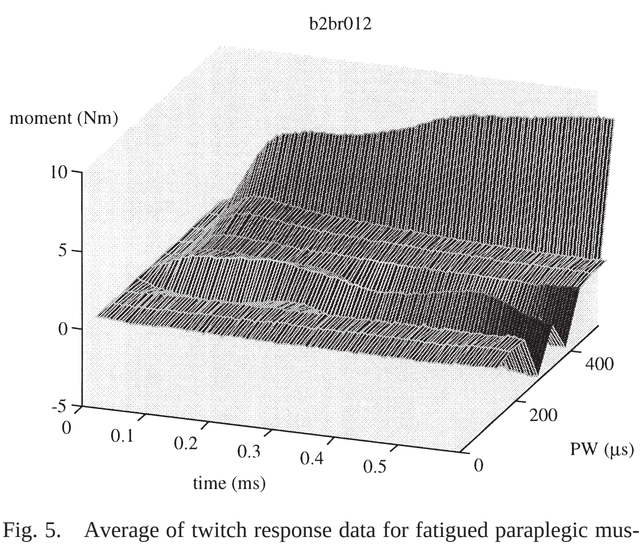

Fig. 5. Average of twitch response data for fatigued paraplegic muscle for the same case as the bold solid trace in Fig. 6f. The curve in the back (500 μs) does not go to zero as T increases due to slow muscle response. Consult Fig. 8f for the step response dynamics of the paraplegic muscle.

- *The twitch response method* (Test B) is similar to that used by Durfee and Maclean [20]. It causes the least fatigue among all three methods applied here, and is necessary to obtain the first, recruitment nonlinearity block in Hammerstein form. Using the early part of the moment response again for linear model (dynamics) estimation does not require any additional experimentation. This was the main reason for considering twitch response models for comparison with the other two models in this study. However, as was pointed out earlier [22,25], the muscle gain and dynamics vary with excitation level. The dynamic aspect should not be forgotten when estimating parameters in the linear block from Test B, which is truly the identification of a local model from zero activation.
- The *PRBS activation sequence* in Test E, designed to give the best excitation of the muscle, can produce frequency rich output response, and proved to be the preferred type of muscle activation [25]. The muscle was excited here in the mid-range of activation 500 ±150 mAct. Due to the deviations of ±150 mAct, too large to be neglected, the inverse recruitment nonlinearity obtained previously in Test B was used before sending pulse-width values to the stimulator to cancel the internal muscle nonlinearity. In this way the muscle was excited in the mid-range, but due to the deviations of ±150 mAct identified locally in that region.
- Test G, using *sinusoidal excitation*, could also in principle be used for model parameter estimation. By carefully selecting excitation frequencies the amplitude and phase of the response gave reasonably good insight into the system's behaviour. Even with the longest identification measurement of the open loop muscle system, with 28 s of time at 20 Hz, the sinusoidal excitation is still used for comparison. The excitation amplitude 500 ±peak 400 mAct was first processed with the inverse recruitment nonlinearity (similarly as before in Test E). Due to the large peak of 400 mAct deviation the muscle behaviour is identified more globally.

2.6. Measures of model performance

For the model parameter estimation two sets of data are usually acquired with the same excitation signal applied to the same system: one for parameter estimation purposes and the second for model validation. Two equal sets of identification signals were also applied to the muscles in this study, as described later. Two unequal excitations were applied for more rigorous validation on a smaller data sample in Hunt et al. [25], giving insight into model (methodology) fidelity and the prediction errors. After obtaining the analytical solution for the optimal parameter estimate for the present data, the model output prediction errors were calculated, as well as errors when validated with the second set of data and all other measured responses (all tests in all groups). To assess model accuracy, a least-squares error criterion with various prediction horizons was used.

Based on the model parameters, the fidelity was examined further by pole/zero positions, covariance matrix with standard deviation of estimated parameters, transfer function gain and step transient response. The standard deviation of the estimated parameters was acquired as the square root of the diagonal elements in the covariance matrix, which was calculated during ARX model identification. The gain was calculated as gain=$B_m(1)/A_m(1)$ in the estimated model.

2.7. Subjects

Five subjects participated in this study. Three neurologically intact subjects were 28, 34 and 43 years old and labelled here as M, N and I. Two paraplegics, abbreviated as S and T, had T5 and T12 lesions, and at the time of the experiments were aged 35 and 42 years, 13 and 9 years after the injury. Prior to the measurements the paraplegic ankle plantarflexor muscles were strengthened with a daily training protocol with surface electrical stimulation using the Stanmore stimulator [39]. Exercise was performed for the ankle plantarflexor and dorsiflexor muscles using bilateral reciprocal stimulation in a sitting position for 30 min daily.

Before the measurement trial the subjects were informed about the nature and timing of the experiment. They were asked to look forward, keep the body relaxed, not to move during the stimulation periods and not to interfere voluntarily with muscle stimulation. Moment traces were later manually checked to assure that no voluntary introduced offset moments or reflex activity were associated. There was no need for any corrections, except in one paraplegic person (not included here), where the plantaflexor stimulation provoked an escalation of spasticity.

2.8. Test groups

The two-block constitution of the Hammerstein model suggests presentation of findings in two parts: (i) first a separate check of nonlinear recruitment curve variations and then (ii) examination of linear transfer function properties. To follow these two points of interest in combination with a set of questions presented at the end of the Introduction section, measurements with intact and paraplegic volunteers were combined in six groups:

Group ‘a’ One intact person (M) was tested in one session, consisting of three identical identification measurement sets. The session started with Test A (left/right side), and was followed by applying identification excitations: twitch response method (Test B, left/right), sinusoidal method (Test G, left/right) and PRBS excitation (Test E, left/right). After 10 min the identification set was repeated and after 10 min once again. The order of the tests was thus: Test A, B, G, PRBS, pause, B, G, PRBS, pause, B, G and PRBS. This protocol was designed to give insight into parameter changes due to *muscle fatigue*.

Group ‘b’ Intact person M, *kept electrodes on the skin* without taking them off for five days with the same identification procedure repeated twice, one after another, every day (Test A, B, G, E, B, G, E). The aim of the measurements in this group was to have insight into day-to-day variance of the muscle model.

Group ‘c’ Similar to group ‘b’, i.e. intact person M, but the same *electrodes were applied every day on the same marked area on the calf*. Again the same identification excitation was used including Test A, B, G, E, B, G and E. In a real application this group mimics the situation where the user applies electrodes regularly every day or perhaps occasionally, but as accurately as possible at the same location.

Group ‘d’ Intact person N, the same set of electrodes was applied for four days *approximately on the same*, but now unmarked, area. Identification excitation signals were as before: Test A, B, G, E, B, G and E. This measurement group mimics the situation of a regular or periodic user, who possesses knowledge of where on the calf to place electrodes properly, but the exact position is not marked.

Group ‘e’ Here parallel models are acquired *among several intact persons* (subjects M, N, I). The same excitation signals were applied in all subjects including Test A, B, G, E, B, G and E, as in previous groups. In this group we compared person-to-person model variations and not the individual properties.

Group ‘f’ Comparison *between two paraplegic persons* (persons S, T). Excitation signals were as in the above groups including Test A, B, G, E, B, G and E. Identification of paraplegic muscles is treated separately for comparison reasons to the intact group ‘e’. More details of this group are also given in the following Results section.

3. Results

By considering all the tests done on the left and right legs, 76 identifications were carried out in all persons with the twitch response test (Test B), 65 with sinusoidal excitation (Test G) and 63 with PRBS signal excitation (Test E). The number is not equal among the three for various reasons, including poor data collection originating from the computer or Wobbler apparatus, or due to subjects' unwanted (unintentional voluntary disturbance) movements during the trials. If encountered, such events were noted during the experiment and the data later examined manually. Some trials were left out from further calculations. Considering the similarity of left/right data, we will in this section focus on results representing only right ankle plantarflexor muscles. The only difference between left/right side and the second reason for observing right side from here on, is that there was practically no useful response in paraplegic person B with the left leg.

The presentation of results consists of two parts: for nonlinear recruitment and then for the linear dynamic block. In Figs. 6a–f recruitment curves for each test group 'a' to 'f' are merged. Only recruitment curves for the first identifications on a subject on a particular day are shown. Results for the second attempt are not shown here for clarity, but fall within the range presented in group 'a', which demonstrates fatigue effect. An exception to this is paraplegic group 'f' (Fig. 6f), where all first identifications with paraplegic S and two measurement sets for paraplegic T are shown. This gives insight into day-to-day variations and the effect of fatigue in paraplegics. Thus, seven separate sessions on seven days are shown for paraplegics. The labeling in Figs. 6a–f has the following system: take as an example Fig. 6f, code TR012 70, the first letter represents subject T, the second, R, addresses the right side, 01 is the session count, and 2 is for the second measurement set in that session; the number 70 after that corresponds to 70 mA current amplitude during stimulation being adjusted in Test A and further used in other tests. All recruitment curves are normalized, meaning that response measured at 500 μs always equals 1000 mAct. The gain information is given further in Fig. 9.

For the estimated linear dynamic block model, normalized step transient responses for the models based on twitch response (Test B) are shown in Figs. 7a–f, and those for the models based on PRBS excitation (Test E) are shown in Figs. 8a–f. To further characterize the estimated linear block, the transfer function gain values are shown in Figs. 9a–f for test groups 'a' to 'f'. Gain figures are for comparison between Test B and E excitations, also including the gain values for Test G. Considering the large prediction errors found for sinusoidal excitations in a previous study [25], Test G was only used for gain comparison. Again, identification results from the first trials are presented.

Many observations regarding the Hammerstein muscle model in Hunt et al. [25] were verified and confirmed on a larger data set here. To demonstrate pole/zero position and movement, Fig. 10a gives pole/zero configurations for twitch-based models and Fig. 10b for PRBS-based models.

4. Discussion

4.1. Nonlinear recruitment curve

The nonlinear recruitment curves calculated from twitch response data are shown in Figs. 6a–f for groups 'a' to 'f'.

- In group '**a**', where measurements were done at 10 min intervals, the differences are mainly between the first curve and the other two. The activation level (gain) is approximately 150 mAct units higher in the first response compared to the other two curves for the majority of pulse widths. The shifts toward lower muscle gain are expected and confirm that the muscles are fatiguing.
- An unexpectedly good agreement between curves was measured in group '**b**' and also group '**c**' tests, suggesting that nonlinear recruitment does not change if surface electrodes are kept on the skin for several days or if electrodes are applied on the same marked area repeatedly. Please note also that in group 'b', 50 mA pulses were used in two trials and 60 mA in the other three, which does not noticeably influence the recruitment nonlinearities.
- Larger discrepancies appear inside group '**d**' where the electrodes were placed only on the same unmarked area every day. Even if 70 mA pulses were used in some trials, in this group 'd', no saturation effects are visible. All curves are more spread out than in groups 'b' and 'c' and they also differ altogether from groups 'b' and 'c' above. The gain fatiguing effect is evident from two subsequent trials shown in the figure.
- For group '**e**', trials were randomly selected from several tests in the same subject. This immediately showed up variations between persons, larger than encountered before in groups 'b', 'c' and 'd'. The middle curve is for 50 mA current pulse, left for 60 mA and right for 70 mA. Therefore, no direct relation between current value and recruitment curve shape is evident, or it is at least much smaller than are variations among subjects.
- In group '**f**', five out of seven traces were recorded from one paraplegic individual, when not tired (SR091 to SR0D1). The other two were acquired from

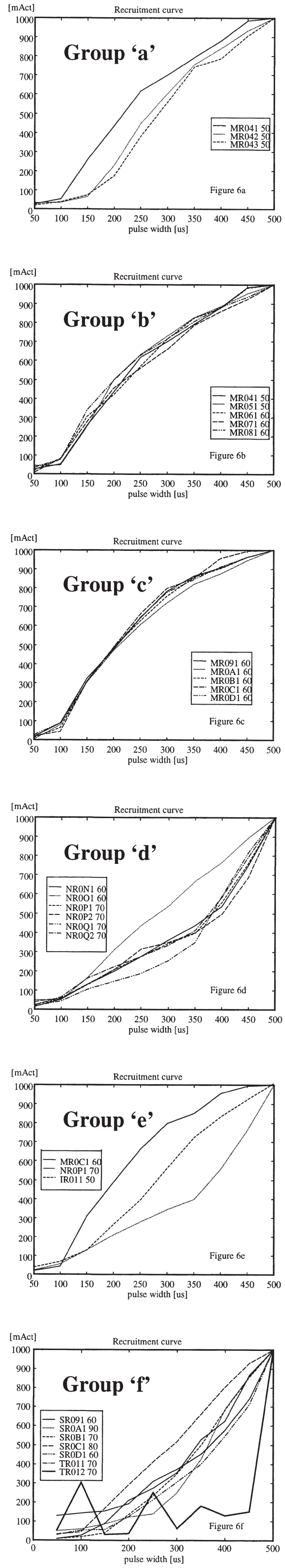

Figure 6

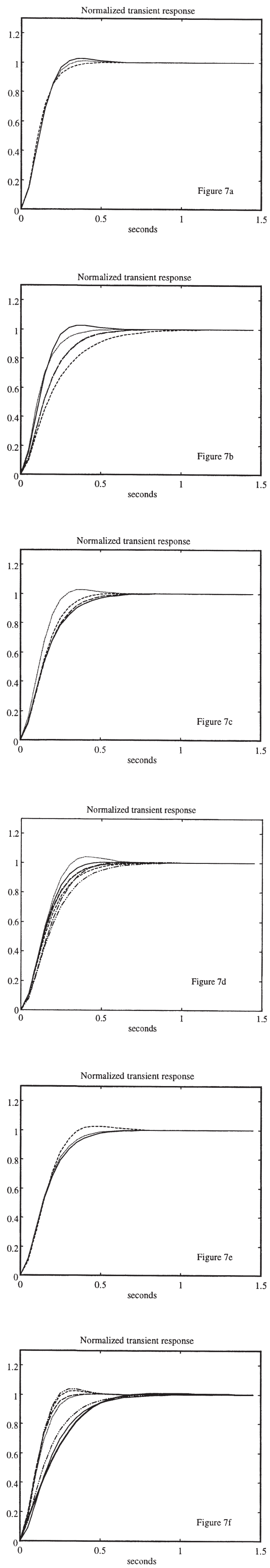

Figure 7

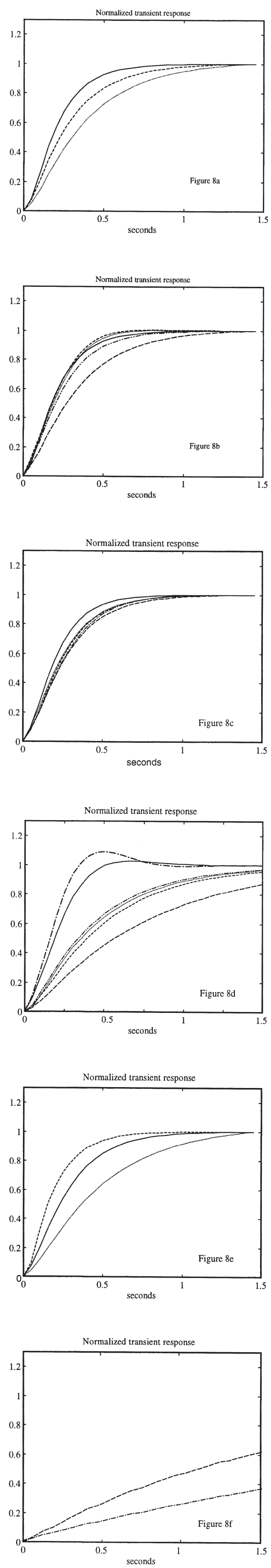

Figure 8

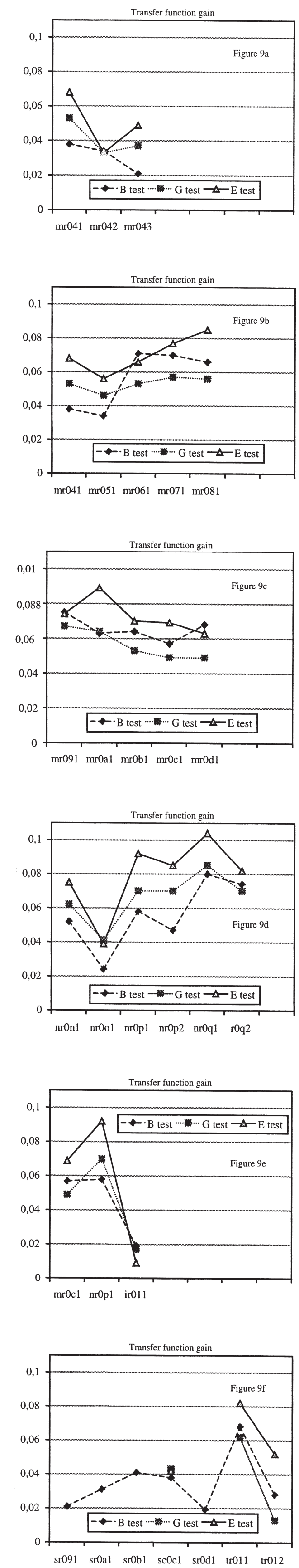

Figure 9

the second paraplegic person in one session. Total muscle fatigue in the second set is evident. We may also note 90 and 80 mA currents resulting in curve shapes being comparable to intact subjects at a lower current value, and rapid fatigue in the last trial.

4.2. Linear transfer functions

Normalized transient responses will be discussed first. Figs. 7 and 8 allow a comparison to be made between the linear models obtained through identification in B tests, and between PRBS identification models.

- The time scale for all the graphs in Fig. 7 Fig. 8 is 1.5 s, including Fig. 8d (several intact subjects) and Fig. 8f (paraplegics). In general, *transient responses* calculated from twitches are faster than the others in all groups. The slower dynamics observed with PRBS excitation is, due to initially higher average muscle excitation, valid for all groups and supports earlier findings [25], where the measured response of muscle was found to be faster at low activation and slower toward saturation levels. The initial conditions in twitch responses were zero, and 500 mAct in PRBS signals. The PRBS input signal amplitude, with a mean value of 500 mAct in this case, due to summation effect excites muscle much more than single excitation pulses in the twitch method. At higher excitation, smaller motor nerve fibres innervating slow motor units [23] start to contribute in (slower) response. It is normal in a nonlinear dynamic system that identification from zero initial point, or from, e.g. 50% (500 mAct) initial point, gives two different models.
  - A detailed comparison for group ‘a’, for twitch data (Fig. 7a), shows similar transient responses as fatigue increases, and also shows similar recruitment curves for all three cases. On the other hand, the dynamics differ [25] in Fig. 8a (PRBS model) as one might expect from the shift of static recruitment in the middle activation region.
  - Groups ‘b’ and ‘c’ can be checked together. Figs. 7b and 8b, 7c and 8c could not confirm the hypothesis that electrodes after several days of use, or any other parameters observed, change muscle response dynamics. The bundle of curves is even less dispersed than traces in Fig. 8a, which represent deviations due to fatigue in one session. The response is very consistent.
  - In group ‘d’ (Figs. 7d and 8d), the model identified from twitch responses is similar to those in previous groups, but very different from the PRBS-identified models, which, for most cases, displays slower dynamics (slower rising) than other trials. Further visual inspection of Fig. 8d (compared to Figs. 8a–c) shows evidently larger curve dispersion.
  - Fig. 8e shows that the dispersion (in dynamics) among various persons is equal or within to that observed in one person with the electrodes attached without marks.
  - Much slower dynamics (slow rising) than found in any other group is shown in both traces of Fig. 8f. If in the virtual experiment, the time scale of Fig. 8f was squeezed five to ten times, normal dynamics would be obtained (in Fig. 8f, at 1.5 s, 40–60% is obtained; e.g. in Fig. 8d, 40–60% is obtained between 0.15 and 0.3 s, meaning a ratio of 10 to 5). Due to regular training, subject S had better-conditioned muscles than subject T. As shown in Fig. 8f, subject S demonstrates faster response (dashed line), higher gain (Fig. 9f) and more moment capability.
- *Gain values* for all identified models are plotted in Fig. 9 for all groups. As a general rule, gains found by PRBS tests were among the highest. This recognition, supported by numerous identification trials, confirms very well the gain increase with activation increase depicted in earlier work [25] where a smaller data sample was used. Fig. 9a for group ‘a’ shows the trend of lowering model gain from MR041 to MR042 and MR043 tests due to muscle fatigue. Fig. 9b indicates increased gain for the last three tests, however, this should be observed carefully and attributed to 60 mA pulse current in these trials. As we might anticipate, the gains from Test B are equal for all the trials in group ‘c’ (Fig. 9c). A fairly constant gain is measured for all PRBS tests. Greater gain variations are shown in Figs. 9d–e for groups ‘d’ and ‘e’. In general, low gains were found for group ‘f’ (paraplegics). The last two measurements, BR011 and BR012, demonstrate rapid muscle fatigue, produced only by muscle stimulation during identification.
- Considering the *qualitative properties* of muscles observed through pole/zero positions, the clustering of model poles is evident from Figs. 10a, b. Clusters in Fig. 10a move toward the real axis in Fig. 10b, leading to an over-damped response.

Fig. 6. Recruitment nonlinearities for groups ‘a’ to ‘f’.

Fig. 7. Normalized step transient responses for the model based on the twitch response for groups ‘a’ to ‘f’.

Fig. 8. Normalized step transient responses for the model based on the PRBS response for groups ‘a’ to ‘f’.

Fig. 9. Transfer function gain values for groups ‘a’ to ‘f’.

5. Conclusions

Isometric muscle behaviour is often described by the Hammerstein model, which consists of two blocks: the

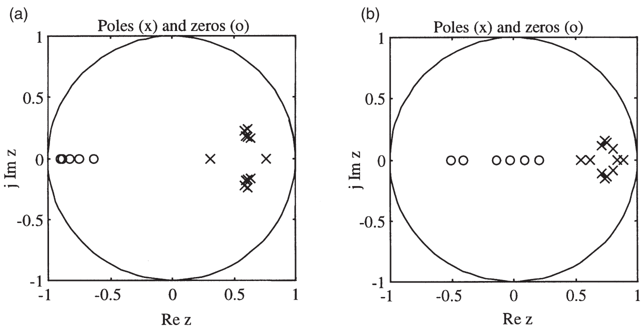

Fig. 10. Pole/zero configuration for twitch-based models (a) and PRBS-based models (b). Both for group 'd'. Subject N was sticking electrodes on his calves every day in approximately the same area.

nonlinear recruitment curve followed by a linear transfer function. In this paper we present data on the variations of the recruitment curve and muscle dynamics with electrode position, among individuals and with fatigue. The measurements were clustered in six groups: 'a' to 'f' as described above. Subjects were three normals and two paraplegics. The nonlinearity was measured by Durfee's method, from the twitch responses to single stimulation pulses in the range 50–500 μs, at 10 levels, having previously set the stimulator current. Five sets of pulse widths were applied in random order for better accuracy. The following conclusions can be drawn for the nonlinear recruitment curve:

- A shift of *recruitment nonlinearity* toward lower muscle gain as a result of fatigue was confirmed.
- In the same intact person the recruitment curve changes little if surface electrodes are kept on the skin for several days.
- When the electrode positions are not carefully reproduced, and over a longer period, significant differences appear in the curve shapes and a similar amount of variation occurs between normals, between paraplegics, and from normals to paraplegics. As expected, variation in recruitment among various persons is larger than in a single person, even if electrodes in that person are stuck on a non-marked area. Variations correspond to fatigue changes.
- Recruitment curves in paraplegics show significant deviations from intact muscles and very large variations with time due to fatigue. The paraplegic curves show wider deadbands. The effect of prolonged stimulation on normals is slight but on paraplegics it is great.

Conclusions for the *linear transfer functions* are:

- Transient response in the same intact person, even if applying electrodes every day, is very similar.
- Transient response variation between different people is similar to the case of electrode placement without any marks on the skin.
- The dynamics in paraplegic subjects are five to ten times slower than in an intact person and vary noticeably with muscle condition (i.e. regular training).
- Models identified with PRBS signals, acquired at higher muscle activation, show on average higher gain than other models.
- Lower muscle gain for fresh muscle at the beginning of measurements together with rapid fatigue were confirmed in paraplegics.

For repeatable responses, the electrodes should be placed carefully in the same position. This probably avoids the need to re-identify the recruitment curve at each session. The recruitment curve shapes vary considerably between individuals (normal and paraplegic), and perhaps with electrode position, so we cannot expect that one recruitment curve will give satisfactory results for all. What is most striking, however, is that the variation due to fatigue in paraplegics is much greater.

At the time of the Wobbler standing experiments [36–38] we were not aware of the degree of muscle fatiguing and change in properties that were reported in this paper (due to fatigue within one session). Regardless of the fact that the muscle model varies as is shown in group 'a', the simple robust linear controllers were made functional. While trying to examine the fidelity of the Hammerstein model [25], the original two-block record proved to lack gain and dynamic scheduling. Joining local models into a single model with interpolation [42], or controllers with controller gains and dynamics progressively scheduled [43] seems to be a logical alternative. Joined local models or scheduling would still be affected by the model variations presented here (i.e. fatigue resulting in gain and dynamic change), but with no doubt to a lesser degree than with simple linear control. A possible alternative seems to be nonlinear control [44]. The optimal alternative, having in mind realization in real-time, seems to be online identification of muscle properties [29] with adaptive scheduling of controller parameters [26]. Even this would not make the system work after a certain level of fatigue, where no feedback is going to help anyway. However, the changes in muscle gain and dynamics presented here could be compensated in this way up to that point.

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