Model-based Dynamic Control Allocation in a Hybrid Neuroprosthesis: Supplemental

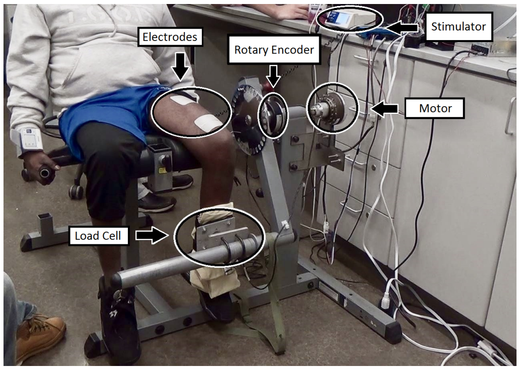


Figure 1: The modified hybrid neuroprosthesis system that combines FES quadriceps muscle with an electric motor.

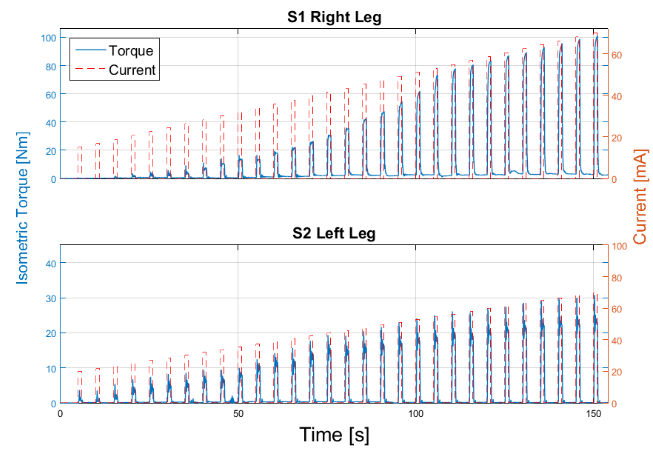


Figure 2: The results from parameter Test 1 for both participants. Stimulation current amplitude ramp to determine the threshold current amplitude, I_t, and saturation current amplitude, I_s of the participant. The threshold is the current amplitude that causes the first significant torque measurement, and the saturation is the current amplitude that produced the last significant change in the torque measurement.

I. Parameter Estimation Procedure

To implement the NMPC-based DCA method, the parameters of the dynamics in (7) must be estimated. This section briefly describes methods, which were also used in our previous work [27], for estimating these parameters based on the methods in [41-43]. The procedure for estimating the fatigue dynamics of the quadriceps muscles are based on methods in [36]. For all parameter estimation procedures, a biphasic $35Hz$ pulse train with a pulse width of $400\mu s$ was used as the wave form parameters of the pulse train. Six tests were performed in the modified leg extension machine to identify the musculoskeletal parameters. During the first five tests it was assumed that the duration of stimulation was short enough and sufficient time to rest were provided, so that muscle fatigue was assumed not to occur (this can be verified from Fig. 8), i.e. $\mu = 1$ during these tests. Results from each test will be shown for the able-bodied participant to exemplify each test of the procedure. Compared to our previous work [27], parameter estimation for the participant with SCI is shown in this paper.

Test 1. With the participant seated in an isometric contraction configuration, 2s long pulse trains of stimulation were administered, and the isometric joint torque was measured using a load cell. The purpose of this test is to estimate the saturation and threshold current amplitudes, I_s and I_t respectively. The stimulation amplitude of the 2s long pulse trains was slowly increased using 20 evenly spaced stimulation amplitudes within $20 - 70mA$. The peak current amplitude was set to $70mA$ based on our experience. In previous experiments we found that most able-bodied participants find the stimulation amplitudes above 70mA to be uncomfortable. We also chose to use this value as the upper limit for the stimulation amplitude for the SCI participant so that the results between able-bodied and SCI can be easily compared. To ensure that the muscles are not fatigued during this test, $4s$ long breaks are provided in between each pulse train. The results of this test were used to estimate the saturation and threshold current amplitudes (I_s and I_t in (5)). The threshold was computed as the current amplitude that produced the first significant contraction, and the saturation was computed as the current amplitude that produced the last significant increase in the measured joint torque. The results of this procedure both participants are shown in Fig. 2.

Test 2. By holding the leg of the participant at different joint angles in the leg extension machine, the passive joint torque and gravitational torque can be measured by using the load cell. The results of these measurements can be used to estimate the passive stiffness (d_i for $i = [1, 3, 4, 5, 6]$ and ϕ_0 in (2)) and mass parameters (m and l_c in (7)) of the participant. Each data point was measured twice and the average value was taken. A nonlinear, least-squares curve fitting algorithm was used to determine parameters that resulted in a best fit between the average data and the function of the passive knee torque, τ_p. The measured data and the resulting best fit determined by the nonlinear, least-squares curve fitting algorithm are shown in Fig. 3 for both participants. The muscles were not stimulated during this test.

Test 3. Isometric contraction torque was obtained by stimulating at the saturation level (from Test 1) for 2s at a number of different joint angles. The purpose of this test is to collect data that may be used to determine the torque-length

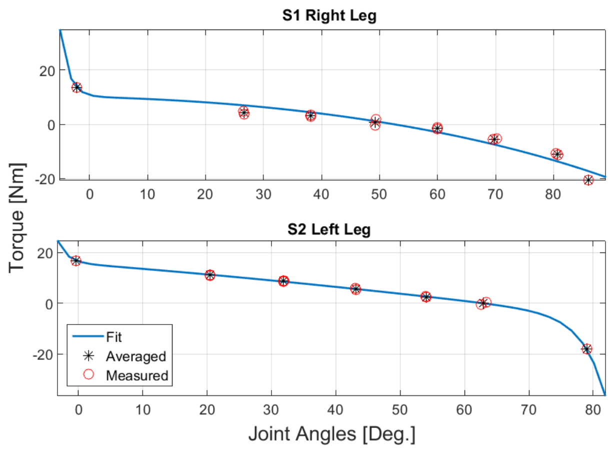


Figure 3: The results from parameter Test 2 for both participants. The exponential terms that model hyperextension and hyperflexion of the anatomical joint angles (ϕ) can be observed around 0º and 85º, respectively.

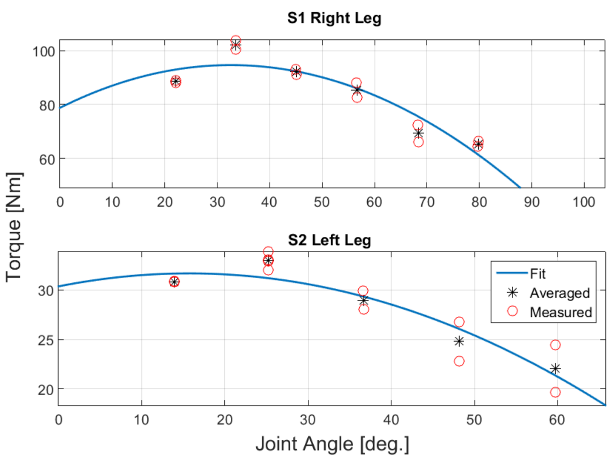


Figure 4: The results from parameter Test 3 for both participants. Torques produced during isometric contraction tests at different anatomical joint angles (ϕ), and the best fit of the measured data to find the torque-angle characteristics given in (3).

parameters (c_i for $i = [0-2]$ in (3)). Like Test 2, a nonlinear, least squares curve fitting algorithm was used to determine torque-length parameters that resulted in a best fit between the measured torque and joint angle at the different positions. The isometric contraction torques were measured at 7 different joint positions, and the best fit to the measured data are shown in Fig. 4. For S2, steady state of the force measurement was used instead of the maximum.

One of the isometric contraction tests from Test 3 was used to determine the muscle activation time constant (T_a in (4)). Because the leg is fixed in an isometric configuration and the muscle was stimulated at the saturation level, which corresponds to a normalized stimulation of 1, the normalized joint torque is equivalent to the normalized muscle activation [27]. The normalized joint torque was measured using the load cell for a step input of stimulation at the saturation

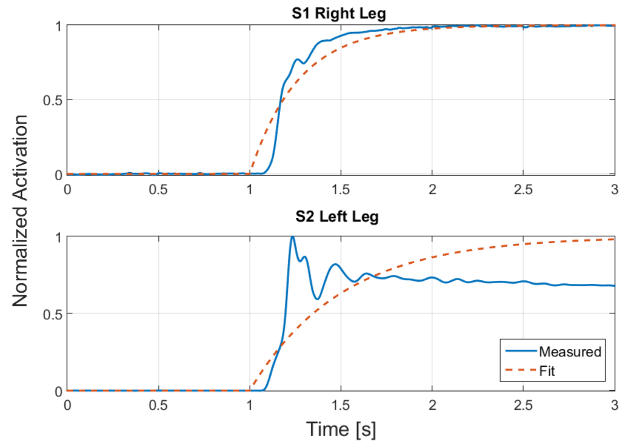


Figure 5: Muscle activation time constant results from parameter Test 3 for both participants. The normalized load cell data were used to determine the muscle activation time constant. The first order system time constant was found by a response that best fits the normalized measured data.

level. Since the normalized joint torque measured by the load cell is equivalent to the normalized muscle activation under these conditions, the normalized load cell measurement was used as an approximate measurement of the first order muscle activation dynamics. The normalized muscle activation from an isometric contraction is shown in Fig. 5, where the stimulation begins at 1s.

Test 4. Pendulum tests were run for determining the damping and inertial parameters of the system (d_2 in (2) and J in (7)). This was done by holding the leg at approximately full extension, then releasing it and letting it drop freely. An optical encoder mounted on the leg extension machine at the knee joint (Fig. 1) was used to measure the response of the leg. Then an optimization was used to determine the best fit damping and inertial parameters. The measured encoder data from the pendulum test and the response from the best fit model are shown together in Fig. 6. The discrepancy between the measured data and fit was also contributed by the previously determined parameters in Test 2.

Test 5. The purpose of this test is to estimate the force-velocity parameter, c_3 in (3). Movement of the knee joints was elicited by applying a sinusoidal stimulation, with a period of 8s, to the quadriceps muscles of the participant. This was measured by using the encoder mounted on the leg extension machine. The amplitude of the stimulation was selected such that for each participant the joint angle was between 10º and 70º. This ensured the muscles to be always in tension and sufficiently far from hyperextension/hyperflexion. The parameters estimated in Tests 1-5 were used to populate the model of knee extension, and then an optimization was performed to identify the force-velocity parameter (c_3 in (3)) that makes the model best match the measured data. Fig. 7 compares the measured knee joint angle to the knee joint angle of the model when given the same input. The RMS error of output of the model that matches the measured joint angle are

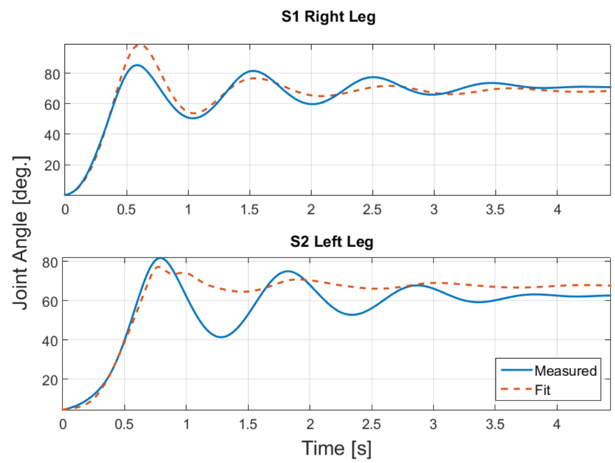


Figure 6: The results from parameter Test 4 for both participants. Plot of the results of the pendulum test with the response of the model that best fits the measured response.

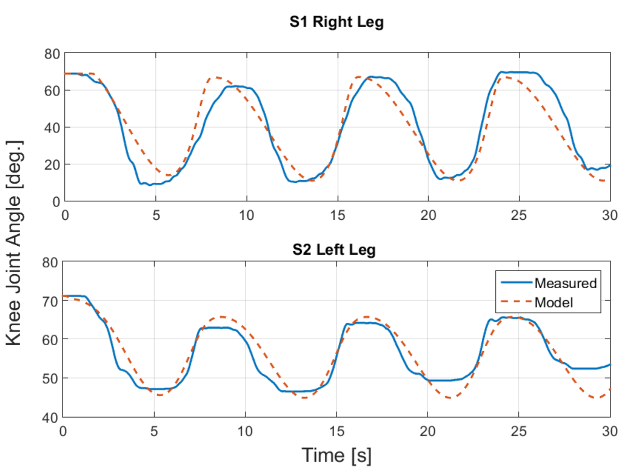
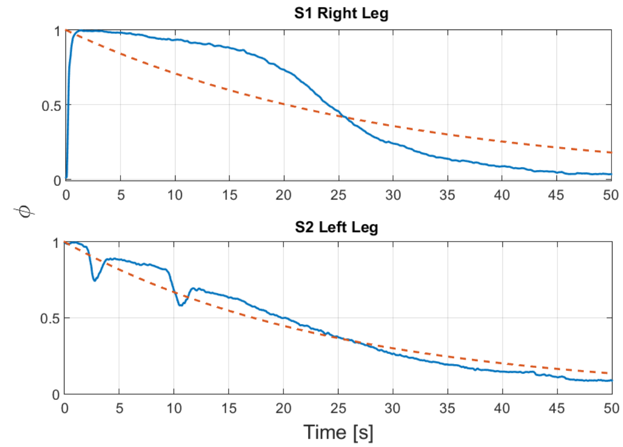
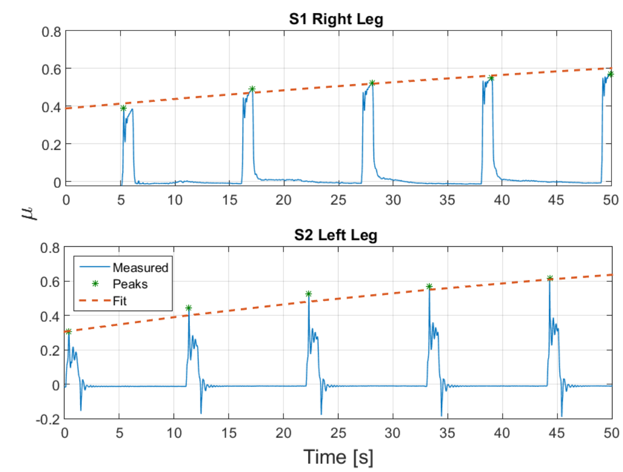


Figure 7: The results from parameter Test 5 for both participants. A known, sinusoidal stimulation was applied to the quadriceps and the resulting joint angle was measured. This plot shows the measured joint angle and the joint angle predicted by the model with the estimated parameters. RMS error of output of the model matches the measured joint angle are listed on the Table. (I).



(a) Fatigue parameter fit.



(b) Recovery data extraction and parameter fit.

Figure 8: This figure demonstrates the results of the muscle fatigue and recovery parameter estimation test from Test 6. In 8b, recovery curves were fitted based on the peaks, which were extracted from the isometric torque measurements.

listed in the Table. I.

Test 6. The purpose of the following two part test is to estimate the parameters of the muscle fatigue dynamics (μ_{min}, T_f, and T_r in (6)). First a constant stimulation with an amplitude equal to the saturation is used to fatigue the muscles. Assuming that muscle activation reaches steady-state relatively quickly, and because the stimulation amplitude is equal to the saturation amplitude (i.e. $a_{ke} \approx u_{ke} = 1$) the fatigue dynamics can be reduced to $\dot{\mu} = \frac{1}{T_f}(\mu_{min} - \mu)$. Assuming that the muscle is initially unfatigued at the start of the test (implies that $\mu(0) = 1$) the solution to the reduced dynamics is $\mu(t) = \mu_{min} - (\mu_{min} - 1)\, e^{-t/T_f}$.

After the muscles are fatigued, $0.5s$ pulse trains with an amplitude that is equal to the saturation were used every ten seconds to measure the rate at which the muscles were recovering. Assuming that the stimulation pulse trains were sufficiently short during recovery ($a_{ke} \approx 0$) the muscle fatigue dynamics can be reduced to $\dot{\mu} = \frac{1}{T_r}(1 - \mu)$, whose solution is $\mu(t) = 1 + (\mu_r - 1)\, e^{-t/T_r}$ where μ_r is the initial condition that is measured from the first contraction during the recovery period. A least-squares nonlinear curve fitting algorithm was used to solve for the parameters μ_{min}, T_f , and T_r that best fit the time responses of the muscle during fatigue and recovery to the normalized load cell measurements from the fatigue and recovery portions of the test. The normalized load cell data and the plot of the fatigue state that best fits the measured data during fatigue and recovery are shown in Fig. 8.

The results of the parameter estimation for the both legs of the S1 participant (able-bodied) and both legs of S2 participant (with SCI) are shown in the Table I. The parameters in Table I are subject specific, and are specific to the stimulation pulse width and frequency used during the parameter estimation.

Parameters of S1		
Parameter	Right Leg	Left Leg
I_t [mA]	18.10	21.21
I_s [mA]	60.00	60.00
m [kg]	4.69	4.68
l_c [m]	0.37	0.37
J [kg m^2]	0.19	0.18
θ_{eq} [rads]	1.20	1.18
d_1 [Nm]	2.66×10^{-14}	2.40×10^{-14}
d_2 [Nm]	1.68	3.50
d_3 [Nm]	1.64	1.48×10^{-9}
d_4	1.59	3.47
d_5 [Nm]	0.76	4.37
d_6	-39.09	-39.97
ϕ_0 [rads]	5.29×10^{-8}	1.88×10^{-8}
c_0 [Nm]	78.78	80.36
c_1 [Nm]	55.76	67.12
c_2 [Nm]	-49.02	-56.25
c_3	1.44	1.04
T_a [sec]	0.26	0.25
μ_{min}	2.61×10^{-9}	3.15×10^{-9}
T_f [sec]	29.17	30.14
T_r [sec]	48.09	41.50
RMS [deg.]	8.10	6.06

Parameters of S2		
Parameter	Right Leg	Left Leg
I_t [mA]	20.00	20.00
I_s [mA]	70.00	70.00
m [kg]	2.10	2.59
l_c [m]	0.18	0.16
J [kg m^2]	0.18	0.24
θ_{eq} [rads]	1.30	1.24
d_1 [Nm]	13.53	11.01
d_2 [Nm]	1.95	2.01
d_3 [Nm]	2.29×10^{-7}	8.22×10^{-9}
d_4	11.71	15.40
d_5 [Nm]	2.59×10^{-6}	0.99
d_6	-6.40×10^{-5}	-21.27
ϕ_0 [rads]	1.03	0.85
c_0 [Nm]	51.08	30.37
c_1 [Nm]	-21.86	9.58
c_2 [Nm]	-17.49	-17.49
c_3	1.69	0.90
T_a [sec]	0.41	0.50
μ_{min}	6.54×10^{-2}	2.55×10^{-10}
T_f [sec]	9.34	24.89
T_r [sec]	68.19	71.46
RMS [deg.]	6.18	2.95

Table I: Musculoskeletal parameters estimated for the right and left legs of each participant.

During control validation experiments these stimulation parameters were kept constant and only the stimulation amplitude was input to the system. Small differences that can be seen between the right and left leg of each participant can possibly be attributed to asymmetry in the muscle structure between the participants right and left legs, and variations in positioning of the electrodes between each leg.