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The robustness of a second order sliding mode approach to human gait simulation

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Abstract:
Nonlinear control techniques such as the sliding mode approach can provide a useful alternative to computationally intensive optimisation strategies in the simulation of human gait. Application of the sliding mode approach has the added advantage that it can provide estimates of internal parameters/signals and can be used to measure and/or monitor the deviation from normal gait in patients who are suffering from conditions such as osteoarthritis, for example. Effectively the principle of the equivalent injection that has been long used for fault detection and diagnostics in engineering machinery can be seen to provide a useful new dimension to gait analysis. Any such model, however, requires apriori estimates to be made of the physical dimensions and muscle characteristics of an individual. The fidelity of any resulting gait analysis will be dependent upon the degree to which the system is sensitive to the selection of such parameters. This paper investigates the effect of inducing errors in the mass, scale and proportions of the individual when simulating the gait cycle of ten normal subjects using a second order sliding mode approach. Seven experiments were carried out to examine the influence of errors up to 50% in the assumed parameters of a three-dimensional musculoskeletal lower body model. The second order sliding mode based simulated gait process was concluded to be sufficiently robust to body segment parameter variation that the use of scaled default parameters can be justified in the gait simulator.

Keywords: Estimation in systems biology; Biomedical system modeling; second order sliding modes

1. INTRODUCTION

Computer modelling is increasingly used as a tool to gain understanding of the processes involved in human movement [Neptune and Kautz, 2001, Anderson and Pandy, 2003, Goldberg et al., 2003, Thelen et al., 2003, Neptune et al., 2004, Thelen and Anderson, 2006]. Simulation provides a means to investigate neural control processes involved in coordinated movement or to develop muscle stimulation controllers to activate paralysed muscles.

A problem faced when simulating the human body is that no two are exactly alike. A group of subjects selected for an experiment will vary in height and proportions, weight and mass distribution. It is impractical to take in-vivo measurements of all of these parameters so simulation usually relies on basic scaling of default parameters obtained from cadaver subjects [Neptune and Kautz, 2001, Anderson and Pandy, 2003, Goldberg et al., 2003, Thelen et al., 2003, Neptune et al., 2004, Thelen and Anderson, 2006].

This study investigates the influence of error in the body segment parameters, specifically mass, scale, proportions and mass distribution using a human locomotor model driven by a second order sliding-mode injection [Lister et al., 2006, 2007]. The degree of influence of body parameter error strongly influences the value of simulation as a tool for gait analysis and the development of therapies using computer models of the human body. Large deviation could render such models worthless as the results they generate could not be assumed to be representative of the individuals simulated.

Using a twelve segment, motion tracking neuro-musculo-skeletal model, experiments adjusted the body parameters by values of up to 50% and the variation in motion tracking, joint torque and muscle activity resulting from these induced errors were recorded.

Kinematic and kinetic data from samples collected from ten normal male subjects were used to drive the model, described in the next section. It should be noted that only the measured joint angles are used as inputs to the model. The muscle activity patterns measured in the gait laboratory are used only for analysis of the estimated muscle activity signals generated from the simulator. The following subsection lists the parameter variations applied to the model during the experiments. The results section includes tables of the RMS tracking errors measured during the experiments as well as graphs of joint angles, torques and muscle activity patterns generated by the model.

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2. METHODS

The human locomotor system is represented by a forward-dynamics model incorporating twelve segments and twenty-three degrees-of-freedom (only twenty-one of which were used in this study) developed in Matlab and Simulink using the SimMechanics toolbox, designed for Newtonian modelling of rigid body machines. Body segments are defined according to their masses, moments of inertia and relative positions of joint centres to the centres of mass based on the 3-dimensional lower limb data supplied by Delp to the International Society of Biomechanics website [Delp, 1990] and described elsewhere [Delp et al., 1990, Delp and Loan, 2000]. One degree-of-freedom 'revolute' SimMechanics joints simulate the toe, subtalar and talocrural joints and three rotational degree-of-freedom 'gimbals' were used to model the hips, knees and lumbar joint. 'Prismatics' were applied to either side of the knee joint, effectively adjusting the length of the thigh and shank segments in order to maintain the centre of rotation measured from the subjects. Varus/valgus motion and internal/external rotation of the knee were also constrained to measured kinematics as a more detailed model of the knee joint would otherwise be required. Measured ground reaction forces cannot be distinguished between toe and foot segments but the toe segments are required for calculation of several musculotendon actuator lengths. Toe joints were accordingly constrained to measured trajectories. Ground reaction forces and torques for the model were generated by applying measured force plate readings (taken for ten normal subjects using a Bertec 4060-10 strain-gauge force plate) directly to the foot and then using a spring-damper system (spring constant; $k = 25000Nm^{-1}$, damper constant; $c = 500Nsm^{-1}$) in order to constrain the subtalar joint to the measured linear motion of the subject during the measured stance phase. Values for the spring-dampers were arrived at experimentally. A spring-damper constraint ($k = 1Nrad^{-1}$ $c = 0.1Nsrad^{-1}$) is also applied to the vertical ground reaction torque while the cross product of the horizontal error in the foot segment vertical axis and the ground reaction force is used to effectively adjust the centre of pressure with foot orientation.

Residual forces were applied to the upper body segment in the form of a spring-damper with parameters $k = 1000Nrad^{-1}$ and $c = 100Nsrad^{-1}$ acting on the orientation error of the segment.

Muscle parameters [Delp, 1990] were incorporated into a set of forward-dynamics modified Hill-type musculotendon actuators [Delp and Zajac, 1992, Zajac et al., 1986, Zajac, 1989] of which forty-three cross the right leg joints (the simulated gluteus maximus, medius, minimus and adductor magnus muscles are each split into three components sharing a common excitation signal which provides more accurate force distribution from these large muscles). Activation dynamics describing the excitation-contraction coupling of the muscle to neural excitation (scaled between 0 and 1) were represented by a first order equation with activation and deactivation time constants of 12 ms and 24 ms respectively [Piazza and Delp, 1996]. Mechanical behaviour of the muscle was simulated by a lumped-parameter model including muscle active and passive force-length, muscle force-velocity and tendon length-force relationships [Delp, 1990, Delp and Zajac, 1992, Zajac et al., 1986, Zajac, 1989]. Musculotendon lengths and torque arms were calculated geometrically from the origin, insertion and wrapping points describing the paths of the muscles across the joints [Delp, 1990]. Further details of the model can be found in [Lister et al., 2007] and must be omitted here because of space limitations.

Synchronised kinematic, kinetic and EMG gait data was collected from ten normal male subjects aged between 19 and 31 approved by the University Hospital Leicester NHS Trust Clinical Ethics Committee. Kinematic motion tracking data was obtained using a six camera Qualysis ProReflex Motion Capture/Analysis System and kinetic measurements were taken using a Bertec 4060-10 strain-gauge force plate.

Motion tracking of the kinematic data was performed via a super-twisting second-order sliding-mode controller from Levant et al. [2000] taking the form:

$$u = u_1 + u_2$$

$$u_2 = \begin{cases} -\lambda|\sigma_0|^\rho sign\sigma, & |\sigma| > \sigma_0 \\ -\lambda|\sigma|^\rho sign\sigma, & |\sigma| \le \sigma_0 \end{cases}$$

$$\dot{u}_1 = \begin{cases} -u, & |u| > 1.5 \\ -\alpha sign\sigma, & |u| \le 1.5 \end{cases}$$

with the switching function $\sigma(e, \dot{e}) = ce + \dot{e}$ where c is a strictly positive design scalar and e is the error in the system variable to be controlled, in this case the difference between the measured and simulated joint positions. Controller parameters used were selected to minimise tracking errors without producing excessive antagonism:

$$\begin{aligned} c &= [15\ 15\ 15\ 15\ 75\ 75] \\ \lambda &= [0.75\ 0.75\ 0.75\ 0.6\ 0.5\ 0.5] \\ \sigma_0 &= 5 \\ \rho &= 0.25 \\ \alpha &= [0.4\ 0.4\ 0.4\ 0.8\ 0.8\ 0.8] \end{aligned}$$

The robustness of the control system ensures that it is not necessary to account for the dynamic behaviour of the muscles when calculating activation signals or to provide optimisation criteria to resolve the muscle actuator redundancy. The control action is used to activate the muscles directly according to their alignment to the desired motion. The instantaneous activity level of each muscle, a_m, is defined according to the degree of coincidence between the three-dimensional control signal for the joint, C_j, and the direction vector of the muscle torque, T_m:

$$a_m = \begin{cases} 0, & C_j \cdot T_m < 0 \\ C_j \cdot T_m, & 0 < C_j \cdot T_m < 1 \\ 1, & C_j \cdot T_m > 1 \end{cases}$$

For bi-articular and tri-articular muscles C_j represents a six- or nine-element array (as appropriate) combining the three-dimensional control signals for all of the joints crossed. T_m is then the unit vector of the torque direction vectors across the joints combined in the same way.

The body segments of the model are defined by mass, moment-of-inertia and the relative positions of the joint centres to the centre of mass and scaled according to the weight, thigh lengths and shank lengths of each subject. The purpose of this study is to quantify the effects of

Table 1. Mean RMS tracking errors of the joints under parameter variation for the seven experiments described.

Exp.	-50%	-20%	-10%	-5%	0%	+5%	+10%	+20%	+50%
1	2.97	2.70	2.65	2.66	2.68	2.71	2.76	2.98	4.97
2	2.66	2.69	2.68	2.68	2.68	2.68	2.68	2.69	2.72
3	2.83	2.65	2.76	2.66	2.68	2.70	2.74	2.85	4.30
4	2.71	2.70	2.69	2.69	2.68	2.67	2.67	2.66	2.58
5	2.65	2.66	2.67	2.67	2.68	2.69	2.70	2.72	2.84
6	2.89	2.69	2.67	2.66	2.68	2.68	2.70	2.75	3.28
7	2.36	2.52	2.60	2.64	2.68	2.71	2.76	2.87	3.34

Table 2. Mean RMS joint torques

Exp.	-50%	-20%	-10%	-5%	0%	+5%	+10%	+20%	+50%
1	22.47	19.77	19.29	19.20	19.14	19.21	19.54	20.79	31.40
2	18.76	19.18	19.14	19.10	19.14	19.16	19.13	19.21	19.02
3	22.44	19.84	20.67	19.22	19.14	19.25	19.54	20.80	26.69
4	19.44	18.96	19.07	19.09	19.14	19.17	19.19	19.31	20.30
5	19.66	19.31	19.24	19.18	19.14	19.09	19.07	19.01	18.94
6	23.44	20.88	19.97	19.54	19.14	18.77	18.37	17.92	19.27
7	19.02	19.13	19.19	19.16	19.14	19.12	19.06	19.07	19.30

variation to these parameters. Note that to compute actual values of such parameters for a range of subjects is impossible and thus a formal theoretical analysis of parameter variations cannot be conducted. A range of variations will be assumed that reflects both likely and extreme variations.

The following experiments were performed across the data set:
0. The control. No parameter adjustments made.
1. The mass of every segment was adjusted by ±5%, 10%, 20% and 50%.
2. The moment of inertia of every segment was adjusted by ±5%, 10%, 20% and 50%.
3. The length of every segment was adjusted by ±5%, 10%, 20% and 50%.
4. ±5%, 10%, 20% and 50% of the mass from the upper body and pelvis segments was redistributed proportionally to the leg segments.
5. ±5%, 10%, 20% and 50% of the mass from the right leg segments was redistributed to the left leg segments.
6. The thigh lengths were adjusted by ±5%, 10%, 20% and 50% and the shank lengths by ∓5%, 10%, 20% and 50%.
7. The pelvis size was adjusted by ±5%, 10%, 20% and 50%.

The ten kinematic data sets were used as ideal reference trajectories for the model and the mean of the model outputs taken. The simulated muscle lengths are calculated using a default geometric data set and based on the joint angles. The output force is then scaled according to the normal segment lengths of the subject.

3. RESULTS

The results presented here refer only to the right leg as displaying the very similar data from the left leg would be largely redundant. To avoid initial condition settling time issues particularly as a result of non-optimal muscle contraction lengths and velocities, one complete gait cycle was taken from each sample and repeated to produce two identical, consecutive gait cycles. Only the second gait cycle is considered.

Table 3. Mean RMS joint torque changes

Exp.	-50%	-20%	-10%	-5%	+5%	+10%	+20%	+50%
1	10.94	5.024	2.751	1.538	1.645	3.201	6.685	19.15
2	3.356	1.824	0.946	0.597	0.579	0.838	1.486	2.801
3	13.06	6.496	4.475	2.024	2.127	3.929	7.645	16.04
4	5.282	1.468	0.852	0.537	0.476	0.809	1.406	4.672
5	3.386	1.549	0.861	0.472	0.496	0.854	1.546	3.523
6	9.303	4.667	2.486	1.346	1.289	2.585	4.957	11.45
7	3.050	1.455	0.826	0.504	0.553	0.917	1.594	3.884

Table 4. Average percentage muscle activity changes

Exp.	-50%	-20%	-10%	-5%	+5%	+10%	+20%	+50%
1	4.475	0.805	0.361	0.263	-0.135	0.040	0.884	5.453
2	0.574	0.310	0.047	0.001	-0.002	-0.086	-0.162	-0.035
3	3.000	0.396	-0.669	0.127	0.054	0.211	0.810	3.242
4	-1.273	0.101	-0.002	-0.011	0.011	-0.066	-0.015	-1.921
5	1.032	0.465	0.270	0.160	-0.097	-0.149	-0.104	-0.191
6	4.510	2.006	1.073	0.514	-0.471	-1.117	-2.103	-4.120
7	1.587	0.662	0.328	0.227	-0.194	-0.383	-0.667	-0.981

The mean of the RMS tracking errors of the ten subjects for the joint angles of the right leg are displayed in table 1 for each experiment. Recall that the model seeks to construct physiological levels of activation and thus the zero errors that could be expected from direct manipulation of a torque input are not achieved. Table 2 displays the mean RMS joint torques for the right leg and Table 3 shows the mean change in RMS torque relative to the control. An average of the percentage change in activity of each muscle (relative to the control level) weighted by the maximum voluntary contraction forces was produced (Table 4). This approximates the total difference in muscle activity of the leg.

4. DISCUSSION

Immediately striking in Table 1 is the lack of variation as physical parameters are adjusted. The largest tracking error occurs with a 50% increase to the body mass in experiment 1 with significant variation in hip adduction/abduction and internal/external rotation evident in Figure 1. The model employs the right hand rule and the sign of the data as represented in all figures reflects this rather than anatomical nomenclature (ie. flexion is positive for the hip and negative for the knee). Note that the tracking performance is maintained with physiological levels of activation. Reduction in parameters such as subject mass and scale in many cases produce similar changes in the tracking behaviour to increases in the parameters (Table 1). It is important to remember that the controller has been tuned to minimise the tracking error when operating with measured subject parameters and as such its performance will degrade as the parameters are moved away from the norm. These results thus report the situation relating to a fixed control strategy and no attempt has been made to redefine the controller for different parameter levels.

Experiment 6 produced some significant change in tracking performance particularly under negative parameter variation which corresponds to a decrease in thigh length and an increase in shank length. While small changes in the tracking performance of one joint are often balanced in another, increasing the thigh length and decreasing the

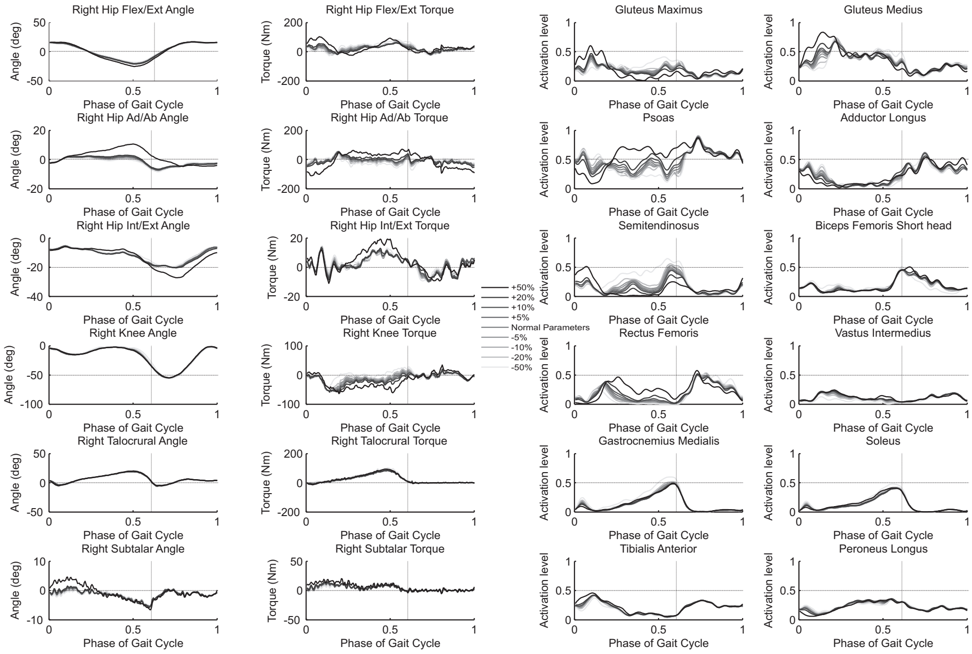

Fig. 1. Mean right leg joint angle, joint torque and muscle activity results for ten subjects when the mass parameters of the model are adjusted by ± 5, 10, 20 and 50%

shank length did demonstrate some small overall improvement in the errors with up to 10% variation. The need for ground clearance during the swing phase is neglected.

Mean RMS torque changes (Table 3) show much larger differences than the joint angles, but this is not reflected in the mean RMS joint torques (Table 2) which vary very little even with significant changes to mass and scale. In experiment 1 there is three times more mass in the model at +50% than at -50% but the joint torques do not approach the same change in amplitude. Likewise in experiment 3 the maximum height of the simulation is triple the minimum but the joint torques and muscle activity amplitudes do not reflect this. As an efficient, pendular process human gait takes advantage of gravity and the elastic properties of muscles, tendon and ligament to reduce energy use. On a flat surface the action of the muscles can in part be considered a counter to the mechanical energy losses of the gait process, losses which are evidently not directly proportional to the mass or height. Changes in parameters appear to affect the distribution of these losses, causing large changes in individual joint torques but having little effect on the total until the system begins to approach the limits of the muscles as in the case of a 50% increase in mass (Table 2 and Figure 1).

The average muscle activity changes seen in Table 4 differ by only a few percent even in the most extreme cases. The switched nature of the controller produces antagonism in the muscle activations and a change in torque moves the point about which this oscillation occurs. As a result the total activation of the muscles will change little until the limits of the muscles are approached as occurs at 50% parameter changes in experiments 1, 3 and 6. Figures 1 and 2 demonstrate the most change in the model outputs as they display the effect of global changes to the mass and physical size of the subjects respectively. The system was almost invariant to variation in the moments of inertia and the graphical results are consequently omitted. With parameter variation up to ±20% the pattern of changes in the curves of Figures 1 and 2 is largely only one of amplitude. It is only with variation of ±50% that real changes in the shapes of the curves begin to occur. With ±50% variation in the parameters, both the sliding mode controller and the physical muscle characteristics are approaching saturation.

Given the relationship between joint torque and muscle activity and the very small changes demonstrated in experiments 4 to 7 only the joint angles and torques are displayed for these experiments. Mass redistribution as demonstrated in Figure 3 shows very little effect on the results, the largest changes occurring when mass is shifted from one leg to the other in experiment 5. When mass is shifted to and from both legs collectively in experiment 4 (Figure 3) the muscle activity changes are smaller suggesting that symmetry is more important a factor than quantity. Changing the lengths of the thigh and shank segments changes the proportions of the leg in a manner increasingly inappropriate for the gait patterns simulated and in practice ground clearance issues would come into play before changes as extreme as 50% were made to the segment lengths but the ground reaction system employed by the model does not allow for ground contact during the measured swing phase. Experiment 6 (Figure 4) demonstrates that there is little effect in moving the position of the knee joint. It is only in the most extreme cases tested that significant distortions emerge in the results. Experiment 7 (Figure 4) demonstrates very little change in anything but the hip adduction/abduction and the internal/external rotation as might be expected from a change in pelvis size. Flexion/extension of the hip and knee show almost no deviation whatsoever even with 50% variation to the pelvis size.

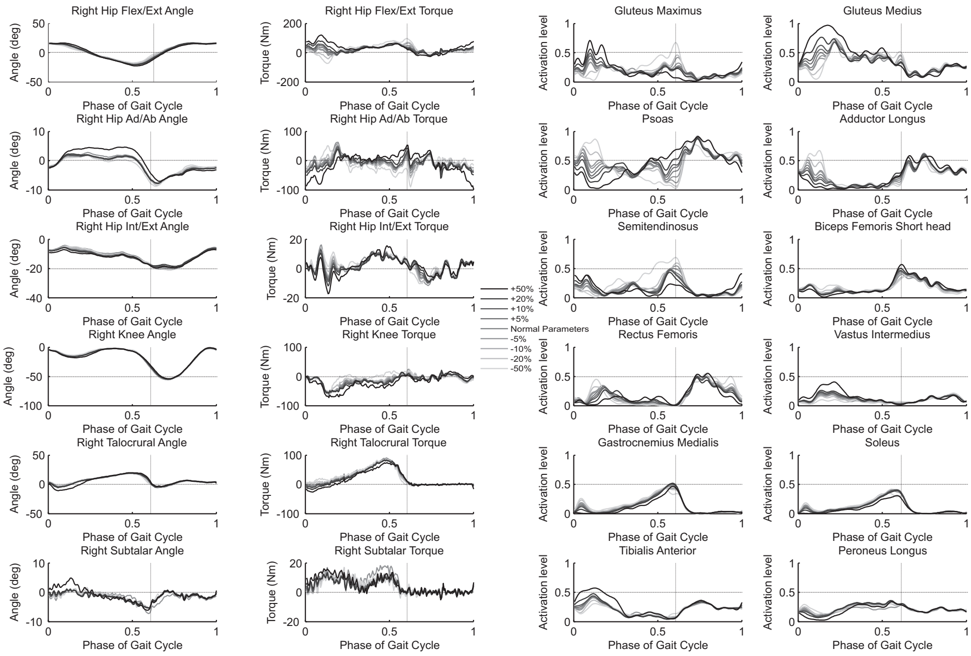

Fig. 2. Mean right leg joint angle, joint torque and muscle activity results for ten subjects when the segment length parameters of the model are adjusted by ± 5, 10, 20 and 50%.

5. CONCLUSION

The minimal changes resulting from errors of 50% in the nominal parameters is evident. The muscle activity patterns generally retain their shapes. Overall the simulated locomotive system is robust enough that the use of default parameters scaled to the height and weight of the subject can be justified within a sliding mode based gait simulator as the effect of any error is negligible.

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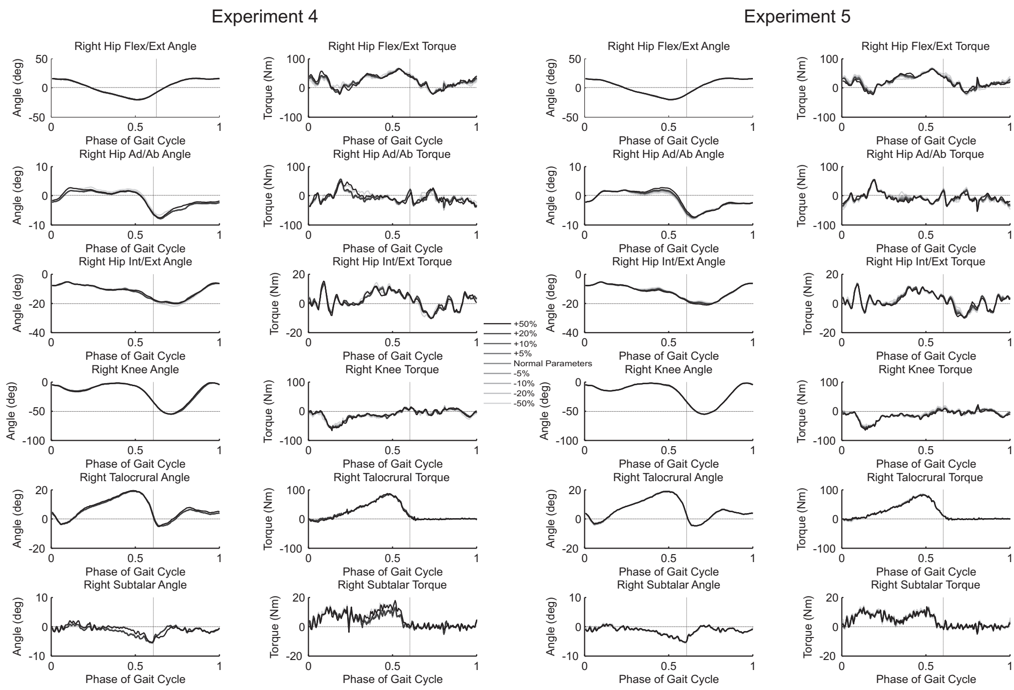

Fig. 3. Mean right leg joint angles and joint torques for experiments 4 (upper body and pelvis mass is redistributed to the legs) and 5 (right leg mass is redistributed to the left leg).

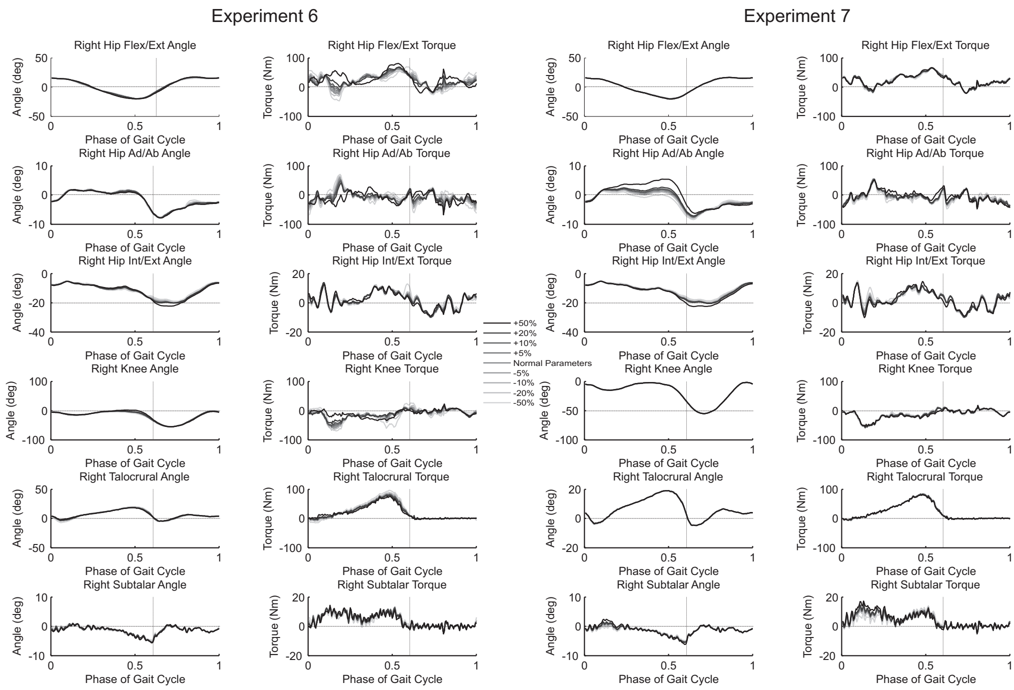

Fig. 4. Mean right leg joint angles and joint torques for experiments 6 (thigh and shank segment lengths are adjusted by opposing amounts) and 7 (pelvis size is adjusted).