Gait & Posture 43 (2016) 24–30

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Rectus femoris transfer surgery affects balance recovery in children with cerebral palsy: A computer simulation study

Misagh Mansouri [a,*], Ashley E. Clark [a], Ajay Seth [b], Jeffrey A. Reinbolt [a]

[a] Mechanical, Aerospace and Biomedical Engineering, University of Tennessee, Knoxville, TN, USA
[b] Bioengineering, Stanford University, Stanford, CA, USA

ARTICLE INFO

Article history:
Received 2 July 2014
Received in revised form 6 July 2015
Accepted 13 August 2015

Keywords:
Balance recovery
Dynamic simulation
Neuromuscular control
Cerebral palsy
Rectus femoris transfer surgery simulation
Postural perturbations

ABSTRACT

Stiff-knee gait is a troublesome movement disorder among children with cerebral palsy (CP), where peak swing phase knee flexion is diminished due to over-activity of the rectus femoris muscle. A common treatment for stiff-knee gait, rectus femoris transfer surgery, moves the muscle's distal tendon from the patella to the sartorius insertion on the tibia. As a biarticular muscle, rectus femoris may play a role in motor control and have unrecognized benefits for maintaining balance. We used musculoskeletal modeling, neuromuscular control, and forward dynamic simulation to investigate the role of rectus femoris tendon transfer surgery on balance recovery after support-surface perturbations for children with CP adopting two different crouched postures. We combined both high-level supraspinal and low-level spinal signals to generate 92 muscle excitations for tracking experimental whole body center of mass positions and velocities. Stability during balance recovery was evaluated by the minimum distance between the extrapolated center of mass and base of support boundary (b_{min}) and the minimum time to reach the boundary (TtB_{min}). The balance recovery of pre-surgical simulations ($b_{min} = 2.3 + 1.1$ cm, $TtB_{min} = 0.2 + 0.1$ s) were different ($p = 0.02$), on average, than post-surgical simulations ($b_{min} = -4.9 + 11.4$ cm, $TtB_{min} = -0.1 + 0.3$ s) of rectus femoris transfers. The moderate crouch simulations ($b_{min} = 2.4 + 0.4$ cm, $TtB_{min} = 0.2 + 0.03$ s) were more stable than the mild crouch simulations ($b_{min} = 1.2 + 0.3$ cm, $TtB_{min} = 0.1 + 0.02$ s) following anterior translations of the support surface. These findings suggest that tendon transfer of rectus femoris affects balance recovery in children with CP.

1. Introduction

Stiff-knee gait is one of the most prevalent and troublesome movement abnormalities among children with CP and its symptoms are treatable with surgery involving the rectus femoris muscle. There are approximately 573,000 individuals affected by spastic CP in the United States. Patients experiencing stiff-knee gait typically adopt energy-inefficient movements to compensate for reduced toe-clearance and avoid tripping or falling due to spastic, over active lower-extremity muscles. Stiff-knee gait is often accompanied by excessive knee flexion during stance (crouch gait) [1], which may also play a role in balance control [2]. Rectus femoris over-activity is attributed as a primary cause of stiff-knee gait [3]. Various surgical procedures can treat CP symptoms by altering the function of problematic muscles including the rectus femoris [4,5]. Rectus femoris transfer surgery is one such procedure, which aims to convert the muscle's knee extension moment to a knee flexion moment. Unfortunately, outcomes following surgery are met with variable success [6,7].

Rectus femoris is a biarticular muscle, acting as both a hip flexor and knee extensor, and may play a role in maintaining balance during dynamic tasks [8,9]. Some have suggested biarticular muscles act as energy transfer straps across joints [10]. Furthermore, others have suggested biarticular muscles play a distinct role in motor control and are among the first muscles affected in persons with CP [11]. As a biarticular muscle, rectus femoris may be important in maintaining balance and may have deleterious effects when transferred to become a knee flexor.

Recommendations for rectus femoris transfer surgeries are generally based on physical examinations along with clinical movement analysis and surgical treatments; the biomechanical consequences and outcomes following rectus femoris transfer

* Corresponding author at: The University of Tennessee, Nathan W. Dougherty Engineering Building, Room 203, 1512 Middle Drive, Knoxville, TN 37996-2210, USA. Tel.: +1 865 974 6808; fax: +1 865 974 5274.
E-mail addresses: misagh@utk.edu, mismanb@gmail.com (M. Mansouri).

http://dx.doi.org/10.1016/j.gaitpost.2015.08.016
0966-6362/

surgery are rarely considered. Physical examination of a single passive joint motion does not address coordinated multi-joint movement. Clinical movement analysis characterizes the motion of limb segments, but not the individual muscle contributions causing this motion. Muscle-actuated simulations and bio-inspired control systems potentially give insights into the complex interaction between neural commands, musculoskeletal geometry, and resulting functional movements. Combining clinical movement analysis and simulation-based approaches may lead to a better understanding of biomechanical consequences of stiff-knee gait and its treatments.

The objective of this study was to determine the effect of simulated rectus femoris transfer surgery on balance recovery after support-surface perturbations for children with CP adopting two different crouched postures. We hypothesized that the stability of pre-surgical simulations would be different (i.e., have better or worse balance recovery) than post-surgical simulations. We tested this hypothesis by comparing the minimum distance from the extrapolated whole body center of mass (CoM_{extrp}) to the base of support (BoS) boundary and the minimum time for the CoM_{extrp} to reach the BoS boundary [12]. We also determined whether moderate crouched postures were more stable than mild crouched postures for balance recovery.

2. Methods

We determined the influence of the biarticular rectus femoris muscle on balance recovery in children with CP by performing the following four steps: (1) creating musculoskeletal models of children with spastic CP before and after rectus femoris transfer surgery adopting either a mild or moderate crouched posture; (2) designing a biologically-inspired, closed-loop control system for balance recovery; (3) generating forward dynamic simulations to examine balance recovery following support-surface translations; and (4) evaluating stability margins using the extrapolated center of mass position and minimum time-to-boundary of the base of support.

2.1. Musculoskeletal models

Three-dimensional musculoskeletal models with 92 muscles and 23 degrees of freedom were constructed in OpenSim [13]. The Hill-type muscle-tendon model was used as the basis for each muscle in the simulations [14]. The foot-ground contact geometry were based on 3D scans of cadaver feet [15] and ground reaction forces were modeled using elastic foundation mesh-based contact [13]. The stiffness and dissipation values were chosen based on material properties of skin (foot) and concrete (ground) materials [16]. The models were scaled to represent the size of children with spastic CP [17]. To simulate rectus femoris transfer surgery, we modified the pre-surgical models to create post-surgical ones by reattaching the distal tendon of the rectus femoris from the patella (Fig. 1a) to the insertion of the sartorius on the tibia (Fig. 1b). The tendon slack length of the transferred muscle was scaled to ensure the muscle fibers operated near their pre-surgical length ranges [3]. The tendon attachments on bones and muscle via-points are defined based on the MRI images of subjects with CP [18] and modeled using OpenSim via points. The unilateral model represented a simulated transfer on the left limb only, while the bilateral model represented a transfer on both limbs. The crouched postures of the models were selected based on average lower-extremity kinematics of children with cerebral palsy adopting mild (ankle, knee and hip angle of 13.8°, 20.9° and 17.5°, respectively) and moderate (ankle, knee and hip angle of 20.6°, 37.3° and 28.1°) crouch gait. For each model to maintain balance during forward dynamic simulations, a control system design was necessary.

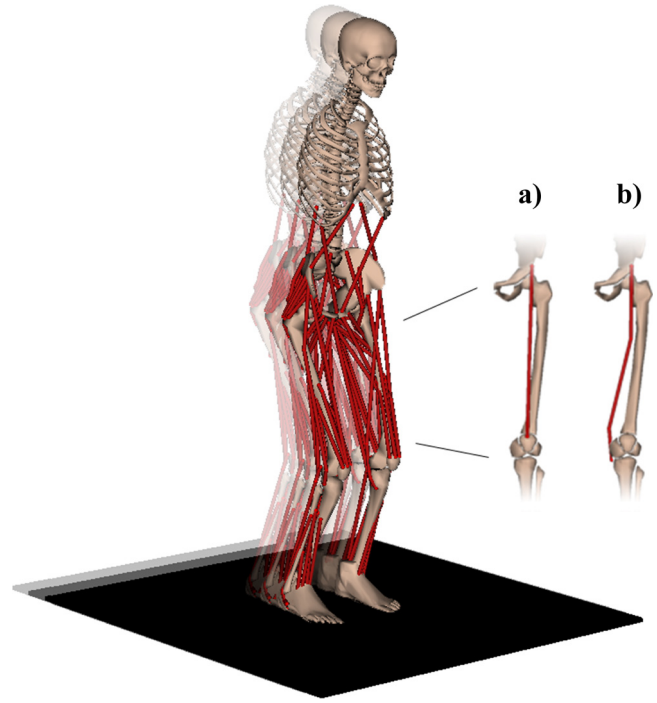

Fig. 1. Three-dimensional, 23 degree-of-freedom musculoskeletal model with 92 muscle-tendon actuators (shown in red) recovering balance on a support surface translating 7.5 cm over 0.55 s with a maximum velocity of 18 cm/s. The musculoskeletal model was used along with a biologically-inspired controller to create forward dynamic simulations of children with mild and moderate crouch to test our hypothesis regarding balance recovery following rectus femoris transfer surgery. Biarticular locations for the rectus femoris muscle are shown as (a) the pre-surgical attachment to patella and (b) the post-surgical transfer to the insertion of the sartorius on the tibia. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of the article.)

2.2. Biologically-inspired controller

Each musculoskeletal model used a biologically-inspired controller to track the experimental CoM displacements similar to the Central Nervous System (CNS) combination of high-level supraspinal and low-level spinal signals in controlling human balance. The high-level controller utilized computed muscle control (CMC) [19] to calculate muscle excitations (peripheral nerve action potential, which initiates the muscles excitation-to-activation process signals) to maintain a static posture despite the forces of gravity. The CMC algorithm utilizes a combination of feedback control, static optimization and forward dynamics to estimate muscles excitations required for tracking a movement [19]. The forward dynamics portion uses muscle excitations calculated from static optimization to drive a model replicating the experimental joint motion. The feedback controller ensures the muscle excitations generate a simulated movement tracking the experimental one. The low-level controller utilized a stretch reflex based on a combination of muscle spindles and Golgi-tendon organ (GTO) [20]. The muscle spindle provides the CNS with information about length and contraction velocity of muscles. The GTO provides the CNS with information about forces in muscle-tendon complex (MTC) and helps to stabilize posture. Together, these afferent mechanisms provide a simple feedback estimate of muscle-tendon length, velocity, and force for controlling postural responses (Eq. (1), Fig. 2).

$$Excitation(t) = CMC(t) + k_p[l_{MTC_{ref}}(t) - l_{CE}(\Delta t) - l_{SE}(\Delta t)] + k_d[-V_{CE}(\Delta t)] \quad (1)$$

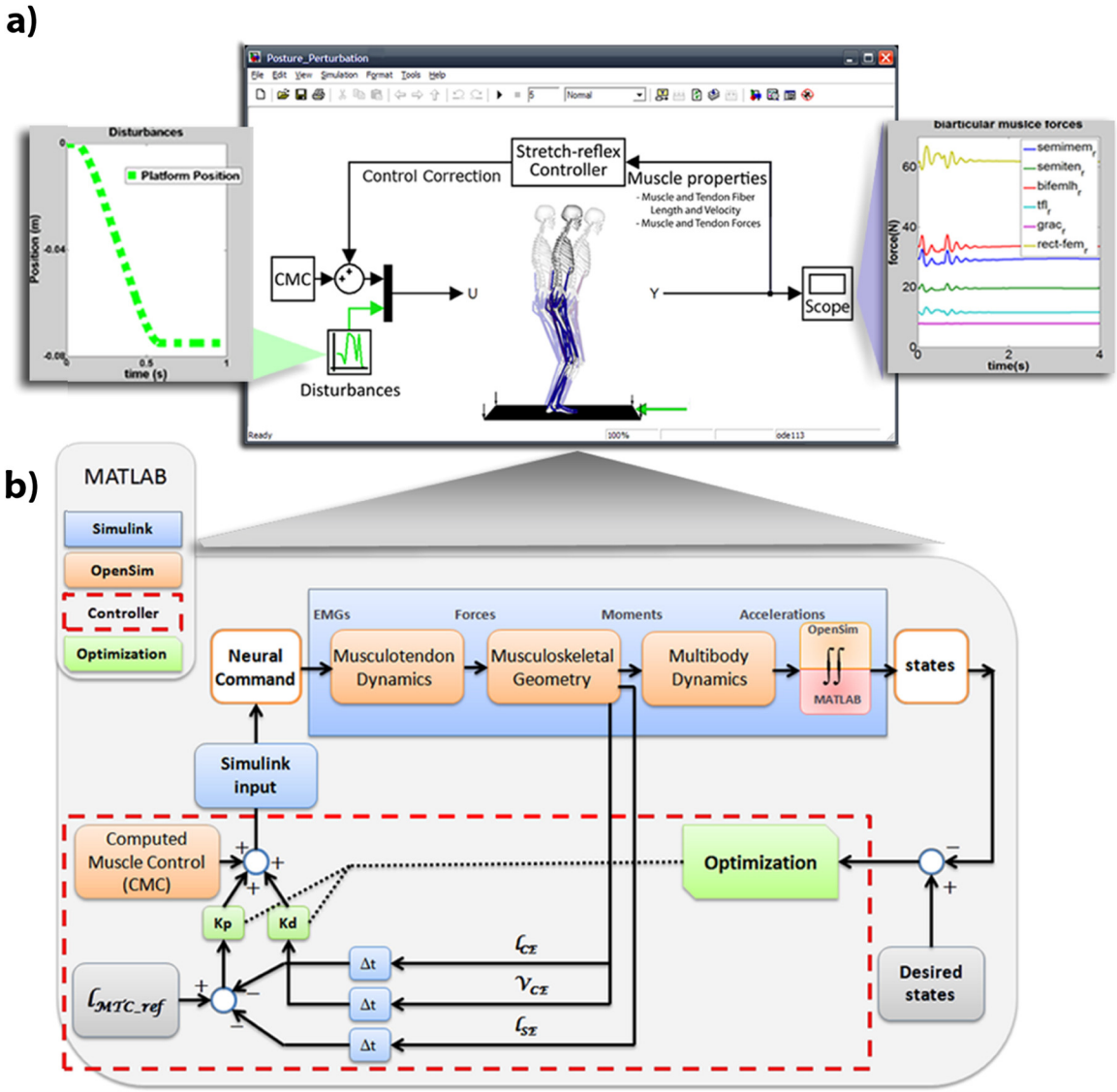

Fig. 2. Schematic of (a) the closed-loop forward dynamic simulation in Simulink and (b) the underlying algorithms. The inputs to the musculoskeletal model, or neural command, were calculated based on the low-level (stretch-reflex controller) and high-level (computed muscle control) feedback control system. Disturbances (e.g., support-surface translations) may also be introduced to the system as inputs. The outputs of the dynamic simulation are the model's states (i.e., joint angles and velocities, muscle activations and fiber lengths). The stretch-reflex controller gains. k_p and k_d, were selected by a non-linear least squares optimization minimizing position and velocity errors between simulated and experimental center of mass displacements. $l_{MTC_{ref}}$ is the reference muscle-tendon complex (MTC) length; l_{CE} is the contractile element (CE) length; l_{SE} is the series elastic (SE), or tendon, length; V_{CE} is the CE velocity; and Δt is the stretch-reflex latency.

$l_{MTC_{ref}}$ is the reference MTC length; l_{CE} is the contractile element (CE), or the muscle fiber length; l_{SE} is the series elastic (SE), or tendon, slack length; V_{CE} is the CE velocity; $\Delta t = t - 0.025$ is the time minus monosynaptic latency (25 ms) [21]; k_p and k_d are the position and velocity gains for the stretch reflex.

Stretch-reflex gains were chosen to represent subjects with non-spastic, typically developing and spastic CP responses to stretch. The gains were determined using a non-linear least squares optimization in MATLAB (lsqnonlin) minimizing the CoM position and velocity errors between simulated and experimental CoM displacements [22] for typically developing subjects. Optimizations were performed over different time periods (0.55 s, 1 s, 2 s and 2.8 s); however, simulation time increase in optimization did not have a major effect on resulting gains after the first 55 s. In addition, we penalized changes in hip, knee and ankle joint angles in the cost function to reduce variability in the resulting motions. Spasticity was simulated by multiplying the non-spastic position and velocity gains by a factor (1.1) to increase the sensitivity of the controller [23].

2.3. Forward dynamics simulations

The musculoskeletal models and biologically-inspired controller were used to create forward dynamic simulations of balance recovery to test our hypothesis regarding stability of children with CP before and after transfer surgery as well as comparing different crouched postures (Fig. 1). We used an OpenSim/MATLAB interface [24] to generate simulations of balance recovery following support-surface translations. The translation was defined using a generalized cross-validation spline function prescribing the position of the support surface as a function of time based on clinical studies [25]. The support surface was translated 7.5 cm in the anterior or posterior directions with a peak velocity of 18 cm/s, which took 0.55 s to complete. A total of 21 forward dynamic simulations were performed for combinations of non-spastic, spastic, mild crouch, moderate crouch, pre-surgical, and post-surgical models (Table 1 and Supplementary Table 1). Each 4-second simulation involved 0.55 s of support-surface translation and 3.45 s of balance

Table 1
Comparison across simulations of balance recovery defined by a spatial stability margin, b_{min}, which was the minimum distance between extrapolated center of mass (CoM_{extrp}) and the BoS boundary, and a temporal stability margin, TtB_{min}, which was the minimum time-to-boundary for the CoM_{extrp} to reach the BoS boundary. The negative values on the table show the CoM_{extrp} is outside of BoS and the model is indeed falling. Additional information about these simulations of balance recovery is available as supplementary material (Supplementary Table 1).

Crouched posture	Support-surface translation	Pre-surgery		Post-surgery			
				Unilateral		Bilateral	
		b_{min} (cm)	TtB_{min} (s)	b_{min} (cm)	TtB_{min} (s)	b_{min} (cm)	TtB_{min} (s)
Mild	Anterior	1.09	0.09	0.98	0.08	1.49	0.12
	Posterior	3.76	0.33	−33.86	−0.55	−8.52	−0.44
Moderate	Anterior	1.96	0.15	2.32	0.19	2.80	0.21
	Posterior	2.42	0.21	−0.20	−0.03	−4.58	−0.33

recovery to record the dynamic stability margins indicating the potential to fall.

2.4. Dynamic stability margins

Based on the criteria for dynamic stability equation from Hof et al. [12], we quantified balance recovery by defining a spatial stability margin (b_{min}) as the minimum distance between extrapolated center of mass (CoM_{extrp}) and the BoS boundary, and defining a temporal stability margin (TtB_{min}) as the minimum time-to-boundary for the CoM_{extrp} to reach the BoS boundary (Supplementary Figure 1). The CoM_{extrp} is the CoM displacement extrapolated in the direction of its velocity. Larger b_{min} values indicate the CoM_{extrp} is farther from the BoS boundary and the simulated condition is more stable than other conditions with smaller b_{min} values. Likewise, larger TtB_{min} values indicate more time is necessary for the CoM_{extrp} to reach the BoS boundary and the condition is more stable than others.

We evaluated our hypothesis regarding the stability of children with CP under different surgical conditions by comparing the changes in TtB_{min} values, which are functions of b_{min} and CoM_{extrp} velocity. A two-tailed, paired *t*-test at the 0.05 significance level was performed against the alternative hypothesis that the stability of pre-surgical simulations is different from post-surgical ones. We analyzed the differences in b_{min} and TtB_{min} stability margins by comparing pre-surgical simulations to post-surgical ones following anterior and posterior support-surface translations; in addition, we analyzed differences between mild and moderate crouched postures.

3. Results

The balance recovery of all pre-surgical simulations of children with CP (b_{min} = 2.31 + 1.12 cm, TtB_{min} = 0.19 + 0.10 s) was significantly different (p = 0.022), on average, than post-surgical ones (b_{min} = −4.94 + 11.38 cm, TtB_{min} = −0.09 + 0.32 s) of both unilateral and bilateral rectus femoris transfers (*note, negative values indicate instability*). Following posterior support-surface translations, all pre-surgical simulations (b_{min} = 3.09 + 0.95 cm, TtB_{min} = 0.27 + 0.09 s) were significantly more stable, on average, than post-surgical ones (b_{min} = −11.79 + 15.10 cm, TtB_{min} = −0.34 + 0.23 s) (Table 1). Following anterior support-surface translations, all pre-surgical simulations (b_{min} = 1.52 + 0.61 cm, TtB_{min} = 0.12 + 0.04 s) were not significantly different, on average, from post-surgical ones (b_{min} = 1.89 + 0.82 cm, TtB_{min} = 0.15 + 0.06 s). The pre-surgical simulations maintained balance following both anterior and posterior support-surface translations (Fig. 3). On the contrary, the post-surgical simulations (unilateral and bilateral transfers) were not able to recover balance following posterior support-surface translations (Fig. 3, bottom row).

Balance recovery was also influenced by crouched posture (Fig. 3, columns). Following anterior support-surface translations, the moderate crouch simulations (b_{min} = 2.36 + 0.42 cm, TtB_{min} = 0.18 + 0.03 s) were more stable than the mild crouch simulations (b_{min} = 1.18 + 0.27 cm, TtB_{min} = 0.095 + 0.02 s) (Table 1). Following posterior support-surface translations, the stability of moderate crouch simulations (b_{min} = −0.78 + 3.54 cm, TtB_{min} = −0.05 + 0.27 s) was not significantly different, compared with mild crouch simulations (b_{min} = −12.87 + 19.18 cm, TtB_{min} = −0.22 + 0.48 s).

4. Discussion

Rectus femoris transfer surgery is a common procedure for treating stiff-knee gait, converting the function of this biarticular muscle from a knee extensor to a knee flexor, which may influence an individual's ability to maintain balance. Several studies have suggested biarticular muscles act as energy transfer straps across joints, are among the first muscles affected in persons with CP, and play an important role in motor control [8,10,11]. The objective of this study was to determine how simulated transfer surgery of biarticular rectus femoris muscle influences the balance recovery of children with CP. Our simulation results confirm that changing the rectus femoris muscle's function as a knee extensor to a knee flexor had a negative effect on balance recovery during simulations following different perturbations. Pre-surgical simulations were significantly more stable than post-surgical ones during posterior support-surface translations. Balance recovery was also influenced by crouched posture. The moderate crouched postures were more stable than mild crouched postures; however, this difference was only significant following anterior support-surface translations.

The significant differences in balance recovery observed in pre- and post-surgical simulations and during anterior versus posterior support-surface translations can be explained by following biomechanical concepts. For anterior support-surface translations, the joint motions to maintain balance require flexing the knee to compensate for an extension induced by the anterior motion of support surface. For posterior support-surface translations, the joint motions to maintain balance require the simulation to extend the knee to compensate for a flexion induced by the posterior motion of support surface. In addition, dynamic coupling has important effects in musculoskeletal dynamics. Zajac and Gordon [26] showed due to dynamic coupling, the biarticular muscles can induce accelerations in direction opposite to the joint moment they generate. Furthermore, Clark [27] showed that removing a monoarticular knee extensor muscle (vastus medialis) had minimal effects on balance recovery of their models. The simulated results show the rectus femoris play a role modulating the knee kinematics to maintain balance. The transfer surgery changes the role of a biarticular rectus femoris, which adversely affects the balance recovery following support-surface perturbations.

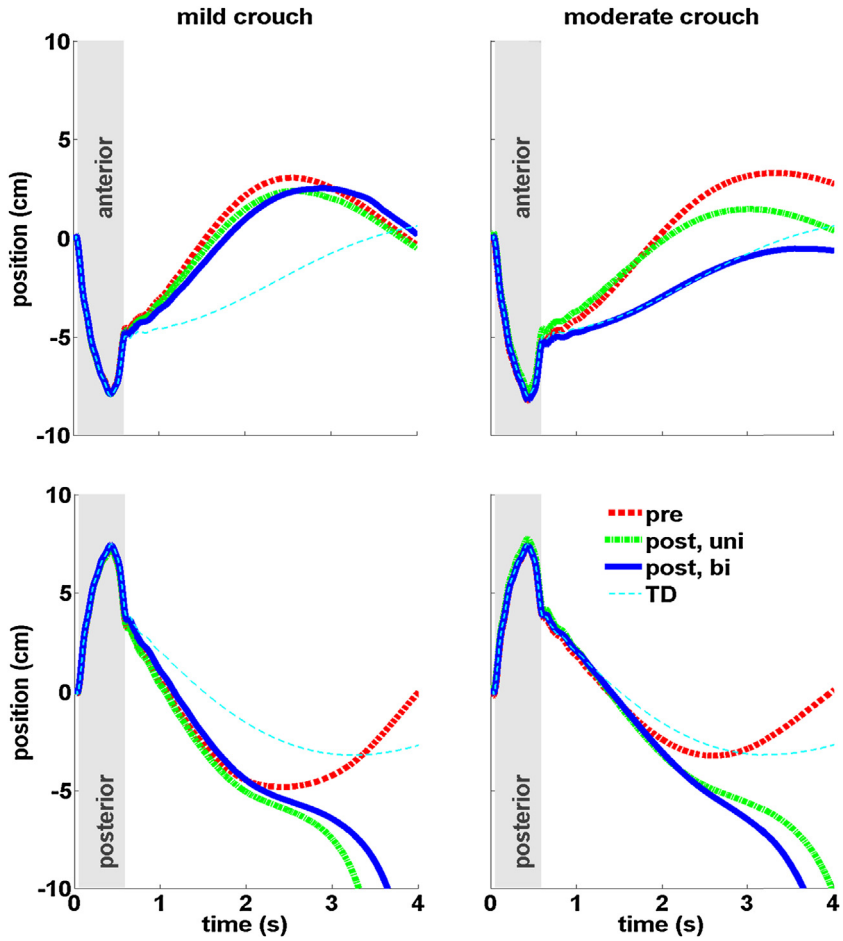

Fig. 3. Extrapolated center of mass displacements relative to the support-surface during anterior (*top row*) and posterior (*bottom row*) translations (*shaded regions*) for simulations before (pre) and after (post, unilateral and bilateral) rectus femoris tendon transfer for mild (*left column*) and moderate (*right column*) crouched postures. The displacements for simulations of the typically developing (TD) model without a spastic stretch reflex are shown as well.

These results for anterior and posterior support-surface translations also hold true for both the mild and moderate crouch simulations. The moderate crouched postures were more stable than mild crouched postures; however, this difference was only significant following anterior support-surface translations. Our simulation results for mild and moderate crouch subjects in this study were similar to crouched postures from Hoang et al. [2] that found several crouched postures afford the lower limb muscles the ability of muscles to generate increased ground reaction forces. Furthermore, our results in both perturbation directions of a control subject undergoing perturbation showed good conformity with the experimental healthy, upright adults from Ting [28] and Henry et al. [22] (Fig. 4). Further research is needed to determine whether mild and moderate crouch simulations possess clinically meaningful differences in the context of balance recovery.

This simulation study had several challenges and our findings should be interpreted within the context of modeling assumptions and capabilities to perform predictive simulations. First, the balance recovery simulations used foot-ground contact modeling that depends upon parameters such as geometry, stiffness and dissipation [23]. The same contact model parameters were used in each simulation [16], so observed differences are not the result of foot-ground differences. Second, the simulations were sensitive to stretch-reflex gains. Multiple optimizations seeded with random initial guesses were used to avoid local minima and determine control parameters for maintaining balance while tracking experimental whole-body CoM position and velocity [22]. Third, we did not consider stretch-reflex latency differences between

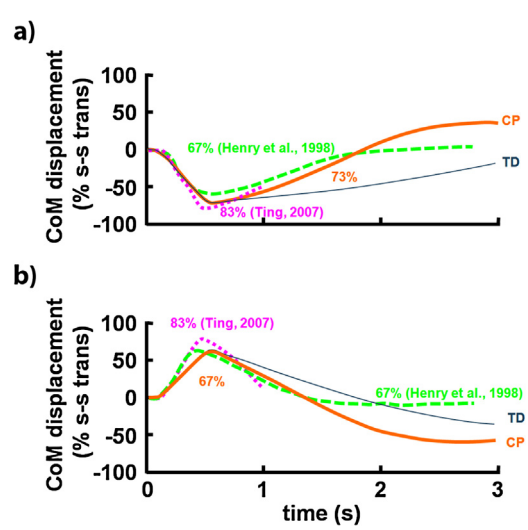

Fig. 4. Simulated center of mass (CoM) displacements for Typically Developing (thin solid line) and pre-surgical, spastic Cerebral Palsy (thick solid line) models with mild crouch compared with experimental CoM displacements of healthy adults with upright posture from Ting, 2007 (square dot line) and Henry et al. 1998 (dash line) during (a) anterior and (b) posterior support-surface translations. To compare the balance recoveries across support-surface translation magnitudes, the CoM displacements were normalized by the support-surface translations in each study.

muscles nor changes in the spasticity after the surgery in this study. Alternatively, we focused on musculoskeletal geometry differences between muscle attachments and muscle moment arms to indicate a change in balance recovery for different postures, tendon locations, and stretch controller gains. Fourth, the difference between the experimental and simulated CoM displacements (Fig. 4) can be explained with the fact that the bio-inspired controller is not a full representation of the CNS including vestibular, visual, and proprioceptive sensory information; however, our controller uses information about the current whole body CoM to adjust the stretch-reflex controller gains. Fifth, muscle recruitment strategies were based upon muscle length and tendon force differences before and after rectus femoris transfer surgery; however, different strategies may be adopted by individual patients. In spite of these challenges, our results remain consistent with experimental balance recovery data [22,28] and do not influence the answers to the underlying questions about what role the biarticular rectus femoris muscle plays in balance recovery before and after transfer surgery for patients adopting mild and moderate crouched postures. Although the predicted stability margins may change if we made different modeling assumptions, the conclusions regarding the surgical alteration of the rectus femoris muscle's function having an influence on balance recovery are unlikely to change.

There are several opportunities for future work to address problems with validating our simulation results and further investigate biomechanical factors contributing to movement abnormalities. Concerning problems with validation, we were unable to find previous studies investigating support-surface perturbations in children with CP following rectus femoris transfer surgeries. Clinical studies [25,29] have reported some data for this population before surgery; however, we were not able to compare our CoM displacements directly with these studies. Unfortunately, for functional tests (e.g., Gross Motor Function Measure), there is no clear indication of the link between balance recovery potential and changes in gross motor function over time. Importantly, it is unethical to separate, or decouple, spinal reflexes (as we have studied here) and supraspinal brain signals in human balance studies. Concerning further investigation, alternative treatment procedures such as distal transfer of the rectus femoris to the iliotibial band for stiff-knee gait and Achilles tendon lengthening for equinus gait may contribute additional insight into the influence of other biarticular muscles in postural control [3]. By using multidirectional (e.g., mediolateral, diagonal, rotational) support-surface perturbations as well as ones with varying magnitudes and velocities, relationships between various biarticular muscles and different directions may be established. Future work may lead to uncovering principles that govern how the CNS selects appropriate muscle patterns to achieve specific movement utilized for balance recovery.

The musculoskeletal modeling, neuromuscular control system design, and forward dynamic simulation in this study identified that distal transfer of the rectus femoris muscle to the insertion of the sartorius may change control of balance provided by the muscle path for patients with spastic CP. The pre-surgical simulations maintained balance following both anterior and posterior support-surface translations while post-surgical ones did not maintain balance following posterior translations. Balance recovery was also influenced by crouched postures where moderate crouch was better than mild crouch in helping recover from anterior support-surface translations but not necessarily posterior translations. This study provides insights about how surgical treatments affects balance recovery and provides a basis for the identification of other biomechanical factors contributing to movement abnormalities and potentially improves surgical and rehabilitation treatments for patients with neurological disorders.

Acknowledgements

We are grateful to Scott Delp and Ayman Habib for helpful discussions. This research was supported in part by the National Science Foundation (CAREER #1253317).

Conflicts of interest statement: We do not have any financial or personal relationships with other people or organizations that could inappropriately influence our manuscript.

Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at http://dx.doi.org/10.1016/j.gaitpost.2015.08.016.

References

[1] van der Krogt MM, Bregman DJJ, Wisse M, Doorenbosch CAM, Harlaar J, Collins SH. How crouch gait can dynamically induce stiff-knee gait. Annals Biomed Eng 2010;38:1593–606.
[2] Hoang HX, Reinbolt JA. Crouched posture maximizes ground reaction forces generated by muscles. Gait Posture 2012;36:405–8.
[3] Fox MD, Reinbolt JA, Ounpuu S, Delp SL. Mechanisms of improved knee flexion after rectus femoris transfer surgery. J Biomech 2009;42:614–9.
[4] Muthusamy K, Seidl AJ, Friesen RM, Carollo JJ, Pan Z, Chang FM. Rectus femoris transfer in children with cerebral palsy: evaluation of transfer site and preoperative indicators. J Pediatr Orthop 2008;28:674–8.
[5] Khouri N, Desailly E. Rectus femoris transfer in multilevel surgery: technical details and gait outcome assessment in cerebral palsy patients. Orthop Traumatol-Sur 2013;99:333–40.
[6] Rethlefsen SA, Kam G, Wren TA, Kay RM. Predictors of outcome of distal rectus femoris transfer surgery in ambulatory children with cerebral palsy. J Pediatr Orthop B 2009;18:58–62.
[7] Sutherland DH, Santi M, Abel MF. Treatment of stiff-knee gait in cerebral-palsy — a comparison by gait analysis of distal rectus femoris transfer versus proximal rectus release. J Pediatr Orthoped 1990;10:433–41.
[8] Prilutsky BI. Coordination of two- and one-joint muscles: functional consequences and implications for motor control. Motor Control 2000;4:1–44.
[9] Schenau GJV, Pratt CA, Macpherson JM. Differential use and control of monoarticular and biarticular muscles. Hum Movement Sci 1994;13:495–517.
[10] Metaxiotis D, Wolf S, Doederlein L. Conversion of biarticular to monoarticular muscles as a component of multilevel surgery in spastic diplegia. J Bone Joint Surg-Br Vol 2004;86B:102–9.
[11] Gage JR, Deluca PA, Renshaw TS. Gait analysis: principles and applications emphasis on its use in cerebral palsy. J Bone Joint Surg 1995;77-A:1607–23.
[12] Hof AL, Gazendam MGJ, Sinke WE. The condition for dynamic stability. J Biomech 2005;38:1–8.
[13] Seth A, Sherman M, Reinbolt JA, Delp SL. OpenSim: a musculoskeletal modeling and simulation framework for in silico investigations and exchange. Proc Iutam 2011;2:212–32.
[14] Thelen DG. Adjustment of muscle mechanics model parameters to simulate dynamic contractions in older adults. J Biomech Eng 2003;125:70–7.
[15] Erdemir A, Sirimamilla PA, Halloran JP, van den Bogert AJ. An elaborate data set characterizing the mechanical response of the foot. J Biomech Eng-T Asme 2009;131.
[16] Anderson FC, Pandy MG. A dynamic optimization solution for vertical jumping in three dimensions. Comput Methods Biomech Biomed Engin 1999;2:201–31.
[17] Hicks JL, Schwartz MH, Arnold AS, Delp SL. Crouched postures reduce the capacity of muscles to extend the hip and knee during the single-limb stance phase of gait. J Biomech 2008;41:960–7.
[18] Asakawa DS, Blemker SS, Rab GT, Bagley A, Delp SL. Three-dimensional muscle-tendon geometry after rectus femoris tendon transfer. J Bone Joint Surg Am 2004;86-A:348–54.
[19] Thelen DG, Anderson FC, Delp SL. Generating dynamic simulations of movement using computed muscle control. J Biomech 2003;36:321–8.
[20] Kistemaker DA, Van Soest AJK, Wong JD, Kurtzer I, Gribble PL. Control of position and movement is simplified by combined muscle spindle and Golgi tendon organ feedback. J Neurophysiol 2013;109:1126–39.
[21] Brodal P. The central nervous system: structure and function. 4th ed. New York: Oxford University Press; 2010.
[22] Henry SM, Fung J, Horak FB. Control of stance during lateral and anterior/posterior surface translations. IEEE Trans Rehabil Eng 1998;6:32–42.
[23] Jansen K, De Groote F, Aerts W, De Schutter J, Duysens J, Jonkers I. Altering length and velocity feedback during a neuro-musculoskeletal simulation of normal gait contributes to hemiparetic gait characteristics. J Neuroeng Rehabil 2014;11:78.
[24] Mansouri M, Reinbolt JA. A platform for dynamic simulation and control of movement based on OpenSim and MATLAB. J Biomech 2012;45:1517–21.

[25] Woollacott MH, Shumway-Cook A. Postural dysfunction during standing and walking in children with cerebral palsy: what are the underlying problems and what new therapies might improve balance? Neural Plast 2005;12:211–9. discussion 63–72.
[26] Zajac FE, Gordon ME. Determining muscle's force and action in multi-articular movement. Exerc Sport Sci Rev 1989;17:187–230.
[27] Clark A. Biarticular muscles influence postural responses. Tennessee; 2012.
[28] Ting LH. Dimensional reduction in sensorimotor systems: a framework for understanding muscle coordination of posture. Prog Brain Res 2007;165:299–321.
[29] Rose J, Wolff DR, Jones VK, Bloch DA, Oehlert JW, Gamble JG. Postural balance in children with cerebral palsy. Dev Med Child Neurol 2002;44:58–63.