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Paralysis of the gastrocnemius medial head differentially affects gait patterns and muscle activity during level and stair ascent locomotion

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ABSTRACT

Background: Prior studies have analyzed the activity of the gastrocnemius (GCM) medial and lateral heads as a single unit because it is technically challenging to separately analyze the function of each component *in vivo*. However, functional variation between the medial and lateral heads is expected due to their anatomical differences.

Research question: What is the independent function of the medial GCM? How does paralysis of the GCM medial head affect gait kinematics?.

Methods: Twelve healthy adults (two males and ten females; age: 28.2 [± 7.72] years) that were scheduled to undergo neurolysis of the tibial nerve branch supplying the medial head of the GCM for aesthetic calf reduction participated in the study. Gait analysis was performed using a computerized opto-electric gait analysis system to measure kinematic data. Surface electromyography (EMG) was recorded simultaneously during the gait analysis. Surface electrodes were placed on seven muscles. Pre-procedure and 1-week and 3-month post-procedure data were compared using a linear mixed model.

Results: During level walking, decreased activity of the GCM medial head did not significantly change gait kinematics. However, a significant increase in GCM lateral head and hamstring activities occurred after a branch nerve block to the GCM medial head. During stair ascent, in contrast to level walking, changes in EMG activity only occurred in the GCM medial head, and post-procedure ankle dorsiflexion angles at the end of the terminal-stance phase significantly increased. Ankle plantarflexion angles during the push-off phase were also decreased when compared with pre-procedure values.

Significance: The human body response to dysfunction of the GCM medial head depended on the type of locomotion.

1. Introduction

The gastrocnemius (GCM) muscle is important for human locomotion including, walking, stair climbing, and running. It generally participates in both ankle plantar flexion and knee flexion, but acts to extend the knee during stance phase of gait. The GCM is composed of the medial and lateral heads and is connected to the calcaneus of the heel by a common tendon, the Achilles tendon, which is shared with the soleus muscle. The soleus and GCM account for 93% of the theoretical plantar flexion torque [1]. The soleus is a one-jointed muscle that runs from the posterior surface of the tibia and fibula to the Achilles tendon. During normal gait, soleus activity commences during the latter part of the loading response phase and continues until the end of the terminal-stance phase. The GCM is a two-jointed muscle that originates from the medial and lateral condyles of the femur.

Within the GCM, functional discrepancies between the medial and lateral heads are anticipated due to their different anatomical locations in the lower leg and different structural characteristics including, muscle thickness, fascicle length, and pennation angle. More specifically, the lateral head has decreased distal extension, has a smaller pennation angle, and the individual muscle fibers of lateral head are 46% longer than the medial head muscle fibers [2–4].

Abbreviations: GCM, Gastrocnemius; EMG, Electromyography; RF, Radio-frequency

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Previous studies have analyzed the function of the medial and lateral heads of the GCM as a single unit because it is technically challenging to analyze each part separately *in vivo*. In addition, selective functional loss of each head due to injury or disease is rarely reported [5].

Thus, the purpose of this study was to investigate the independent function of the medial head of the GCM. Changes in gait patterns and lower extremity muscle activity during level and stair ascent movements were analyzed following neurolysis of the tibial nerve branch, which supplies the medial head of the GCM.

2. Methods

This was a prospective study conducted in a university-affiliated hospital. Ethical approval was granted by our hospital institutional review board and ethics committee (4-2016-0505).

2.1. Participants

The study group consisted of 12 healthy adults (two males and ten females; age: 28.2 [± 7.72] years; height: 162.44 [± 5.04] cm, mass: 57.31 [± 7.07] kg) that were scheduled to undergo neurolysis of the tibial nerve branch for aesthetic calf reduction using 400-kHz radiofrequency nerve ablation. Subjects with a previous history of orthopedic surgery in the lower extremities and/or other problems that influence gait pattern were excluded.

2.2. Protocol

We performed 3D gait analysis and surface electromyography measures before neurolysis of the tibial nerve branch and again 1 week and 3 months after the procedure (Fig. 1).

2.3. Gait analysis

We performed gait analysis using a computerized opto-electric gait analysis system (VICON MX-T10 Motion Analysis System, Oxford Metrics Inc., Oxford, UK; sampling frequency = 100 Hz) to measure kinematic data. We attached 16 reflective markers to the subjects' body according to the VICON Plug-in-Gait model guidelines. Data were collected while subjects walked at a comfortable, self-selected speed on an 8-m pathway. We analyzed the following parameters to compare kinematic data (Table 1).

Table 1
Kinematic data abbreviations.

Joint	Key (°)	Description
Ankle	**ASA1**	Ankle dorsiflexion angle at initial contact
	ASA2	Ankle maximum dorsiflexion angle during stance phase
	ASA3	Ankle maximum plantar flexion angle during push-off
	ASA4	Ankle dorsiflexion angle at the end of the terminal swing
	ATA1	Ankle internal rotation angle at initial contact
	ATA2	Ankle maximum internal rotation angle during stance phase
	ATA3	Ankle maximum external rotation angle during swing phase
	ATA4	Ankle internal rotation angle at the end of the terminal swing
Knee	**KSA1**	Knee flexion angle at initial contact
	KSA2	Knee minimum flexion angle during stance phase
	KSA3	Knee maximum flexion angle during swing phase
	KSA4	Knee flexion angle at the end of the terminal swing
Hip	**HSA1**	Hip flexion angle at initial contact
	HSA2	Hip maximum extension angle during stance phase
	HSA3	Hip flexion angle at the end of the terminal swing
	HCA1	Hip adduction angle at initial contact
	HCA2	Hip max adduction angle during stance phase
	HCA3	Hip adduction angle at the end of the terminal swing

2.4. Electromyography (EMG) measurements

We recorded surface EMG (MA300-XVI, Motion lab systems, USA; sampling frequency = 1000 Hz) simultaneously with the gait analysis. Surface electrodes were placed on the peroneus longus, hamstring, tibialis anterior, vastus medialis, soleus, and medial and lateral heads of the GCM according to SENIAM guidelines [6]. Ultrasonography (Accuvix V10c system; Samsung Medison Co., Seoul, South Korea) was used to confirm that the electrodes were placed correctly. Data were filtered by a 10- to 500-Hz band-pass filter and time-normalized by dividing one gait cycle into 16 equally spaced intervals. Root mean square (RMS) values were calculated for individual muscles during each time interval. Pre- and post-procedure RMS values for each muscle were expressed as a ratio of the maximum pre-procedure RMS value [7]. All data were averaged for three left and three right gait cycles.

2.5. Operation

The shape and contour of the medial head of the GCM were inspected while the subjects stood on tiptoe to induce maximum contraction of the muscles. Ultrasonography in the prone position was used to identify nerve and vessel locations in the popliteal fossa and calf areas. The muscle margins and nerve locations were marked with a surgical pen. After administration of local anesthesia using 1% lidocaine, a radio-frequency (RF) probe that was connected to a 400-kHz-electrical surgical unit (ITC-300D, ITC Co., Korea) was inserted through a small incision in the skin below the popliteal crease. The motor nerve branches that supplied the medial GCM were ablated using an electrical stimulator to transmit RF energy. Insulation of the RF probe, except for the needle tip, facilitated precise ablation of nerve branches with minimal injury to adjacent tissues. To confirm procedure success, leg shape was visually inspected at the first post-operative visit. After visual inspection, the procedure was considered successful when the maximum EMG value decreased to at least 30% of the maximum pre-operative value.

2.6. Statistical analysis

Pre-procedure and 1-week and 3-month post-procedure data were compared using a linear mixed model. Two-sided p-values < 0.05 were considered statistically significant.

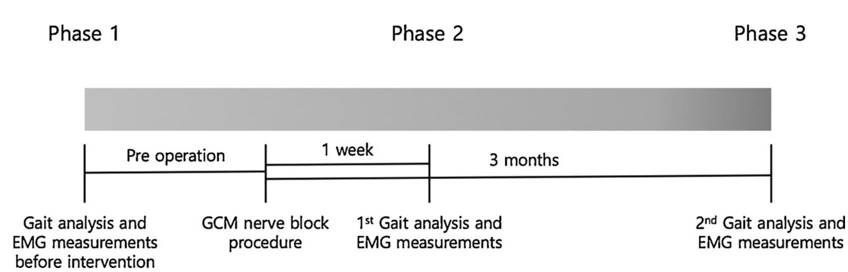

Fig. 1. Study protocol.

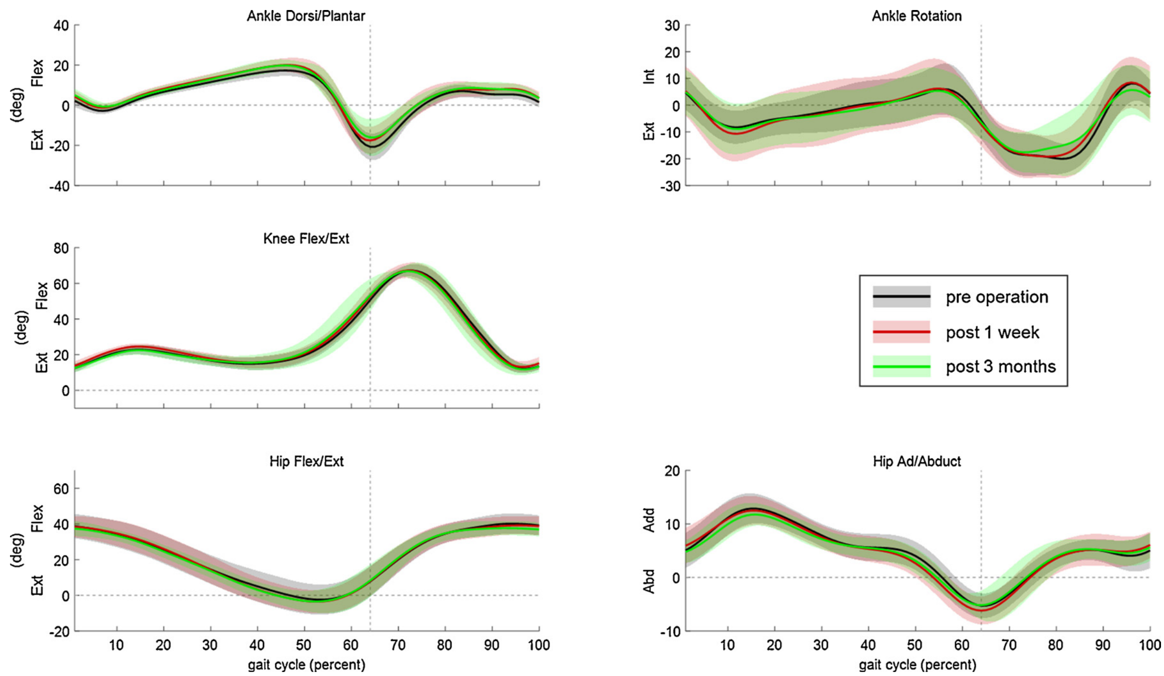

Fig. 2. Changes in gait kinematics during level walking. Vertical dashed lines divide the stance and swing phases of the gait cycle.

3. Results

3.1. Level walking

All kinematic data pertaining to the ankle, knee, and hip joints in the sagittal, coronal, and transverse planes revealed statistically insignificant changes. (Fig. 2)

The RMS values of the GCM medial head during the mid-stance phase significantly decreased after the procedure compared with pre-procedure values ($p < 0.01$). In contrast, the GCM lateral head RMS values at this phase significantly increased after the procedure when compared with pre-procedure values ($p < 0.01$). The hamstring muscle also showed a statistically significant increase in RMS values during the initial contact, mid-stance, and late swing phases 1 week after the procedure ($p < 0.05$) (Fig. 3).

3.2. Stair ascent

Angle joint kinematic data during stair climbing showed statistically significant changes. Post-procedure ankle dorsiflexion angles at the end of the terminal-stance phase (ASA2) increased 3.29 (± 0.82) degrees 1 week after the procedure when compared with the pre-procedure value ($p = 0.007$). At three months post-procedure, the ankle dorsiflexion angle had increased 4.36 (± 1.18) degrees when compared with the initial pre-procedure angle ($p = 0.015$). The ankle plantarflexion angle during the push-off phase (ASA3) also significantly decreased 9.51

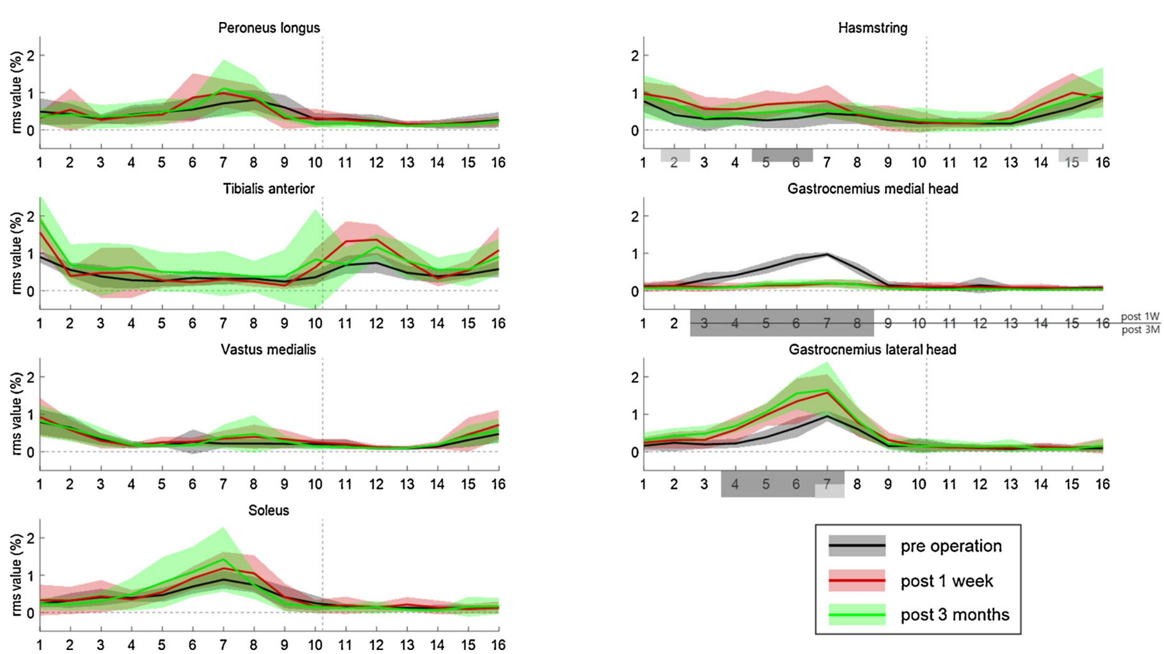

Fig. 3. Changes in linear envelope muscle activity after the medial GCM nerve block during level walking. Dark gray and light gray areas of the x-axis indicate statistically significant differences ($p < 0.01$ and $p < 0.05$, respectively).

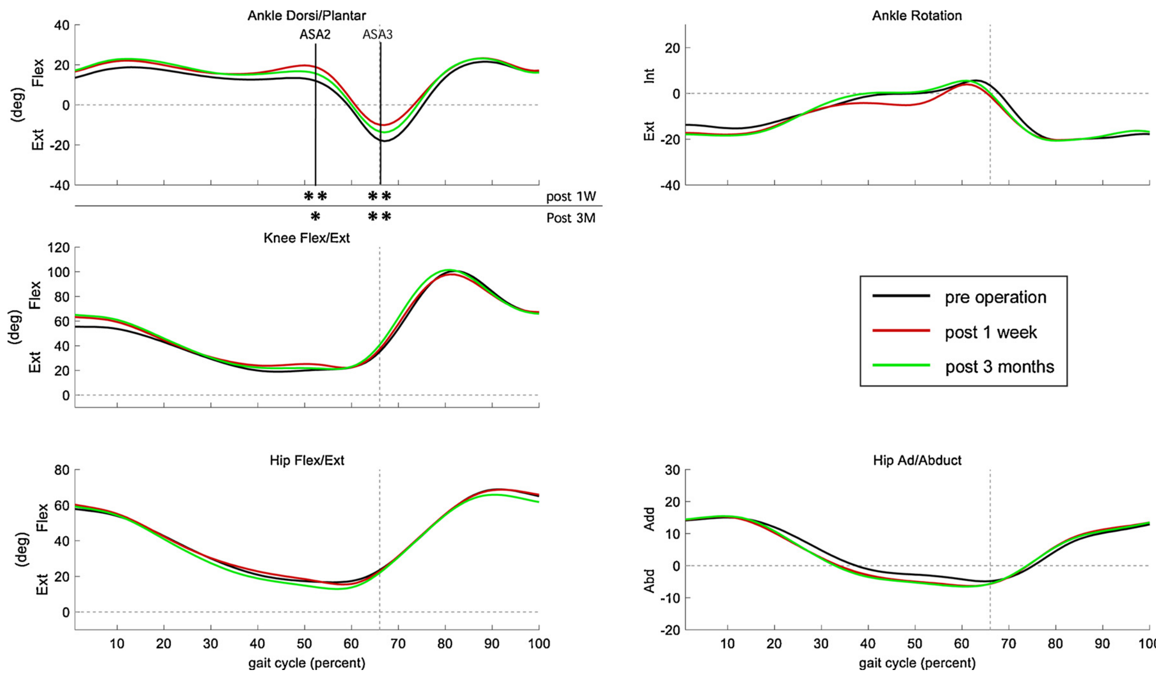

Fig. 4. Changes in gait pattern during stair ascent.
* $p < 0.05$, ** $p < 0.01$, linear mixed model.

(± 1.60) degrees when compared with pre-procedure values 1 week after the procedure. Furthermore, at the 3-month follow-up, ankle plantarflexion had decreased 6.93 (± 1.36) degrees when compared with the pre-procedure value ($p < 0.001$) (Fig. 4).

Post-procedure RMS values of the GCM medial head during the mid-stance phase of the stair ascent were significantly decreased after the procedure compared to pre-procedure values ($p < 0.05$). However, EMG data from the remaining muscles revealed statistically insignificant changes during the stair ascent (Fig. 5).

4. Discussion

To evaluate the effects of GCM medial head paralysis on gait pattern and muscle activity in the lower extremities, we performed gait analysis and EMG measurements during level walking and a stair ascent. Gait kinematics and muscle activity were altered after aesthetic neurolysis of the tibial nerve branch that supplied the GCM medial head. Notably, altered kinematic and muscle activity were specific to the movement being performed, either level walking or stair ascent. Our study revealed that gait kinematics did not change significantly during level walking, but were significantly different during stair ascent.

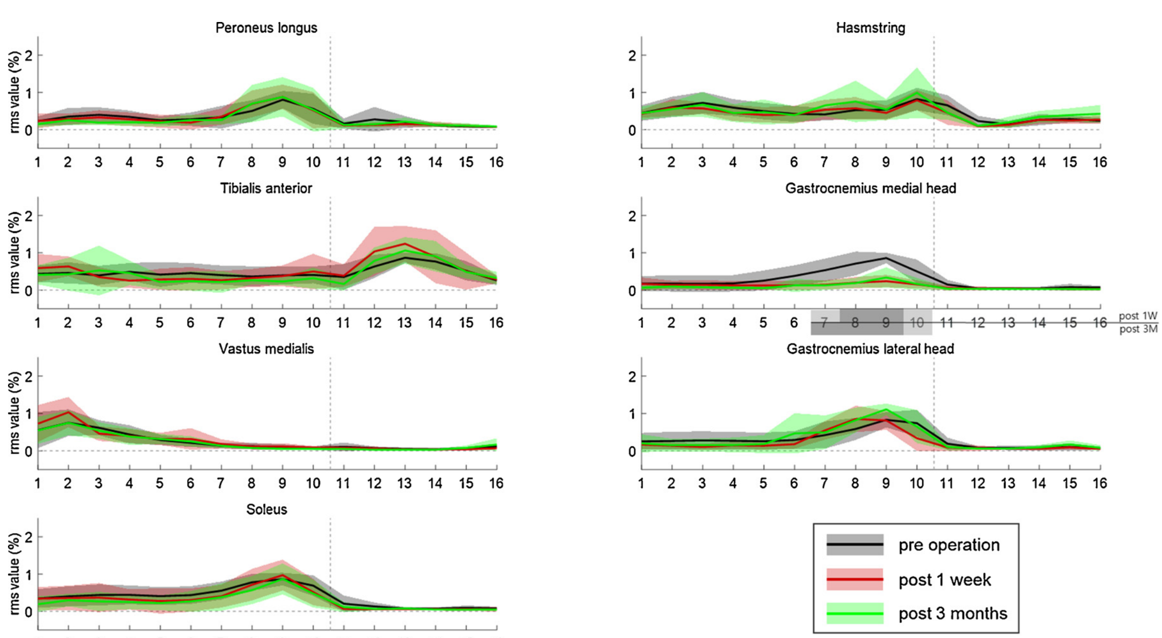

Fig. 5. Changes in linear envelope muscle activity after the medial GCM nerve block during stair ascent. Dark gray and light gray areas of the x-axis indicate statistically significant differences ($p < 0.01$ and $p < 0.05$, respectively).

4.1. Level walking

The GCM medial head plays an important role as the ankle plantar flexor during the terminal-stance and push-off phases. Therefore, we hypothesized that maximum ankle dorsiflexion would increase during the terminal-stance phase and that maximum ankle plantar flexion would decrease during the push-off phase due to paralysis of the medial head. It was reported that ankle dorsiflexion and knee flexion increased in the stance phase during level walking after a tibial nerve block that paralyzed all of the ankle plantar flexors, including the gastrocnemius, soleus, flexor hallucis longus, flexor digitorum longus, and tibialis posterior [5]. In contrast to our hypothesis, in our study, decreased activity of the GCM medial head did not significantly change gait kinematics during level walking. However, the GCM lateral head activity significantly increased 1 week and 3 months after the nerve block to the GCM medial head. Additionally, hamstring activity was significantly increased, but only at the 1-week post-procedure timepoint. The compensatory hyperactivities of other muscles might help to maintain gait kinematics, which were unchanged during level walking after the procedure.

Human bipedal walking is known to be very energy efficient, and proper kinematics of the lower extremities is important for this energy efficiency [8–10]. During walking, the GCM plays an important role beginning from the end of the loading response and extending to the terminal-stance phase. Dynamic stability is altered by continued realignment of the ground reaction force vector to the joints. Focusing on the ankle joint, the ground reaction force vector is located posterior to the ankle joint during the loading response. In mid-stance, the ground reaction force vector advances from the heel (posterior to the ankle joint) to the foot (anterior to the ankle joint), in response to the momentum from the limb swing and bodyweight forward fall, to produce the dorsiflexion moment at the ankle joint. The ankle plantar flexors act to prevent rapid ankle dorsiflexion that decelerates the progression of tibia during this phase. These actions help to keep the ground reaction force vector anterior to the knee joint and maintain a straight leg without collapsing. By the end of mid-stance, the base of body vector lies in the forefoot. With the ankle virtually locked by the plantar flexors, the heel rises as the tibia continues to advance [11]. In the terminal-stance, the vector at the metatarsal head initiates progression over the forefoot rocker. There is also a demand for plantar-flexor activity to support the heel rise. The heel rise opposes the falling body weight and the dorsiflexor torque that is generated by the falling body weight.

This dynamic walking mechanism is based on an inverted pendulum theory, and the alignment of the lower limbs during walking is very important [12–14]. Proper alignment of the lower extremities must be maintained to keep the positional relationship between the ground reaction force vector and the joint center, which allows for minimal muscle activity during walking. If the positional relationship is not maintained, energy consumption increases. In several studies of patients with cerebral palsy, alignment of the lower limbs was disrupted during crouch gait and caused some muscles to work more. The metabolic cost of walking increased and eventually knee pain and degeneration occurred [15–18]. Additionally, in the crouched posture, an increase in joint flexion acceleration, which is caused by gravity, and a decrease in the muscle capacity to produce joint extension accelerations, requires these individuals to produce larger muscle forces to maintain the crouched posture [19]. It is essential to maintain proper alignment of the lower limbs to achieve effective level walking.

Therefore, it appears the ankle plantar flexors play a key role in level walking efficiency. As for the hamstrings that act as a hip extensor, they contribute to upright trunk posture and limb stabilization around initial contact, and contribute to knee extension during stance phase [19,20]. The results of this study indicate that the GCM lateral head and hamstrings compensate for the GCM medial head to maintain gait kinematics during level walking.

4.2. Stair ascent

Notably, there was a kinematic change during the stair ascent. Ankle dorsiflexion increased at the end of the terminal-stance phase (ASA2) and the maximum ankle plantar flexion angle (ASA3) decreased during push-off due to ankle plantar-flexor weakness. In contrast to level walking, compensatory muscle hyperactivities in the GCM lateral head and hamstrings were not observed during stair ascent.

A striking feature of normal walking is the continued smooth, forward movement of the human body center of gravity (CoM). In addition, normal gait patterns are energy efficient and use the natural dynamics of the whole body. However, the movements during a stair ascent are different when compared to normal walking [21–24]. The stair-climbing motion is an action that consumes energy because it moves the CoM upward and forward. In addition, when the magnitude of the forward acceleration is small, the deviation between the pressure center and the zero-moment point is small, and the center of pressure (CoP) is primarily in the support base. Additional voluntary knee flexion during the swing phase is also required to maintain foot clearance during a stair ascent. The most noticeable difference between stair ascent and level walking is that knee extension during the stance phase is very important for moving the CoM upward. The knee joint movement from flexion to extension during the stance phase is very large, which requires the knee extensor muscles to produce a large extension torque and positive vertical power to move the upper body against gravity. Although the ankle joint is also plantarflexed during the mid- to terminal-stance phases, and the joint produces positive work, its role to move the CoM upward is lesser than that during the knee extension motion. In our study, ankle dorsiflexion during the terminal-stance phase increased, but knee kinematics were unchanged during the post-nerve ablation stair ascent. Contrarily, during normal level walking, the knee extensors produced less torque, including negative power during most of the stance phase. The ankle plantar flexors produced greater torque to keep the ground reaction force vector anterior to the knee joint and to maintain a straight leg without collapsing.

Because the pendulum model for level walking does not apply to a stair ascent, proper kinematics that align the hip, knee, and ankle joints appropriate for this model are less important. In addition, higher muscular demand required for stair ascent may exacerbate the effects of weakness, making it more difficult to compensate and maintain normal kinematics. These reasonings may explain why kinematics were changed only during the stair ascent and not during level walking after paralysis of the GCM medial head.

4.3. Limitations

To our knowledge, this is the first study to explore the effects of discrete GCM medial head functional loss on level walking and stair ascent in subjects with tibial nerve ablation. We did not evaluate the impact of altered kinematics and EMG compensation on gait energy efficiency. In a future study, it might be necessary to investigate whether the differences in compensatory strategies affect energy efficiency during walking and stair ascent.

5. Conclusion

Our study, which investigated level walking and stair ascent, revealed that the human body differentially compensated for GCM medial head dysfunction and that the compensation depended on the type of locomotion.

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References

[1] H.A. Haxton, Absolute muscle force in the ankle flexors of man, J. Physiol. (Lond.) 103 (3) (1944) 267–273 http://sfx-82yum.hosted.exlibrisgroup.com/yum?sid=Entrez%3APubMed&id=pmid%3A16991644.
[2] B.L. Riemann, G.K. Limbaugh, J.D. Eitner, R.G. LeFavi, Medial and lateral gastrocnemius activation differences during heel-raise exercise with three different foot positions, J. Strength Cond. Res. 25 (3) (2011) 634–639 https://www.ncbi.nlm.nih.gov/pubmed/20581696.
[3] E.S. Park, E. Sim, D.-W. Rha, S. Jung, Architectural changes of the gastrocnemius muscle after botulinum toxin type a injection in children with cerebral palsy, Yonsei Med. J. 55 (5) (2014) 1406–1412 http://synapse.koreamed.org/DOIx.php?id=10.3349%2Fymj.2014.55.5.1406.
[4] T. Antonios, P.J. Adds, The medial and lateral bellies of gastrocnemius: A cadaveric and ultrasound investigation, Ori. Commu. 21 (1) (2008) 66–74, https://doi.org/10.1002/ca.20533.
[5] D.H. Sutherland, L. Cooper, D. Daniel, The role of the ankle plantar flexors in normal walking, J. Bone Joint Surg. Am. 62 (3) (1980) 354–363 http://sfx-82yum.hosted.exlibrisgroup.com/yum?sid=Entrez%3APubMed&id=pmid%3A7364808.
[6] H.J. Hermens, B. Freriks, C. Disselhorst-Klug, G. Rau, Development of recommendations for SEMG sensors and sensor placement procedures, J. Electromyogr. Kinesiol. 10 (5) (2000) 361–374 http://www.sciencedirect.com/science/article/pii/S1050641100000274.
[7] J. Romkes, A.K. Hell, R. Brunner, Changes in muscle activity in children with hemiplegic cerebral palsy while walking with and without ankle–foot orthoses, Gait Posture 24 (4) (2006) 467–474 http://www.sciencedirect.com/science/article/pii/S0966636205002584.
[8] G.A. Cavagna, N.C. Heglund, C.R. Taylor, Mechanical work in terrestrial locomotion: two basic mechanisms for minimizing energy expenditure, Am. J. Physiol. 233 (5) (1977) R243–R261 http://sfx-82yum.hosted.exlibrisgroup.com/yum?sid=Entrez%3APubMed&id=pmid%3A411381.
[9] C.R. Lee, C.T. Farley, Determinants of the center of mass trajectory in human walking and running, J. Exp. Biol. 201 (21) (1998) 2935–2944 http://sfx-82yum.hosted.exlibrisgroup.com/yum?sid=Entrez%3APubMed&id=pmid%3A9866878.
[10] A.D. Kuo, The six determinants of gait and the inverted pendulum analogy: a dynamic walking perspective, Hum. Mov. Sci. 26 (4) (2007) 617–656 http://sfx-82yum.hosted.exlibrisgroup.com/yum?sid=Entrez%3APubMed&id=pmid%3A17617481.
[11] J. Perry, S.T. k, J.R. Davids, Gait analysis: normal and pathological function, J. Sports Sci. Med. 12 (6) (1992) 815 https://journals.lww.com/pedorthopaedics/Fulltext/1992/11000/Gait_Analysis_Normal_and_Pathological_Function.23.aspx.
[12] A.D. Kuo, J.M. Donelan, Dynamic principles of gait and their clinical implications, Phys. Ther. 90 (2) (2010) 157–174 http://sfx-82yum.hosted.exlibrisgroup.com/yum?sid=Entrez%3APubMed&id=pmid%3A20023002.
[13] R.L. Waters, S. Mulroy, The energy expenditure of normal and pathologic gait, Gait Posture 9 (3) (1999) 207–231 http://sfx-82yum.hosted.exlibrisgroup.com/yum?sid=Entrez%3APubMed&id=pmid%3A10575082.
[14] S.R. Simon, R.A. Mann, J.L. Hagy, L.J. Larsen, Role of the posterior calf muscles in normal gait, J. Bone Joint Surg. Am. 60 (4) (1978) 465–472 http://sfx-82yum.hosted.exlibrisgroup.com/yum?sid=Entrez%3APubMed&id=pmid%3A670267.
[15] R. Jahnsen, L. Villien, G. Aamodt, J.K. Stanghelle, I. Holm, Musculoskeletal pain in adults with cerebral palsy compared with the general population, J. Rehabil. Med. 36 (2) (2004) 78–84 http://sfx-82yum.hosted.exlibrisgroup.com/yum?sid=Entrez%3APubMed&id=pmid%3A15180222.
[16] A. Opheim, R. Jahnsen, E. Olsson, J.K. Stanghelle, Walking function, pain, and fatigue in adults with cerebral palsy: a 7-year follow-up study, Dev. Med. Child Neurol. 51 (5) (2009) 381–388 http://sfx-82yum.hosted.exlibrisgroup.com/yum?sid=Entrez%3APubMed&id=pmid%3A19207296.
[17] J. Rose, J.G. Gamble, A. Burgos, J. Medeiros, W.L. Haskell, Energy expenditure index of walking for normal children and for children with cerebral palsy, Dev. Med. Child Neurol. 32 (4) (1990) 333–340 http://sfx-82yum.hosted.exlibrisgroup.com/yum?sid=Entrez%3APubMed&id=pmid%3A2332124.
[18] T.A. Wren, S. Rethlefsen, R.M. Kay, Prevalence of specific gait abnormalities in children with cerebral palsy: influence of cerebral palsy subtype, age, and previous surgery, J. Pediatr. Orthop. 25 (1) (2005) 79–83 http://sfx-82yum.hosted.exlibrisgroup.com/yum?sid=Entrez%3APubMed&id=pmid%3A15614065.
[19] J.L. Hicks, M.H. Schwartz, A.S. Arnold, S.L. Delp, Crouched postures reduce the capacity of muscles to extend the hip and knee during the single-limb stance phase of gait, J. Biomech. 41 (5) (2008) 960–967 http://sfx-82yum.hosted.exlibrisgroup.com/yum?sid=Entrez%3APubMed&id=pmid%3A18291404.
[20] K.M. Steele, A. Seth, J.L. Hicks, M.S. Schwartz, S.L. Delp, Muscle contributions to support and progression during single-limb stance in crouch gait, J. Biomech. 43 (11) (2010) 2099–2105 HTTP://sfx-82yum.hosted.exlibrisgroup.com/yum?sid=Entrez%3APubMed&id=pmid%3A20493489.
[21] T.P. Andriacchi, G.B. Andersson, R.W. Fermier, D. Stern, J.O. Galante, A study of lower-limb mechanics during stair-climbing, J. Bone Joint Surg. Am. 62 (5) (1980) 749–757 http://sfx-82yum.hosted.exlibrisgroup.com/yum?sid=Entrez%3APubMed&id=pmid%3A7391098.
[22] B.J. McFadyen, D.A. Winter, An integrated biomechanical analysis of normal stair ascent and descent, J. Biomech. 21 (9) (1988) 733–744 http://sfx-82yum.hosted.exlibrisgroup.com/yum?sid=Entrez%3APubMed&id=pmid%3A3182877.
[23] R. Riener, M. Rabuffetti, C. Frigo, Stair ascent and descent at different inclinations, Gait Posture 15 (1) (2002) 32–44 http://sfx-82yum.hosted.exlibrisgroup.com/yum?sid=Entrez%3APubMed&id=pmid%3A11809579.
[24] J.E. Zachazewski, P.O. Riley, D.E. Krebs, Biomechanical analysis of body mass transfer during stair ascent and descent of healthy subjects, J. Rehabil. Res. Dev. 30 (4) (1993) 412–422 http://sfx-82yum.hosted.exlibrisgroup.com/yum?sid=Entrez%3APubMed&id=pmid%3A8158557.