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Joint moment contributions to swing knee extension acceleration during gait in children with spastic hemiplegic cerebral palsy

Evan J. Goldberg [a], Philip S. Requejo [b], Eileen G. Fowler [a,*]

[a] Department of Orthopaedic Surgery, University of California at Los Angeles, 1000 Veteran Ave, 22-64 Rehabilitation Center, Los Angeles, CA 90095-1795, USA
[b] Rancho Los Amigos National Rehabilitation Center, Downey, CA 90242, USA

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ABSTRACT

Inadequate peak knee extension during the swing phase of gait is a major deficit in individuals with spastic cerebral palsy (CP). The biomechanical mechanisms responsible for knee extension have not been thoroughly examined in CP. The purpose of this study was to assess the contributions of joint moments and gravity to knee extension acceleration during swing in children with spastic hemiplegic CP. Six children with spastic hemiplegic CP were recruited (age=13.4 $\pm$ 4.8 years). Gait data were collected using an eight-camera system. Induced acceleration analysis was performed for each limb during swing. Average joint moment and gravity contributions to swing knee extension acceleration were calculated. Total swing and stance joint moment contributions were compared between the hemiplegic and non-hemiplegic limbs using paired *t*-tests ($p<0.05$). Swing limb joint moment contributions from the hemiplegic limb decelerated swing knee extension significantly more than those of the non-hemiplegic limb and resulted in significantly reduced knee extension acceleration. Total stance limb joint moment contributions were not statistically different. Swing limb joint moment contributions that decelerated knee extension appeared to be the primary cause of inadequate knee extension acceleration during swing. Stance limb muscle strength did not appear to be the limiting factor in achieving adequate knee extension in children with CP. Recent research has shown that the ability to extend the knee during swing is dependent on the selective voluntary motor control of the limb. Data from individual participants support this concept.

1. Introduction

Inadequate peak knee extension during the swing phase of gait is a major deficit in individuals with spastic cerebral palsy (CP) (Sutherland and Davids, 1993; Crenna, 1998; Tuzson et al., 2003), which may lead to a shorter stride length and decreased walking velocity. The mechanisms contributing to swing knee extension are not fully understood. Mochon and McMahon (1980) reported that knee motion during swing can be modeled as a multi-link pendulum that moves passively. In contrast, others have reported that muscles, such as the hamstrings and vasti, are active during terminal-swing (Perry, 1992), and metabolic energy is consumed (Doke et al., 2005). Studies of joint moments (Davis and Ounpuu, 2004) have described a small knee extensor moment immediately following toe-off as knee flexion is decelerated. This is followed by a knee flexor moment for the majority of swing, as motion of the knee reverses from flexion to extension. Davis and Ounpuu (2004) proposed that knee extension occurs when the swing hip extensors decelerate the forward rotation of the thigh. This action causes the proximal end of the shank to decelerate while momentum carries the distal shank forward, passively extending the knee. Resulting knee extension is then controlled by the eccentric activation of knee flexors.

Determining how muscle groups accelerate specific joints during movement is complex because muscles are capable of accelerating joints they do not span through intersegmental forces (Zajac and Gordon, 1989). Induced acceleration analysis (IAA) is a technique that quantifies joint accelerations produced by muscle groups (joint moments) using musculoskeletal modeling. This technique has been used extensively to determine the roles of muscle groups during gait in individuals without disability (Kepple et al., 1997a; Neptune et al., 2001, 2004; Siegel et al., 2004; Arnold et al., 2007a, 2007b). Arnold et al. (2007a, 2007b) used IAA to examine muscle contributions to terminal-swing knee extension in six individuals without disability and found that the swing knee was accelerated toward extension primarily by the stance hip muscles and swing vasti. These investigators suggested inadequate terminal-swing knee extension in CP may result from weakness of the stance hip muscles. The analysis becomes more complex when examining gait in CP

* Corresponding author. Tel.: +1 310 825 4028; fax: +1 310 825 5290.
E-mail address: efowler@mednet.ucla.edu (E.G. Fowler).

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as IAA results are dependent on the orientation of the body segments (Fregly and Zajac, 1996); therefore, contributions to knee extension acceleration (KEA) may differ due to altered gait kinematics. Previous studies have examined contributions to swing-phase knee flexion in CP (Goldberg et al., 2003, 2006); however, studies describing contributions to swing knee extension using IAA in CP could not be found.

Impaired neurological motor control is another important factor that must be considered when assessing gait in CP. Recently, a valid and reliable selective voluntary motor control (SVMC) assessment called Selective Control Assessment of the Lower Extremity (SCALE) was developed (Fowler et al., 2009). A linear relationship between SCALE scores and the interjoint coordination of the hip and knee during swing was found in children with spastic CP (Fowler and Goldberg, 2009). Children with good SVMC are more capable of moving out of synergy during the swing phase of gait (i.e., hip flexion with knee extension), while children with poor SVMC are constrained to move in synergy (i.e., simultaneous hip and knee flexion/extension). Since SVMC has been shown to be an important factor in achieving simultaneous knee extension and hip flexion, the mechanisms responsible for accelerating the knee into extension should be influenced by SVMC.

Siegel et al. (2004) used IAA on experimental data collected from five individuals without disability. They found that pairs of joint moments with opposing energetic effects are necessary to balance energy flow through the limbs suggesting that intralimb coordination is important in controlling energy flow throughout the lower extremities. As individuals with impaired SVMC lack normal interjoint coordination (Fowler and Goldberg, 2009), they may lack the ability to efficiently balance energy flow through their lower extremities. As a result, they may rely on alternative strategies to accelerate the knee into extension during swing.

The purpose of this study was to assess the contributions of lower extremity joint moments and gravity to KEA during the swing in children with CP. Since both stance and swing limbs contribute to the production of swing-phase knee extension, we focused on children with spastic hemiplegia. This allowed us to evaluate potential compensation mechanisms by the less involved limb and differences between limbs within the same individual.

2. Methods

2.1. Study population

Six children with spastic hemiplegic CP who were referred for gait analysis were recruited for this study (Table 1). All participants were able to able to walk independently without assistive devices (Gross Motor Function Classification System (GMFCS) Levels I and II) (Palisano et al., 1997). Informed consent and assent, approved by the Institutional Review Board Human Subject's Protection Committee of our institution, were obtained for all participants and their legal guardians. Four participants had no history of musculoskeletal surgery. One participant had a tendon-Achilles lengthening, and the other had a hamstring release and femoral derotational osteotomy. We did not include participants with substantial rotational deformities at the hip or knee as knee extension would not occur in the direction of walking. A previous derotational osteotomy was not cause for exclusion.

2.2. Selective voluntary motor control assessment

SVMC was assessed using the SCALE evaluation (Fowler et al., 2009). Specific isolated movement patterns at the hip, knee, ankle, subtalar and toe joints were tested bilaterally. Hip flexion/extension with the knee extended was tested in side lying, and the following tests were performed in sitting: knee extension/flexion; ankle dorsiflexion/plantar flexion with the knee extended; subtalar inversion/eversion; and toe flexion/extension. Participants were asked to move in a reciprocating pattern to a verbal cadence (e.g., "flex, extend, flex"). SVMC was graded as "Normal" (2 points), "Impaired" (1 point) or "Unable" (0 points) at each joint. An overall SCALE score was calculated for each limb by summing the points assigned to each joint for a maximum of 10 points per limb.

2.3. Gait analysis

Gait analysis was performed in the Kameron Gait and Motion Analysis Laboratory. An eight-camera system (Motion Analysis Corporation, Santa Rosa, CA, USA) sampling at 60 Hz was used to collect motion data. Two forceplates (Kistler Instrumentation Corporation, Amherst, NY, USA) were concealed in the walkway to record ground-reaction forces sampled at 1 kHz. Fifteen reflective markers were placed on the participants using a modified Helen Hayes marker set (Davis III et al., 1991). Participants wore shorts and walked barefoot at a self-selected pace. Each walked back and forth on a 25-foot walkway until at least 3 forceplate hits had been recorded per limb. Data were collected in Cortex 1.0 (Motion Analysis Corp., Santa Rosa, CA, USA). Data were smoothed using a Butterworth filter at 6 Hz (Kidder et al., 1996).

2.4. Induced acceleration analysis

One representative trial for each limb was selected and imported into Visual 3D Basic/RT (C-Motion, Inc., Germantown, MD, USA). To minimize speed-related differences between limbs (Arnold et al., 2007a), trials for both limbs were selected such that walking speeds were within $\pm 5\%$ of each other. Kinematic data from the selected trial was plotted against average kinematic data to verify that the selected trial kinematics were within one standard deviation of his/her average kinematics. The contributions of joint moments and gravity to swing knee acceleration were calculated using the IAA Module. Kinematic and kinetic data were calculated from experimental data. A biomechanical model described and validated by Kepple et al., (1997a) was used, which had seven segments: a combined head, arms and trunk segment, and right and left thighs, shanks and feet. The hip was modeled as a spherical joint allowing flexion/extension, abduction/adduction and internal/external rotation. The knee was a revolute joint allowing flexion/extension. The ankle was a universal joint allowing dorsi/plantar flexion and inversion/eversion. A constraint was placed on the foot that fixed it to the floor when the foot was flat and allowed the foot to rotate about the center of pressure when it was not flat.

IAA was performed as described by Kepple et al. (1997a). The model was configured at each frame according to the experimental data. All joint moments and gravity were set to zero, and one joint moment was entered into the model. The resulting swing limb knee acceleration was calculated for the input. The input moment was then set back to zero, and all other joint moments and gravity were

Table 1

Participant	Age (yr)	Height (cm)	Weight (kg)	Walking speed (cm/s)		GMFCS	Hemiplegic side	SCALE score		Synergy (%)
				Right trial	Left trial			Non-hemiplegic limb	Hemiplegic limb	
P1	20.1	167.5	54.5	81.6	78.7	I	Left	8	4	53.6
P2	9.5	142.9	45.9	89.7	89.1	II	Right	10	6	15.8
P3	9.7	127.6	27.7	88.8	90.2	I	Left	10	4	83.3
P4	11.9	152.0	59.9	114.9	115.2	I	Right	9	6	11.1
P5	18.8	170.0	42.1	88.6	89.4	II	Left	8	4	20.0
P6	10.0	135.5	35.5	134.2	131.9	II	Left	9	7	7.1
Mean	13.4	149.3	44.3	99.6	99.1	-	-	9.0	5.2	31.8
SD	4.8	17.1	11.9	20.4	20.1	-	-	0.9	1.3	30.2

GMFCS=Gross Motor Function Classification System (Palisano et al., 1997)
Synergy indicates the percent of the extension phase in which simultaneous flexor contributions from the hip and knee occur on the hemiplegic limb.

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sequentially entered into the model. IAA was conducted during swing (i.e., from toe-off to initial foot contact). On average, toe-off occurred at 58.7% of the gait cycle on the hemiplegic limb and 64.1% of the gait cycle on the non-hemiplegic limb. The final output provided the knee angular acceleration generated by each joint moment and gravity. Contributions were averaged during the period in which knee acceleration was positive (i.e., the extension phase) (Fig. 1), and total stance and swing moment contributions were compared between the hemiplegic and non-hemiplegic limbs using paired t-tests ($p<0.05$). For comparison purposes, average KEA and joint moment and gravity contributions to this acceleration for six individuals without disability are shown in Fig. 2A and B (age $=11.8\pm1.9$ years, height $=151.2\pm15.5$ cm, weight $=45.5\pm11.1$ kg, walking speed $=120.0\pm17.7$ cm/s).

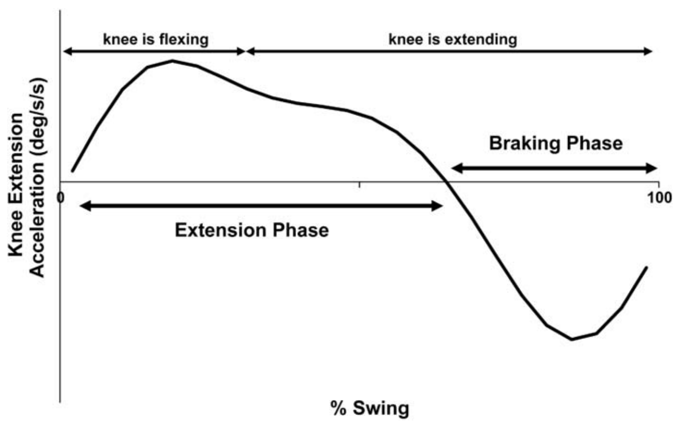

Fig. 1. Knee extension acceleration during the swing phase of gait for an individual without disability. The extension phase is defined as the period in which the knee acceleration is positive (extension), and the braking phase is defined as the period in which the knee acceleration is negative (flexion). The arrows above indicate when the knee is flexing and extending based on kinematic data.

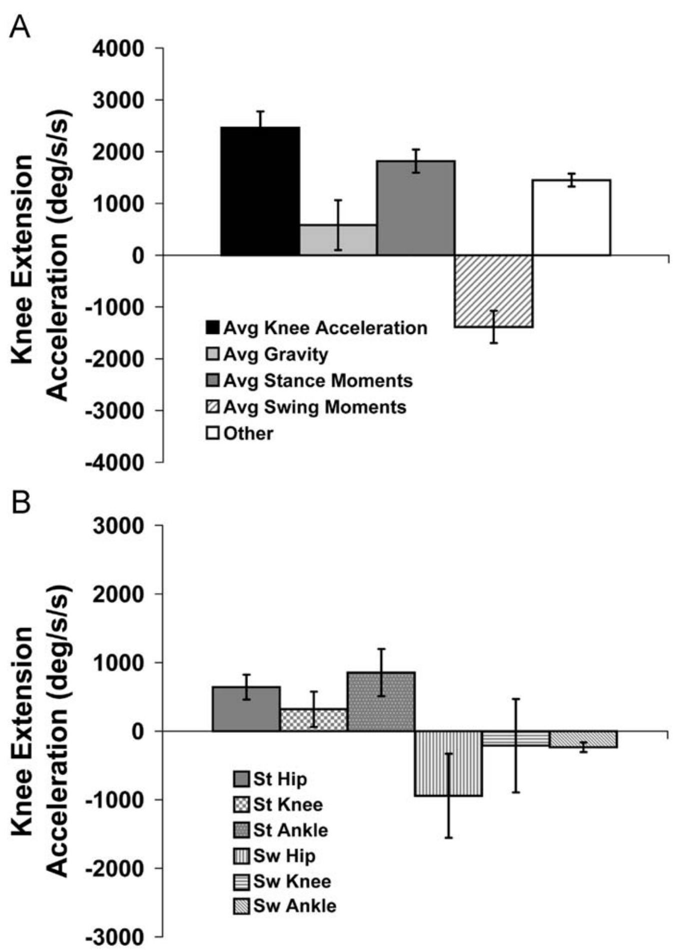

Fig. 2. (A) Average knee extension acceleration during the extension phase of swing and contributions to this acceleration from gravity, stance joint moments, swing joint moments and "other" sources and (B) average contributions to swing knee acceleration from the stance and swing hip, knee and ankle joint moments for six individuals without disability. "Other" sources were the velocity and constraint terms. Positive indicates knee extension.

3. Results

Average KEA values were greater in the non-hemiplegic limb for all participants except participant 6 (Fig. 3F, black bars). While stance limb contributions were similar between limbs, swing limb contributions from the hemiplegic limb decelerated swing knee extension significantly more than those of the non-hemiplegic limb ($p<0.05$) (Table 2) resulting in reduced KEA (Fig. 3, black bars). The contributions from the swing knee moment to swing KEA appeared to cause the overall difference in total swing moment contributions (Table 2), as negative contributions were substantially greater in the hemiplegic limb in all but one participant analyzed (Fig. 4E, horizontal-striped bars). On the stance limb, strategies varied by participant and limb (Fig. 4); however, total stance limb contributions only differed by $\sim$2% between limbs and were not statistically different (Table 2).

When comparing swing phase IAA results between the hemiplegic and non-hemiplegic limbs, the largest differences were found in the sagittal plane hip and knee moment contributions. The non-synergistic action of swing hip and knee moments is evident when assessing their contributions to swing KEA in the non-hemiplegic limb (Fig. 5B and E), and the pattern is similar to that of an individual without disability (Fig. 5A and D). Immediately following toe-off, an internal hip flexion moment decelerates knee extension while an internal knee extension moment accelerates the knee toward extension. Approximately halfway through the extension phase, the actions of these moments reverse. For the hemiplegic limb of the same participant (Participant 3, Fig. 5C and F), both the knee and hip moments decelerated knee extension for most of the extension phase and had a substantially larger magnitude of deceleration as compared to the non-hemiplegic limb. A similar pattern was observed for Participant 1 and to a lesser extent for Participants 2 and 5. Similar gait strategies were found between those who had orthopedic surgery and those who did not.

4. Discussion

The results of this study indicate that a significant difference was found between swing limb contributions to swing knee extension deceleration between the two limbs, with a greater average extensor deceleration provided by the hemiplegic limb. These excessive negative (or flexor) contributions may be caused by impaired SVMC, spasticity and/or static contractures. The largest differences between limbs among the swing joint moment contributions were the knee flexor/extensor moments in which the hemiplegic limb knee moment, on average, decelerated knee extension substantially more than that of the non-hemiplegic limb. This excessive knee flexor moment contribution may be attributed to premature hamstring activity due to an inappropriate hip and knee flexor synergy pattern or spasticity. Previous electromyography studies have found inappropriate activity of the knee flexors during early swing on the hemiplegic limb in individuals with CP (Patikas et al., 2005).

Since the ability to produce non-synergistic movement at the hip and the knee during swing (i.e., hip flexion with knee

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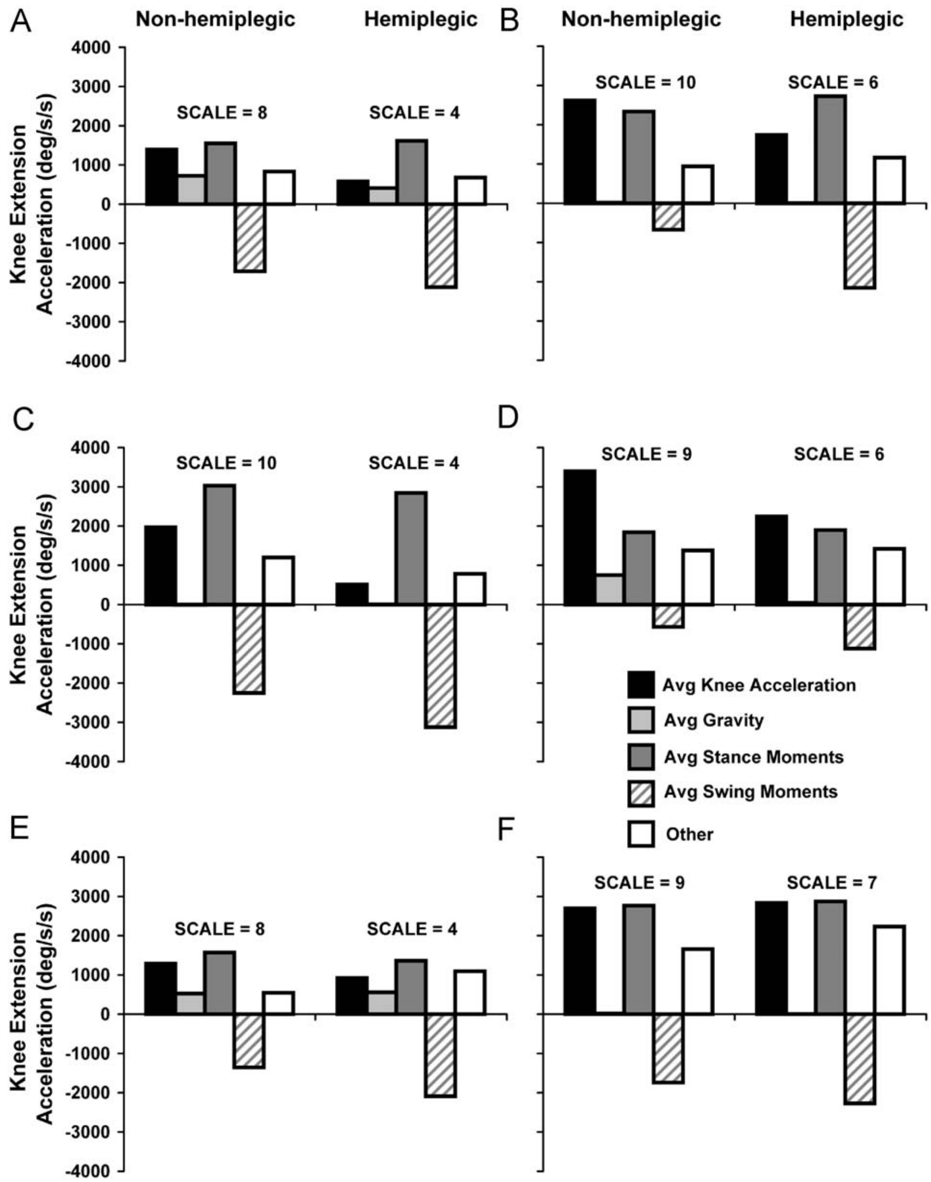

Fig. 3. Average knee extension acceleration and average contributions to swing knee acceleration from gravity, the stance limb joint moments and the swing limb joint moments during the extension phase of swing for participants 1–6 (A–F). "Other" indicates the velocity and constraint terms. Positive indicates extension.

Table 2

	Non-hemiplegic	Hemiplegic	Mean difference
Average swing knee extension acceleration (in deg/s/s)	2226 ± 823	1471 ± 954	$756 \pm 571^{*}$
Average contribution from:			
Gravity	336 ± 369	170 ± 246	166 ± 294
Stance joint moments	2183 ± 628	2219 ± 675	-37 ± 220
Swing joint moments	-1384 ± 658	-2147 ± 637	$763 \pm 389^{*}$

Note: Stance limb contributions to the hemiplegic limb knee acceleration are contributions from the non-hemiplegic limb, and stance limb contributions to the non-hemiplegic limb knee acceleration are contributions from the hemiplegic limb. Positive indicates extension.

* $p < 0.05$

extension) has been shown to be related to SVMC in participants with spastic diplegic CP (Fowler and Goldberg, 2009), the degree of SVMC impairment may also play a role in the joint moment contributions to swing KEA. Hemiplegic limbs that exhibited simultaneous flexor contributions from swing limb sagittal plane hip and knee moments for over 50% of the extension phase had low SCALE scores (4). The two participants with the shortest duration of simultaneous flexor contributions from the hip and knee (7% and 11%) had higher SCALE scores (6 and 7, respectively). Although the sample size is small, the data suggests that the

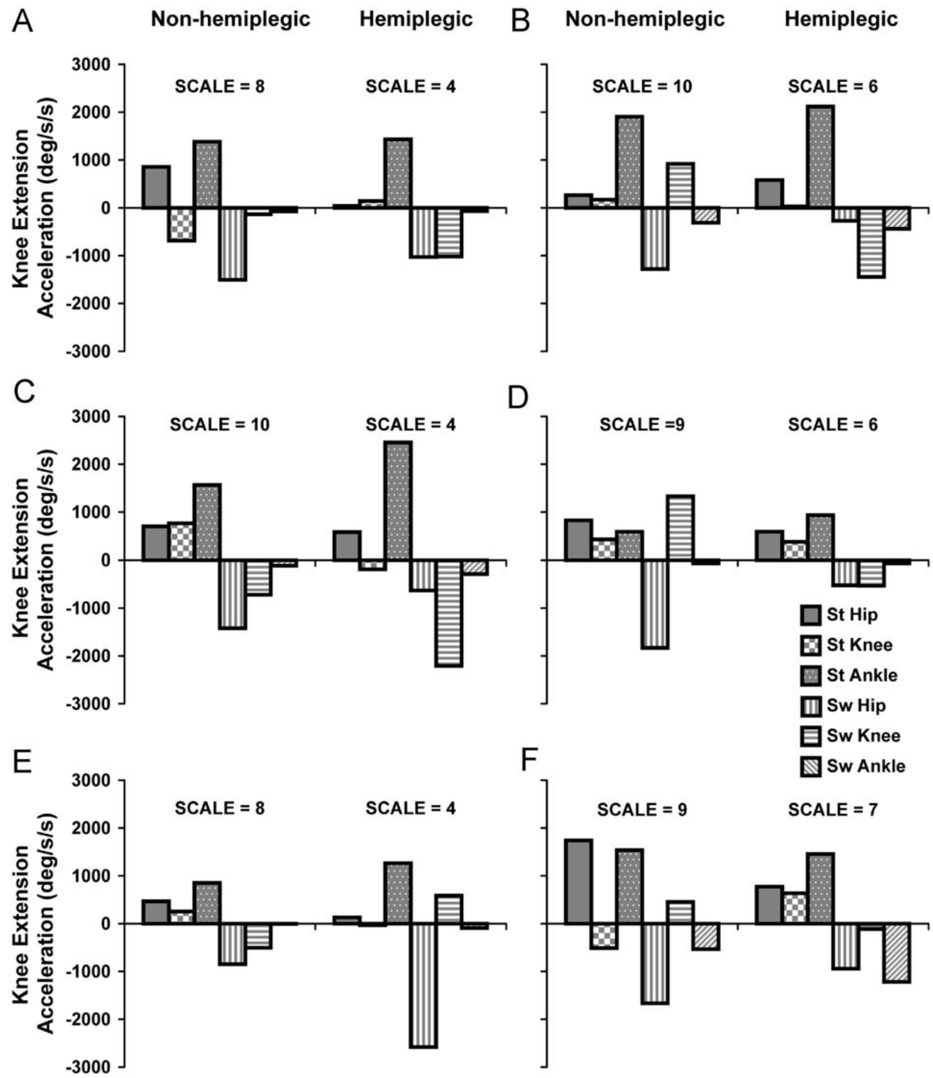

Fig. 4. Average contributions to swing knee extension acceleration from the stance and swing hip, knee and ankle joint moments during the extension phase of swing for participants 1–6 (A–F). Positive indicates extension.

amount of time in which simultaneous hip and knee flexor contributions occurs during the extension phase may be related to the SVMC of the limb.

While previous studies suggest that inadequate terminal-swing knee extension in children with CP may be caused by stance limb muscle weakness (Arnold et al., 2007b), our data demonstrate that stance limb contributions to swing KEA were similar for the hemiplegic and non-hemiplegic limbs. These results suggest that the hemiplegic and non-hemiplegic limb stance joint moments are similar in their ability to accelerate the swing knee toward extension despite impaired SVMC. During this phase of gait (from the beginning of single limb support through mid-stance), simultaneous hip and knee extension normally occurs and can be accomplished by individuals with CP who use synergistic movement patterns. Therefore, a high level of SVMC is not necessary for the stance limb to accelerate the swing knee toward extension. Siegel et al. (2004) suggested that lower extremity extensor moments tend to add energy to the trunk, and Arnold et al. (2007a, 2007b) found that stance limb muscles that accelerate the pelvis upward create reaction forces on the swing limb that accelerate the knee toward extension. Two participants generated stance hip and knee moments out of synergy on the hemiplegic limb (i.e., hip extension with knee flexion). In both cases, the stance knee flexion moment accelerated the swing knee toward flexion. Both participants walked with increased stance limb hip extension moments, which accelerated the swing knee toward extension (Fig. 4A and F, solid gray bars) and appeared to compensate for the negative knee moment contributions.

One limitation of the study is the small sample size (n=6). Despite this sample size, important differences were found between the mechanisms responsible for swing KEA in the hemiplegic and non-hemiplegic limbs. Joint moment-induced accelerations only represent a "snapshot in time" of the contributions to lower extremity joint accelerations, and do not represent cumulative effects of past forces acting on the body, such as the Coriolis, centripetal and constraint forces (Zajac et al., 2002). Therefore, we did not explicitly calculate the component of the swing KEA due to the passive dynamics of the limb or the model constraints. The difference between the actual KEA and the joint moment contributions to the KEA and gravity represent the summed contributions from velocity terms and model constraints.

The present study used joint moments as inputs to the IAA. The primary disadvantage in using joint moments instead of muscle forces is that they provide limited insight into the actions of

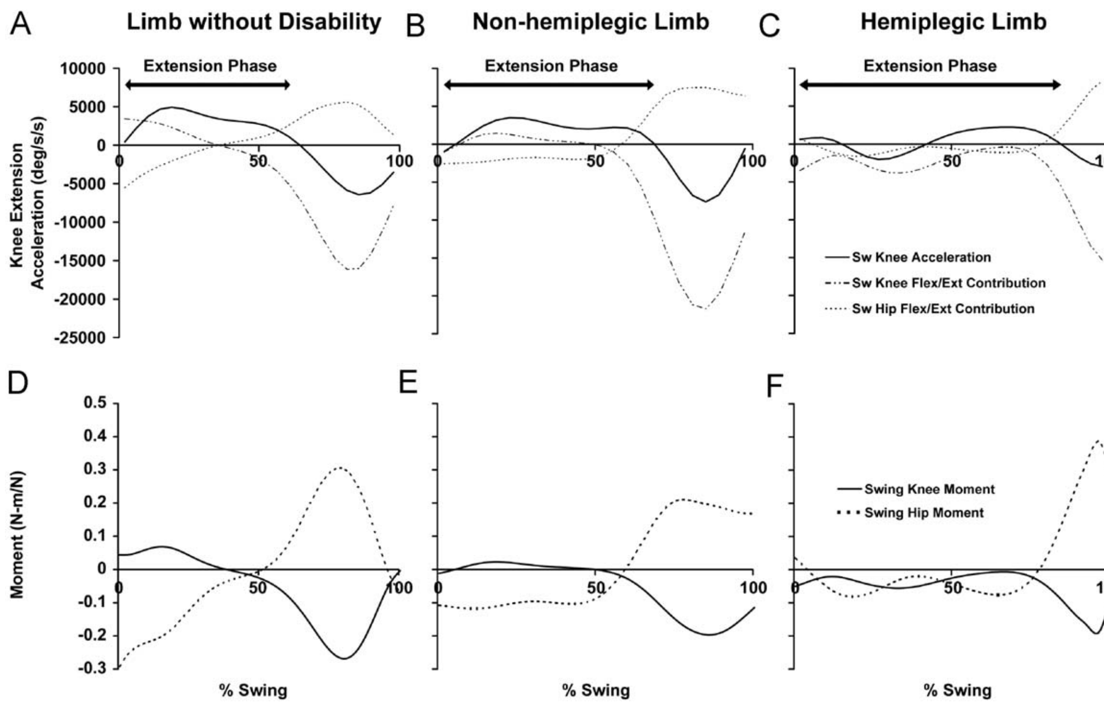

Fig. 5. Swing hip and knee flexion/extension moment contributions to swing knee extension acceleration during the swing phase of gait for (A) an individual without disability and one participant with hemiplegic CP (Participant 3) for the (B) non-hemiplegic limb and the (C) hemiplegic limb and swing hip and knee flexion/extension moments for (D) an individual without disability and one participant with hemiplegic CP for the (E) non-hemiplegic limb and the (F) hemiplegic limb. Positive indicates extension.

biarticular muscles (e.g., hamstrings), and we can only speculate how individual muscles accelerate the knee. While many studies have used muscle-driven models in IAA and segmental power analysis to assess muscle function in individuals without disability (e.g., Neptune et al., 2001, 2004; Arnold et al., 2007a, 2007b), few studies have used muscle-driven models for induced acceleration in CP. Limitations in the current models, including the accuracy and efficiency of characterizing musculoskeletal geometry in CP and changes in muscle-tendon mechanics due to pathology and surgery, must be addressed before muscle-driven models are used in IAA in CP (Arnold and Delp, 2005).

The foot constraint used in the model produced a discontinuity in the data at heel-off. This foot model was compared to a visco-elastic foot–floor model in previous studies, and IAA results were similar (Kepple et al., 2002). In all three participants in which the constraint was applied, both feet had a period in which the foot was flat, so the constraint was applied bilaterally. Therefore, differences between limbs in each participant were not due to the use of the foot constraint. When the constraint was not applied and the foot was allowed to rotate around the center of pressure, stance ankle moment contributions may have been overestimated while stance knee moment contributions may have been under-estimated (Kepple et al., 1997b, 2002). Regardless of the model used, the total stance contribution would most likely remain unchanged. The overestimation of the ankle plantar flexors on the stance limb may explain the discrepancy between our induced acceleration results in individuals without disability (Fig. 2B) and previous studies (Arnold et al., 2007a, 2007b), which modeled the foot/floor interaction with spring-damper units.

In summary, swing limb joint moment contributions that decelerate knee extension appear to be the primary cause of inadequate KEA during swing, specifically the contributions from the swing knee moment. These findings show the need to assess these contributions in diplegic participants across the spectrum of SCALE scores to see if there is a relationship between SVMC and the mechanisms that accelerate the knee toward extension during swing. It is possible that excessive contributions during the braking phase also contribute to inadequate terminal-swing knee extension, and contributions during this phase should be assessed in future studies. Since no significant difference was found between total stance joint moment contributions from the hemiplegic and non-hemiplegic limbs to swing KEA, stance limb muscle strength does not appear to be the limiting factor in achieving appropriate knee extension in spastic CP. In patients who have sufficient SVMC treating the spasticity of the swing limb may improve swing knee extension by decreasing these negative contributions.

Conflict of interest statement

There is no conflict of interest.

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