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Muscle adaptation patterns of children with a trans-tibial amputation during walking

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Abstract

Background. Many studies have shown that trans-tibial amputation involves modifications of resultant muscle patterns during gait. However, these experiments did not estimate the contribution of simultaneous agonist and antagonist muscle action (co-contraction) during gait tasks. Diminution of co-contraction could create joint instability and, thus, change joint integrity, which is particularly important in the etiology of degenerative diseases, such as osteoarthritis, present at the knees of amputated limbs, and particularly in non-amputated limbs. The purpose of this study was to determine if there is any difference in the production of co-contraction about the knee between able-bodied children and children with a trans-tibial amputation during gait.

Methods. Six children with a trans-tibial amputation vs. six able-bodied children paired for gender, age, weight and height participated in this study. Four one-way ANOVAs ($P < 0.05$) were used to observe differences in resultant, agonist and antagonist moments, power, and co-contraction index during different phases of gait between able-bodied children limbs, the amputated and the non-amputated limbs of children with trans-tibial amputation.

Findings. Children with a trans-tibial amputation modified muscle patterns at their amputated limb and produced smaller co-contraction ($P < 0.05$) during single limb support, for both the non-amputated and amputated limbs when compared to able-bodied children.

Interpretation. These results suggest that children with a trans-tibial amputation altered their muscle patterns to perform locomotion. These changes produced a diminution of co-contraction during single limb support for both the amputated and non-amputated limbs and, thus, could create joint instability.

Keywords: Trans-tibial amputation; Children; Agonist muscles; Antagonist muscles; Co-contraction; Gait; Muscle patterns

1. Introduction

Human locomotion involves complex multi-joint and multi-muscle coordination arising because of the redundancy inherent in the musculoskeletal system. Especially during walking, able-bodied (AB) subjects coordinate motion in the lower limbs at the hip, knee and ankle joints for safer ambulation in their environment (Winter, 1991).

For subjects with a trans-tibial amputation (TTA), loss of the ankle and foot, which is responsible for 80% of the propulsion in normal gait (Winter, 1991), requires changes in hip and knee joint coordination to perform efficient gait (Winter and Sienko, 1988; Sanderson and Martin, 1997; Powers et al., 1998; Sadeghi et al., 2001). Particularly during weight-bearing, Sanderson and Martin (1997) observed significant changes in the resultant knee joint moment between able-bodied adults and six adults with a TTA

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(32.8 (SD 6.7) years). The moment pattern showed a transition from extensor to flexor in the control limb (CL) of AB adults and in the non-amputated limb (NAL) of adults with a TTA whereas it remained flexor in the amputated limb (AL) of adults with a TTA. Furthermore, Winter and Sienko (1988) noted a very low or near zero resultant moment in elderly subjects with a TTA for the first half of stance and a normal knee flexor moment for the rest of the stance phase. In contrast, Powers et al. (1998) recorded a knee extensor moment throughout the stance phase in 10 males with a TTA (62.3 (SD 6.9) years). In summary, the findings of the previously mentioned authors were not consistent with one another regarding knee joint kinetics.

On the other hand, it has been demonstrated that the knee joint is commonly affected in joint pathologies associated with TTA (Burke et al., 1978; Melzer et al., 2001). Joint instability (or joint laxity) is a major factor in the etiology of degenerative diseases (Felson et al., 2000; Issa and Sharma, 2006) such as osteoarthritis seen at the knee of the NAL and AL of subjects with a TTA (Burke et al., 1978; Melzer et al., 2001). Articular instability could be attributed to changes in co-contraction (i.e., co-activation of the agonist and antagonist muscles), creating additional stresses on the internal structures of the joints (Arms et al., 1984; Solomonow et al., 1989; Miller et al., 2000). More precisely for AB adults during walking, Falconer and Winter (1985) found a higher percentage of co-contraction during weight acceptance compared to the other phases of the gait cycle. The higher production of co-contraction during weight acceptance is not surprising, given that this phase requires augmented knee joint stability. Hence, the development of premature osteoarthritis at the knees of subjects with a TTA could result from modifications in the activity of the agonist and antagonist muscle groups associated with changes observed in net joint kinetics. Indeed, the human system has more muscles acting around each joint than theoretically needed for coordinated motion, and there are multiple possibilities of muscle coordination to perform a given movement (Prilutsky and Zatsiorsky, 2002). The premature development of osteoarthritis observed in both AL and NAL of subjects with a TTA could be associated with long-term effects of changes induced in muscle patterns following amputation. Hence it could be of special clinical interest to study the modifications in muscle patterns during locomotion as early as childhood in subjects with a TTA.

The purpose of the present study was to investigate differences in the production of co-contraction about the knee between AB children and children with a TTA. To gain insight into how children with a TTA adapted the production of co-contraction during gait, we quantified the resultant, flexor and extensor moments developed at the knee joint in AB children and in the NAL and the AL of children with a TTA. Based on kinematic, kinetic and EMG data, co-contraction was estimated with an updated version of the EMG-assisted optimization model proposed by Amarantini and Martin (2004). The present experiment tested the hypothesis that children with a TTA have lower production of co-contraction at the knee than AB children because of changes in both agonist and antagonist muscle group moments. For subjects with a TTA, such changes could affect joint stability and, thus, might predispose subjects with an amputation to the development of osteoarthritis in both their AL and NAL.

2. Methods

2.1. Subjects

Six children with a TTA (4 males, 2 females) with age 11 (SD 5) years, height 153.5 (SD 16.1) cm and body mass 53.4 (SD 21.3) kg were selected from the Musculoskeletal Clinic of the Centre de Réadaptation Marie Enfant of Hôpital Sainte-Justine. Four children had congenital limb deficiencies while the other two underwent amputations as a result of meningococcemia disease. A certified prosthetist of the Centre de Réadaptation Marie Enfant conducted individual evaluations to ensure that the lower limb prosthesis performed normally, that each subject was comfortable using it and performed normally with his or her prosthesis. To provide some control over the effect of prosthetic design on gait, children with a TTA using Seattle-Light foot prostheses were recruited. Additionally, amputee recruitment focused on children who were fully ambulatory.

Six age-, height-, mass- and sex-matched AB children (4 males, 2 females; age 12 (SD 4) years; height 1.56 (SD 0.17) m; body mass 55.5 (SD 13.9) kg) with no known musculoskeletal problems that could affect their ability to perform gait participated in this study.

All the children were physically active and in good health. The procedures for this study were approved by the Research Ethics Committee of Hôpital Sainte-Justine, and all parents gave informed, written consent.

2.2. Experimental design

Each subject was instructed to walk along a 10-m walkway at a self-selected speed. Shoes were worn during testing. Three to five practice trials were allowed to familiarize the subjects with the testing environment. Three acceptable trials were recorded for each limb where the subjects were instructed to put each foot on a force platform while walking with fluidity. The equipment employed in this study comprised eight 3D digital cameras (Vicon Peak, CA, USA), two AMTI force platforms (Advance Mechanical Technology Inc., MA, USA) and eight double-differential pre-amplifier EMG electrodes from a multi-channel EMG system (Model MA-300-16, Motion Lab Systems, Inc., LA, USA).

The 2D kinematic data were acquired at 60 Hz. Twenty-four passive, reflective markers were placed bilaterally on anatomical landmarks (front and back of head, acromion, elbow, wrist, anterior superior iliac spines, thigh, knee, shank, ankle, heel and toe) and one on the sacrum. Raw coordinates were filtered with a zero-lag, low-pass Butterworth filter (4th order, 6 Hz cut-off frequency) before computing joint angular and linear displacements. At each joint, angular velocity and acceleration were calculated by differentiating cubic smoothing splines (De Boor, 2004). Ground reaction forces and moments were recorded at 900 Hz and were filtered with a zero-lag 4th order Butterworth filter with a 9 Hz low-pass cut-off frequency. To collect muscle activity for AB children, EMG electrodes were fixed bilaterally on the rectus femoris, vastus medialis, medial hamstring (MH) and gastrocnemius (GA). For children with a TTA, the EMG electrodes were applied on the same muscles, without the GA electrode for the AL. Thus, only the MH was taken as the representative knee flexor muscle for the AL. The skin was cleaned with alcohol and, when necessary, body hair was shaved. The electrodes were oriented parallel to muscle fibre direction and were positioned on muscle belly. The EMG signals were sampled at 900 Hz, bandpass-filtered (zero-lag, 4th order, 30–300 Hz cut-off frequencies), and then full wave-rectified.

2.3. Gait variables

The vector of resultant moments was calculated at the knee with inverse dynamics, according to Lagrangian formalism (Zajac and Gordon, 1989; Amarantini and Martin, 2004). For sign convention, the moment was positive for knee extension. Net joint power was computed at each joint as the scalar product of angular velocity and net joint moment. Concentric contraction (energy generation) was represented by positive power, while eccentric contraction (energy absorption) was represented by negative power.

Agonist and antagonist moments were estimated with an updated version of the EMG-assisted optimization model proposed by Amarantini and Martin (2004). For the present study, the estimation of the coefficients α establishing the isometric moment-EMG relationships was directly incorporated into the routine used for dynamic conditions. Thus, the *isometric calibration* step was appropriately removed from the experimental design without affecting the results. This improvement is an obvious advantage for clinical application since it reduces the complexity of the experimental setup. The low-pass cut-off frequency and the exponent applied to the EMG signals were respectively set to 2.5 Hz and 1 for all muscles according to the recommendations of Amarantini and Martin (2004). Consequently, the optimization problem was rewritten as:

$$\text{find}: \alpha = \{\alpha_{\mathrm{RF}}, \alpha_{\mathrm{VM}}, \alpha_{\mathrm{MH}}, \alpha_{\mathrm{GA}}\}, \quad \beta = \{\beta_h, \beta_k, \beta_a\}, \quad \delta = \{\delta_h, \delta_k, \delta_a\} \text{ and } w = \{w_{\mathrm{RF}}, w_{\mathrm{VM}}, w_{\mathrm{MH}}, w_{\mathrm{GA}}\}$$

$$\text{that minimize:} \quad C = \frac{1}{2} \cdot \sum_t (M_K(t) - \widehat{M}_K(t))^2 \tag{1}$$

α, β, δ and w

with

$$\widehat{M}_k(t) = w_{\mathrm{ext}}(t) \cdot \widehat{M}_{\mathrm{ext}}(t) + w_{\mathrm{flex}}(t) \cdot \widehat{M}_{\mathrm{flex}}(t) \tag{2}$$

$$\widehat{M}_{K_i}(t) = [\alpha_i \cdot r\mathrm{EMG}_i(t)] \cdot [1 \pm E \cdot (\beta \cdot \Delta\theta) \pm E \cdot (\delta \cdot \dot{\theta})], \quad i = \{\mathrm{RF,VM,MH,GA}\} \tag{3}$$

$$\text{subject to:} \quad \begin{cases} \alpha_{\mathrm{MH}} \text{ and } \alpha_{\mathrm{GA}} < 0, \quad \alpha_{\mathrm{RF}} \text{ and } \alpha_{\mathrm{VM}} > 0 \\ \beta > 0 \text{ and } \delta > 0 \\ \widehat{M}_{\mathrm{flex}} < 0 \text{ and } \widehat{M}_{\mathrm{ext}} > 0 \\ 0 < w < 1 \end{cases} \tag{4}$$

where rEMG is the vector of EMG signals processed using filter characteristics, and α is the matrix of coefficients establishing EMG-moment relationships. Variable w is the matrix of muscle group gains, and E is the matrix of biarticularity. Unlike the model of Amarantini and Martin (2004), w was applied to the extensor (w_{ext}) and flexor (w_{flex}) muscular groups to gain processing times. This change is possible, given that the model accounts for the total force of each muscular group. β and δ are matrices of stiffness and viscosity coefficients depending respectively upon angular changes ($\theta_j - \theta_{j\mathrm{Tiso}}$) and angular velocity ($\dot{\theta}_j$) vectors. $\Delta\theta$ was defined as the difference between the vector of angular displacement (θ) and that of theoretical isometric calibration angles ($\theta_{\mathrm{Tiso}} = \langle\theta_{h\mathrm{Tiso}}, \theta_{k\mathrm{Tiso}}, \theta_{a\mathrm{Tiso}}\rangle^{\mathrm{t}}$), calculated as half of the excursion of movement at each joint (van Dieen and Visser, 1999).

The above constrained nonlinear optimization problem was solved by sequential quadratic programming (Boggs and Tolle, 1996) with w, $|\alpha|$, β and δ initially set to 0.5, 50, 5 and 5, respectively. Flexor and extensor moments were used to compute the co-contraction index (CI) at the knee at each time t according to the expression given by Falconer and Winter (1985):

$$\mathrm{CI} = \left(\frac{2 \cdot |M_{\mathrm{antago}}|}{|M_{\mathrm{ago}}| + |M_{\mathrm{antago}}|}\right) \times 100\% \tag{5}$$

where M_{ago} is the knee agonist moment, and M_{antago} is the knee antagonist moment at each time t.

2.4. Statistics

The present study focuses principally on the knee because joint kinetic differences between AB and subjects with a TTA are most apparent at this joint (Sanderson and Martin, 1997), and major musculoskeletal problems occur at the knee joint for subjects with a TTA (Burke et al., 1978; Melzer et al., 2001).

After the calculations, all biomechanical patterns were normalized relative to time, and joint kinetics were normalized by the subjects' mass. For each subject, mean sagittal kinematic and kinetic profiles were obtained by averaging three gait cycles.

For comparisons, the dependent variables were the knee CI (expressed in %), the averaged peaks of knee angular positions, of the net, agonist and antagonist knee joint moments and of the knee joint power (K1–K4). Each variable was compared during weight acceptance, single limb support, push-off, mid-swing and end of swing period. After having verified the absence of difference between the right and left legs of AB children with one-way repeated measures ANOVA, the data from the left and right sides of each AB subject were averaged to form one set of "control" data called "CL". One-way ANOVAs were then performed on each of the dependent variables to compare AL versus NAL (repeated measures ANOVA), CL versus AL and CL versus NAL (independent ANOVAs). Significant difference was set at $P < 0.05$.

3. Results

3.1. Kinematics

Step length, swing time, stance time and single limb support time were not statistically different between CL, AL and NAL (Table 1). Furthermore, no significant difference was observed for the stride length (Table 1), and for the averaged speed and cadence. Speed and cadence values were respectively 1.12 m/s (SD 0.17) and 113.1 steps/min (SD 12.5) for AB children compared to 1.04 m/s (SD 0.14) and 111.2 steps/min (SD 12.6) for children with a TTA.

No significant differences were observed for peak knee angular excursion values during the support and swing periods (see Fig. 1). This invariance, obtained from the kinematics data, showed high consistency in the behaviour of AB and children with a TTA at the knee during gait.

3.2. Muscular strategies

The updated version of the EMG-assisted optimization model proposed by this study gave an accurate estimation of the resultant moment when compared to the resultant moment obtained by the inverse dynamic calculated with Lagrangian formalism. Indeed, we found a coefficient of determination of 0.97 for AB children and of 0.93 for children with a TTA. Thus, for the rest of this paper, the resultant moment estimated from the EMG signals has been taken to compare the three limbs.

Moments were plotted during the gait cycle, with heel contact representing 0%, toe-off 65%, and just before subsequent heel contact 100% of the gait cycle. As shown in Fig. 1, the CL produced a positive (extensor) moment for the first 0–12% of the gait cycle, as the quadriceps acted eccentrically (negative power K1) to control knee flexion and thus using the quadriceps for agonist muscles. From 12% to 30% of the gait cycle, the knee carried out extension with concentric contraction of the agonist quadriceps muscles (positive power K2) to raise the centre of the mass. Then, the resultant moment changed polarity (flexor) from 30% to 50% of the gait cycle as the agonist hamstring muscles acted eccentrically (negative power) to slow forward progression of the body. From 50% to 75%, the end of the stance phase to the beginning of the swing phase, the resultant moment was extensor, and the agonist quadriceps muscles acted eccentrically (negative power K3) to control knee flexion during push-off, and continued during early swing to decelerate the backward swinging leg. Finally, for the rest of the gait cycle (75–100%), the resultant moment was negative (flexor), and the agonist hamstring muscles acted eccentrically as the hamstring controlled knee extension prior to subsequent heel contact (negative power K4). The NAL showed the same pattern (see Fig. 1).

As illustrated in Fig. 1, the AL did not perform the same muscular pattern during the stance phase compared to the CL and NAL. The AL had a positive resultant moment (extensor) for all of the stance phase compared to the transition (extensor to flexor) observed for the CL and NAL. Indeed, significant differences were apparent for the minimum resultant moment value during stance where CL (−0.21 (SD 0.11) N m/kg) and NAL (−0.32 (SD 0.31) N m/kg) were flexor while AL (0.30 (SD 0.27) N m/kg) was extensor ($P < 0.05$). That is to say, children with a TTA used their extensors as agonist muscles for all of the stance phase. Finally, no significant

Table 1
Spatio-temporal parameters (mean and standard deviation (SD)) for the control limb (CL) in AB children and for the non-amputated (NAL) and the amputated (AL) limbs in children with a TTA

	AB children		Children with a TTA	
	Mean	SD	Mean	SD
Stride length (m)	1.18	0.15	1.15	0.18

	AB children		Children with a TTA			
	CL		NAL		AL	
	Mean	SD	Mean	SD	Mean	SD
Step length (m)	0.60	0.08	0.54	0.10	0.56	0.10
Stance time (%)	65.2	1.2	65.9	0.9	65.3	0.9
Swing time (%)	34.8	1.2	34.1	1.0	34.7	0.9
Single limb support (%)	11–50	2	12–50	3	15–51	3

difference was evident for peak values of the resultant, extensor and flexor moments during the stance and swing phases. Similar results were obtained for the peak muscular power values of K1–K4, where no significant difference was found for the stance and swing phases (see Fig. 1).

3.3. Co-contraction

For the CI, no significant difference was discerned for weight acceptance and change of direction of knee angular displacement. During the single limb support, a significant difference was observed for the CI. Indeed, the CI for the

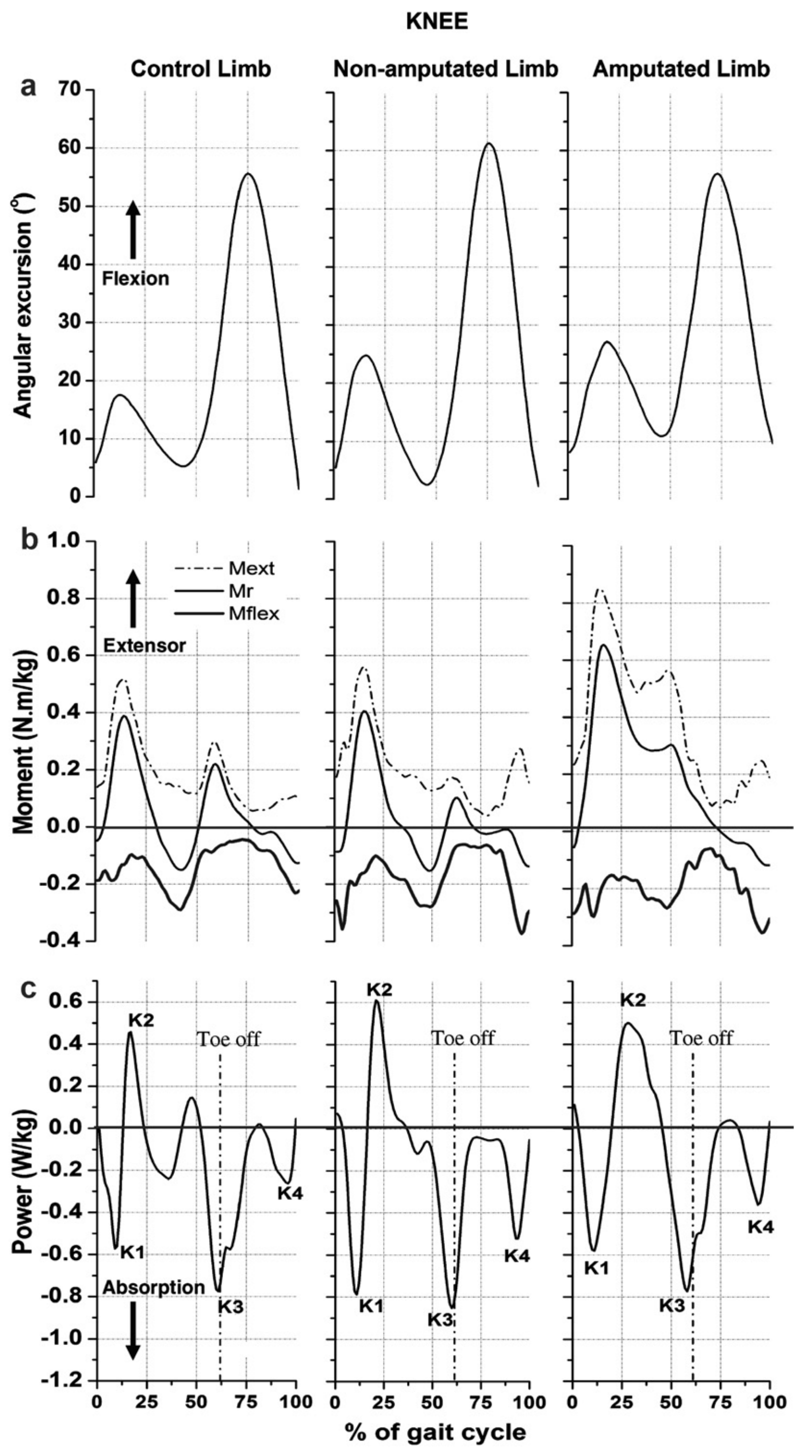

Fig. 1. (a) Knee angular displacement. (b) Resultant (M_r), extensor (M_{ext}) and flexor (M_{flex}) knee moments in N m/kg for the entire gait cycle of the amputated limb, non-amputated limb and control limb. (c) Knee power in W/kg for the entire gait cycle of the three different limbs. The data are from two representative subjects.

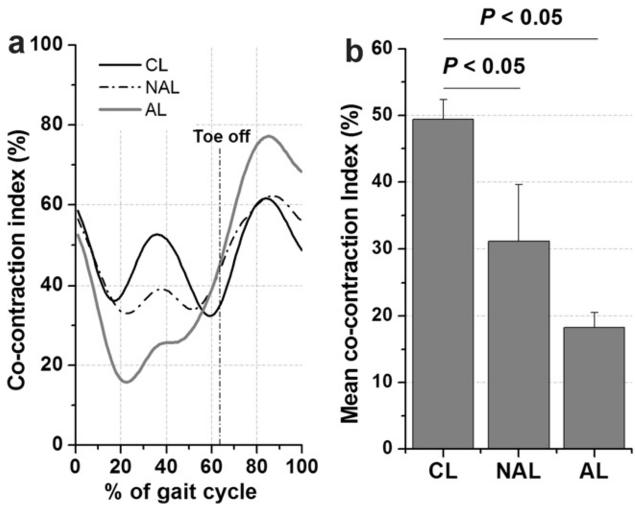

Fig. 2. (a) Co-contraction index for the gait cycle of the amputated limb (AL), non-amputated limb (NAL) and control limb (CL). (b) Mean values of the co-contraction index during the single limb support phase for the three different limbs.

CL (49 (SD 7.3) %) was greater ($P < 0.05$) compared to the NAL (31.1 (SD 20.8) %) and AL (18.2 (SD 5.6) %) (see Fig. 2).

4. Discussion

4.1. Kinematics

As reported by Sanderson and Martin (1997), the present study did not find differences for the spatio-temporal variables (stride length, cadence, speed and percentage of support, and swing phases) for the entire gait cycle.

Considering kinematics, the knee joint reflected considerable similarity between the three limbs for the entire gait cycle as detected previously for amputee gait (Sanderson and Martin, 1997; Lewallen et al., 1986) and for amputee gait obstacle avoidance (Hill et al., 1997). These results reveal at which point the behaviour of children with a TTA is comparable to that of AB children for knee kinematics.

4.2. Muscle strategies

During the stance phase of the gait cycle, maintenance of upright posture and forward propulsion of the body are assured by the lower extremities. These functions can be accomplished with various combinations of moments of force about the three joints of the lower extremity during the entire task (Prilutsky and Zatsiorsky, 2002).

The present study identified differences in knee muscle patterns between AB and children with a TTA during gait. Indeed, at their AL, children with a TTA used their extensor muscle as agonist for all of the support phase compared to their NAL and AB children who had a change of agonist muscle during the stance phase (extensor to flexor). This finding goes along with the observation of Powers et al. (1998) who noted that children with a TTA had an extensor resultant moment for all of the stance phase of the gait cycle. For their AL, children with a TTA used their extensor muscle to keep the knee in extension compared to the CL where the agonist flexor muscle was deployed eccentrically to slow forward progression of the body. For the NAL, children with a TTA used the same muscle coordination as AB children during the stance phase. By employing muscle redundancy to compensate for their loss of ankle and foot and their knee muscular atrophy (Isakov et al., 2000), children with a TTA can perform the same movement by modifying the efforts of the other muscles. However, this new pattern creates changes on the level of muscular co-contractions.

4.3. Co-contractions

Generally, the results obtained in the present study are in agreement with the experiments of Falconer and Winter (1985) where the greatest value of the CI was during weight acceptance and the swing phase for all children. However, in the current investigation, children with a TTA had less co-contraction during the single limb support phase compared to AB children, especially for the AL where a smaller value with small variability was apparent (see Fig. 2b). With this diminution of co-contraction, children with a TTA could present instability at their knee joint for both the NAL and AL. Knee joint stability is especially important for the single limb support phase because this period of the gait cycle requires high maintenance of balance. Thus, the near zero value of the resultant moment during the first half of the stance phase (0–30% of the gait cycle), observed by Winter and Sienko (1988), could be attributed to a decrease of muscular activity and co-contraction for adults with a TTA.

In regard to articular stability, during knee extension (action by the quadriceps), the hamstring assists the work of the anterior cruciate ligament to maintain knee joint stability and to produce a force opposed to the former translation movement of the tibia (Miller et al., 2000). The observed diminution of co-contraction could then create degenerative diseases, such as osteoarthritis, at the knee of the AL and particularly in the NAL (Burke et al., 1978; Melzer et al., 2001). The fact the NAL was mostly affected could be explained because, during support, children with a TTA have greater vertical, lateral and posterior ground reaction forces on the NAL compared to AL and even the CL (Engsberg et al., 1991; Engsberg et al., 1993; Lewallen et al., 1986), thus creating additional stresses on the structure of the knee joint of the NAL. One of the limit of this study is that we focus principally on joint laxity mechanisms and even if it is a major factor in the development of osteoarthritis (Felson et al., 2000; Issa and Sharma, 2006) others factors like increase contact forces between the tibia and the femur, nutritional factor, bone mineral density, etc. could affect the development of osteoarthritis (Felson et al., 2000).

Finally, this lack of co-contraction during single limb support might reduce the muscular energy expenditure of children with a TTA. Unnithan et al. (1996) suggested that co-contraction is a major factor responsible for the high energy cost of walking by children with cerebral palsy. Children with a TTA would thus choose an energy-saving strategy compared to a joint stability strategy which co-contraction can bring. The absence of difference at the level of co-contraction for the other phases of the gait cycle is also striking. Indeed, children with a TTA have less muscle to carry out these co-contractions; therefore, they demand more activity from the remaining muscles that might change joint constraints.

5. Conclusion

In the present study, children with a TTA used different agonist and antagonist muscle knee patterns during walking but showed consistent kinematic behaviour (angular excursion) compared to AB children. This reorganization, observed at the AL and NAL, occurs to maintain functionality of the gait task for children with a TTA, with some disadvantages. More precisely, changes in muscle patterns matched with kinematic invariance result in a diminution of knee co-contraction, reducing knee stability (or increasing joint laxity), especially during the single limb support period which is the most demanding period of the gait cycle regarding the maintenance of balance. This reduction of stability might be responsible for premature degenerative disease of both the NAL and AL knee joints. Since early modifications are apparent for subjects with a TTA, it is important to quantify the impact of the muscular modifications (atrophy, weakness, etc.) on the components of the musculoskeletal system to avoid the physical problems seen in adults with a TTA.

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