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Lateral stabilization improves walking in people with myelomeningocele

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Abstract

Muscle weakness and sensory deficits in people with myelomeningocele (MMC) make their walking control a greater challenge. We know little about how people with MMC optimize their walking balance. Recently, researchers have argued that medial–lateral control of gait requires more active neural input than the anterior–posterior direction, which is more passive. Our goal was to investigate the effect of providing external lateral stabilization (ELS) on walking patterns in people with MMC. We examined 12 people with MMC who could perform at least 4–6 independent steps. We found that the normalized step width (SW) was decreased 20% from without stabilizer to with stabilizer, where as the normalized step length (SL) was increased 4.17% from without stabilizer to with stabilizer. The ELS resulted in 25.10% reduction in centre of mass (COM) ranges of motions in the medial–lateral direction and 13.43% reduction in pelvic range of motions in the frontal plane. Our results suggested that by decreasing the medial–lateral control demands in people with MMC, we could improve gait with smaller SW, longer SL as well as reduced COM and pelvic ranges of motion in the frontal plane. In addition, ELS decreased energy cost and muscle co-activation of soleus and vastus lateralis that may help in diminishing the chances of pain and fatigue in people with MMC. Exploring the effect of the ELS provided us information that might be used to increase mobility safety and to develop a superior rehabilitation intervention for people with MMC.

Keywords: Gait stability; EMG; Energy expenditure; Variability; Spina bifida; Assistive device

1. Introduction

Myelomeningocele (MMC) is a form of neural tube defect. It most often occurs at the lumbar or sacral levels of the spine and causes leg muscle weakness and sensory deficits that affect locomotion pattern. The observed deviations from typical gait in this population include *decreased* cadence, SL, and walking speed, as well as *increased* lateral trunk sway, pelvis rotation, and pelvis obliquity (Bartonek et al., 2002a, b; Duffy et al., 1996a, b; Gutierrez et al., 2003; Thomson et al., 1999; Vankoski et al., 1995). The altered walking patterns (Gupta et al., 2005; Lim et al., 1998; Williams et al., 1993), muscle weakness, poor sensory function, reduced balance ability (Bartonek and Saraste, 2001), and decreased bone mineral density (Quan et al., 1998; Quilis, 1974) increase the risk of fatigue, joint pain, falling, and bone fracture in people with MMC. Furthermore, children with MMC utilize more energy than children without any neuromuscular disorder (Agre et al., 1987; McDowell et al., 2002; Bartonek and Saraste, 2001). Increased pelvis motions in the frontal plane have been hypothesized to be one of the underlying factors for increased energy expenditure since it was significantly related to oxygen cost (Bare et al., 2001). With a high energy cost during walking, people with MMC fatigue easily.

From a neural control perspective, controlling the foot placement in the medial–lateral (ML) direction requires more active neural involvement than does foot placement in the anterior–posterior (AP) direction for healthy people (Donelan et al., 2004; Kuo, 1999). One useful reflection of this is the variability of foot placement, which is significantly higher in the ML direction compared to in the AP direction in healthy people (Townsend, 1985; Bauby and Kuo, 2000). Step variability has been reported as a useful predictor of falling in the elderly (Hausdorff et al., 1997, 2001; Maki, 1997). Gabell and Nayak (1984)

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suggested that SW is related to balance control and that an increase in SW lead to greater stability, a possible compensation for instability. Maki (1997) suggested that decreased SW variability was related to falls with individuals. Recently, Brach et al. (2005) indicated that extreme SW variability was associated with falls during the past year in older persons who can walk at or near normal gait speed. All of these findings suggested that ML stability is very important for people to maintain balance and avoid falling.

Muscle weakness and sensory deficits in people with MMC make their walking control a greater challenge. We know little about how they manage the force needed to remain upright and move forward. Given their unique neuromotor constraints, their solutions to walking forward may or may not follow similar strategies as healthy people do. We propose the following hypotheses. With reduced sensory information from the lower limbs and poor muscle strength to control leg movements, people with MMC showed wider SW and increased active control in ML direction (increased SW variability).

To enhance mobility safety and to reduce risk of falling, it is imperative to improve walking stability in people with MMC. ELS provided for healthy people has been shown to decrease their need for active control in ML direction, that is reduced SW variability and metabolic costs (Donelan et al., 2004). Thus, ELS may be one of the most effective and efficient ways to enhance walking in people with MMC. By decreasing the ML control demands in people with MMC, we may significantly improve their walking ability with less variability of SW. In addition, the ELS may assist in decreasing energy cost and diminishing the chances of joint pain and fatigue in people with MMC. Exploring the effect of the ELS will provide us fundamental information to develop a better health care for people with MMC.

The *hypotheses* were: when provided with ELS during walking, people with MMC will (a) decrease SW, (b) decrease SW variability, (c) decrease muscle co-activation, and (d) decrease energy expenditure.

2. Methods

2.1. Participants

We recruited 12 people with lumbar or sacral level MMC (5 females, 7 males, age range = 6–26 years) who could walk independently without external support and any warning sign of neurological progression for at least 4–6 steps by working with physicians of the University of Michigan Hospital (Table 1). Participants and their parents signed assent and consent forms approved by the Institutional Review Board of the University of Michigan Medical School (IRBMED). Each person received a monetary gift for participating in this study.

Table 1
The mean and standard deviation (SD) of age, height, weight, body mass index (BMI) of our participants

	Age (year)	Height (m)	Weight (kg)	BMI
Mean	14.17	1.46	56.31	24.64
SD	6.07	0.21	28.85	6.47

2.2. Equipment and procedures

Each participant changed into their bathing suits at the Motor Development Lab of the Division of Kinesiology in the University of Michigan (Fig. 1). We collected participants' EMG baseline activity for each leg during sitting for 30 s and the resting heart rate by using a Polar heart rate monitor. We used a 6-camera Peak Motus™ real-time system to collect reflective marker position data at 60 Hz and a video camera to record overall behavior. Participants walked at their preferred speed (for 3 trials) over a GAITRite mat, which captures footfall data used for calculating participant's average overground walking velocity. Next, participants walked on the treadmill at their comfortable speed in two conditions: with and without ELS. All walking, overground and on the treadmill, was barefoot and without any orthotics or braces. We recorded heart rates and muscle activity at 1200 Hz by using the Therapeutics Unlimited EMG equipment (with six channels). The sequence of the conditions was randomized. Participants performed 6 trials (3 trials for left

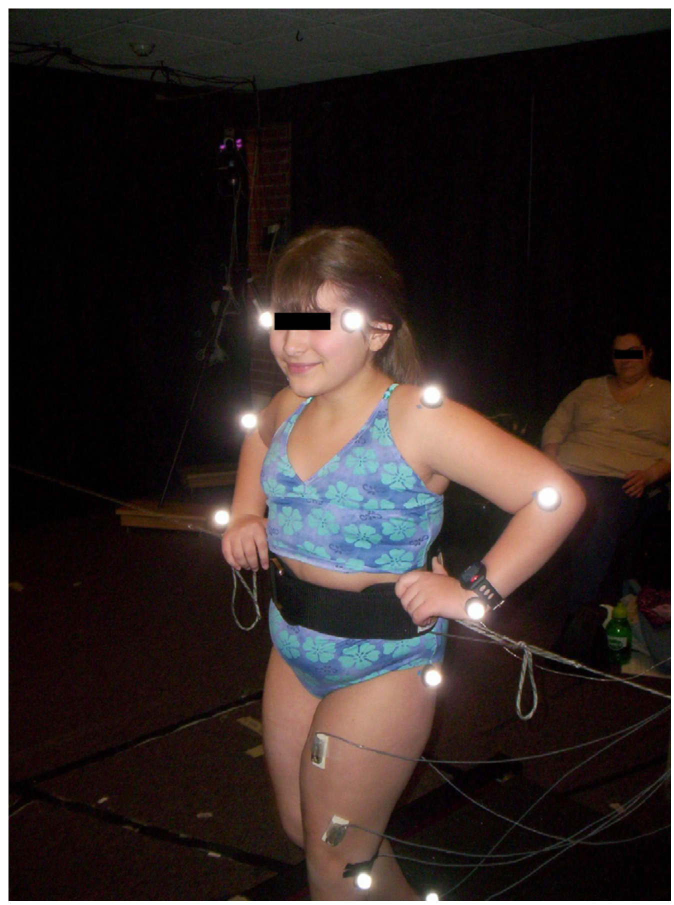

Fig. 1. . An example of one participant with myelomeningocele standing with the ELS. To collect kinematic data, we attached reflective markers (2.5 cm diameter) to the lateral sides of the body. After cleaning the skin surface by using alcohol pads, we placed surface preamplified bipolar electromyographic (EMG) electrodes over the following muscles: tibialis anterior (T), gastrocnemius (G), soleus (S), quadriceps (rectus femoris, QR; vastus lateralis, QV), and hamstrings (biceps femoris, H). We placed a chest strap of Polar heart rate monitor around the participant's chest at the approximate level of the 5th intercostal space where the heart is located. The chest strap is covered by the bathing suit.

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side and 3 trials for right side EMG) for each condition. Each trial was 10 s with a 30 s resting period.

The ELS consisted of a padded belt from which two adjustable cords (two light-weight nylon cords and two rubber tubings) extended to opposite walls (Fig. 2). The stiffness and damping of this spring–damped–mass system, calculated by applying a second-order damped oscillator model (Donelan et al., 2004), were approximately 2000 N/m and 15 N s/m, respectively.

2.3. Data reduction

Raw kinematic data were converted to 3D data via the Peak Motus system software and filtered with a second order Butterworth filter at a cut-off frequency of 6 Hz (Chang et al., 2006). The gait events, touchdown and toe-off, were determined via behavior coding. The time of touchdown was at the frame in which any part of the foot contacted the ground at the beginning of the stance phase. The time of toe-off was identified when the foot was off the ground at the beginning of the swing phase. We used touchdown to identify onset of each stride cycle. The frontal plane motions of the trunk were calculated via the Peak Motus system software and the custom-written MatLab programs. The SL was determined by the AP direction distance between the toe marker of one side and the toe marker of the other leg for each touchdown. The distance traveled in the AP direction during each step was corrected by the treadmill speed. The ML direction distance between the toe markers of both legs determined the SW. The SL and SW were normalized to the leg length. The SW and SL variability were standard deviations of SW and SL that were normalized by leg length. We determined the center of mass (COM) position by using Winter's method (1990) and calculated the COM ranges of motion (lateral displacement) in the frontal plane.

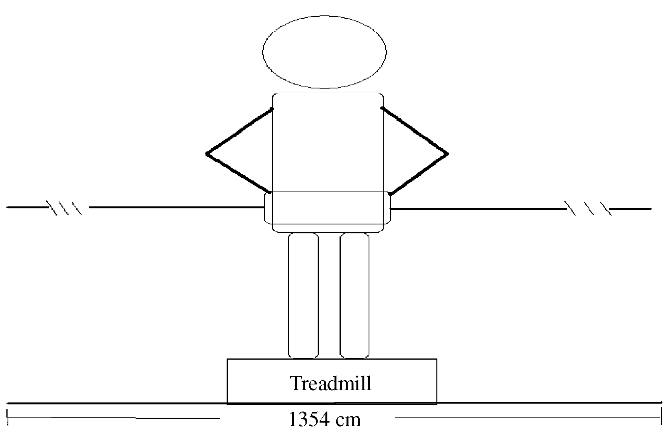

Fig. 2. An illustration of the external lateral stabilization with a padded belt from which two lightweight adjustable cords extended to the opposite walls. The distance between the walls was 1354 cm. A long distance between the walls enabled us to use long length adjustable cords. This ensured that any non-lateral forces imposed on the participants were very small and negligible since the anterior–posterior displacement of each participant during walking was very small relative to the long length of the cord. The cords were perpendicular to the treadmill and stabilized participants in the medial–lateral direction. Since the cords of the ELS around the waist level interfered with normal arm swings, each participant walked with hands placed on the lateral side of the trunk with and without ELS. If the participant was not able to walk with hands placed on the lateral side on the trunk on the treadmill, they could flex their arms to their comfortable position without touching the cords. Each participant performed the same arm position that he or she felt comfortable in both conditions (with and without ELS). The same ELS with the same stiffness and damping were used for all participants, approximately 2000 N/m and 15 N s/m, respectively.

The raw EMG data were high-pass filtered at 20 Hz, rectified, and low-pass filtered at 6 Hz. To determine on–off activity (Hodges and Bui, 1996), threshold value was 3 standard deviations beyond the mean of baseline activity for that muscle. The number of sequential samples during which muscle activity must exceed the threshold value was 50 ms. Due to the differences in stride cycle duration among individuals, we normalized the burst duration by the stride cycle duration. The co-activation indexes for each muscle pair, T and G, T and S, QR and H, QV and H, G and QR, G and QV, S and QR, as well as S and QV, were calculated based on the method originally designed by Winter (1990). The co-activation equation we used is defined below.

Co-activation index $= 2 \times$ (overlapping area of muscle A and muscle B activity)/(muscle A activity + muscle B activity).

Previous studies reported the strong linear relation between oxygen uptake and heart rate in different velocities which supports the use of the energy expenditure index (EEI) as a walking energy expenditure indicator (Rose et al., 1989, 1991). The EEI was calculated by using the following equation:

EEI = (walking heart rate (beats/per min)—resting heart rate (beats/per/min))/walking speed (cm/s)

2.4. Data analysis

The statistical methods were 2 (Condition) $\times$ 2 (Direction) ANOVAs with repeated measures on condition to test the impact of the ELS on foot placement and walking control in people with MMC. The two conditions were (a) without ELS and (b) with ELS. The two directions were (a) ML direction and (b) AP direction. For the question of impact on foot placement, the dependent variable was mean of distance. The distance in ML direction was SW. The distance in AP direction was SL. For the question of impact on walking control, the dependent variable was variability of distance. One-way ANOVAs with repeated measures on condition were used to examine the effect of the ELS on COM, trunk, and pelvis motions in the frontal plane, muscle co-activation, and energy expenditure. The dependent variables were COM, trunk, and pelvis ROM in the frontal plane, muscle co-activation index, and EEI.

3. Results

The 2 (Condition) $\times$ 2 (Direction) ANOVA with repeated measures on condition for means of normalized SW and normalized SL showed significant Condition effect ($F(1, 23) = 6.855$, $p = 0.015$), Direction effect ($F(1, 23) = 31.445$, $p < 0.001$), and interaction effect ($F(1, 23) = 13.415$, $p = 0.001$). The significant Condition effect showed the normalized SW was narrower with stabilizer than without stabilizer (Figs. 3 and 4) as well as the normalized SL was larger with stabilizer than without

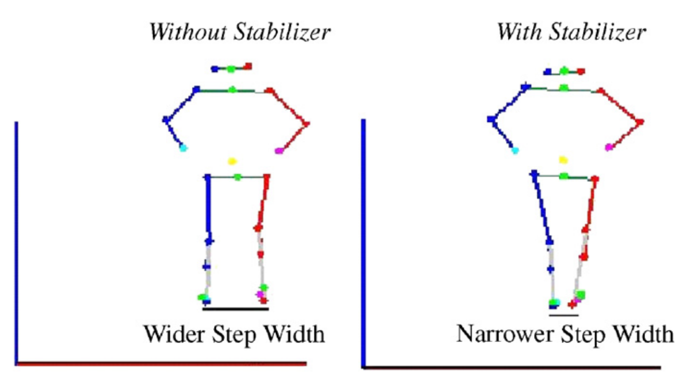

Fig. 3. Examples of step width on the treadmill, with and without the lateral stabilizer device, plotted in the frontal plane.

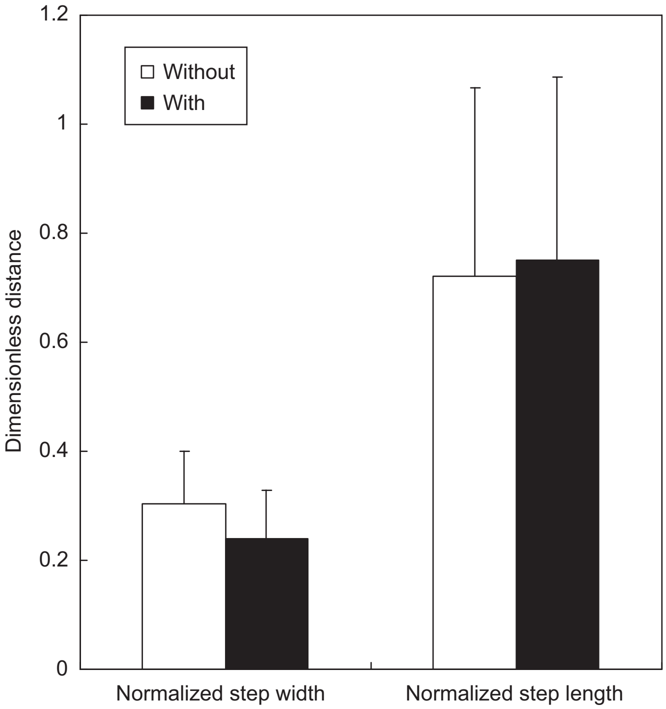

Fig. 4. . The means of normalized SW on the treadmill for with and without lateral stabilization were 0.24 (SD = 0.089) and 0.3 (SD = 0.097), respectively. The means of normalized SL on the treadmill for with and without lateral stabilization were 0.75 (SD = 0.336) and 0.72 (SD = 0.345).

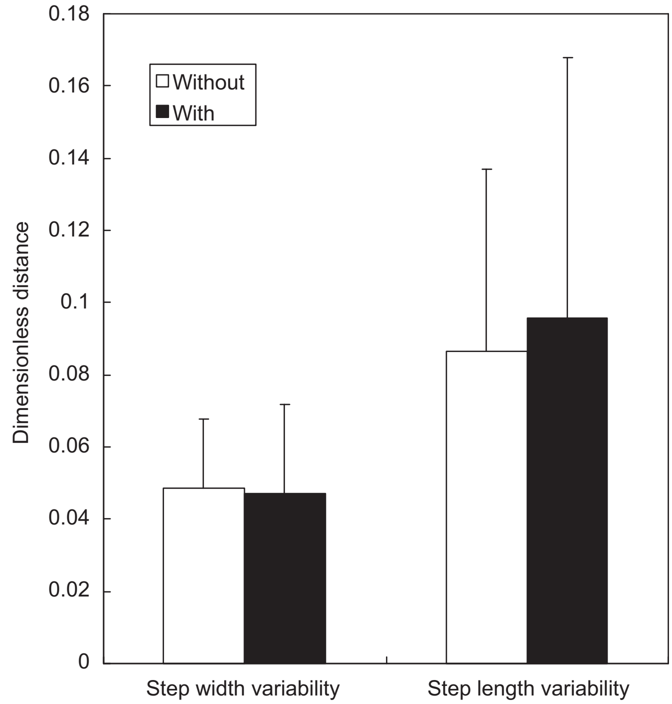

Fig. 5. The SW variability on the treadmill for with and without lateral stabilization was 0.0470 (SD = 0.0249) and 0.0484 (SD = 0.0194), respectively. The SL variability on the treadmill for with and without lateral stabilization was 0.0955 (SD = 0.0724) and 0.0866 (SD = 0.0503).

stabilizer. The significant Direction effect showed the normalized SL (AP direction) was larger than normalized SW (ML direction). The significant interaction effect showed the normalized SW was *decreased* 20% from without stabilizer to with stabilizer where as the normalized SL was *increased* 4.17% from without stabilizer to with stabilizer.

The results of 2 (Condition) $\times$ 2 (Direction) ANOVA with repeated measures on condition for variability of foot placement showed significant Direction effect ($F(1, 23) = 23.377$, $p < 0.001$), but no significant Condition and interaction effects (Fig. 5). The SW variability was 44.11% lower than the SL variability when walking without ELS. With ELS, the SW variability was 50.79% lower than the SL variability. The SW variability was 47.6% lower than the SL variability regardless of with or without ELS.

The COM ranges of motion in the frontal plane, with and without ELS, were 0.0358 (SD = 0.01835) and 0.0478 (SD = 0.016), respectively. Our analysis showed significant Condition effect ($F(1, 23) = 34.525$, $p < 0.001$). The ELS resulted in 25. 10% reduction in COM ranges of motions in the frontal plane.

The pelvic ranges of motion in the frontal plane, with and without ELS, were 6.2258 (SD = 3.3157) and 7.1915 (SD = 4.2449), respectively. The results showed significant Condition effect ($F(1, 23) = 5.944$, $p = 0.023$). The ELS resulted in 13.43% reduction in pelvic range of motions in the frontal plane. We did not find significant Condition effect for trunk ranges of motion in the frontal plane, with and without ELS.

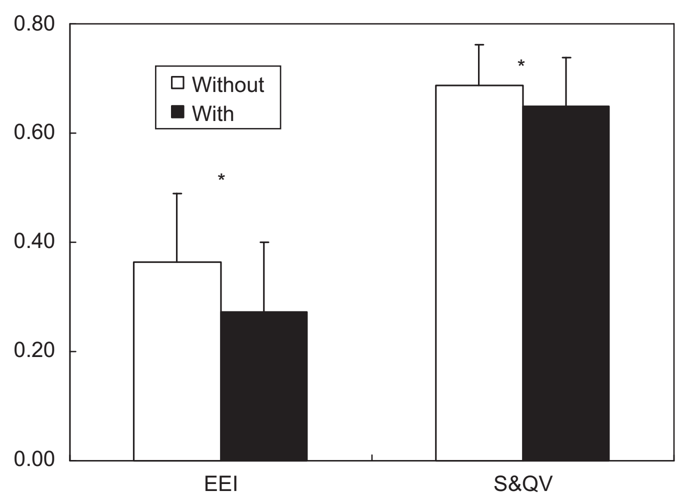

Fig. 6. The means for the energy expenditure index (EEI) during treadmill walking, with and without lateral stabilization, were 0.2731 (SD = 0.1266) and 0.3632 (SD = 0.1261), respectively. The co-activation index of S and QV during treadmill walking for with and without lateral stabilization was 64.84% (SD = 9.075%) and 68.76% (SD = 7.473%), respectively.

The EEI values were significantly different between with and without ELS ($F(1, 23) = 85.379$, $p < 0.001$) (Fig. 6). The ELS resulted in a decrease of 24.81% in energy expenditure. The co-activation index of S and QV were also significantly different between conditions ($F(1, 23) = 15.175$, $p = 0.001$) (Fig. 6). The ELS resulted in a decrease of 5.7% in muscle co-activation of S and QV. We did not find statistically significant Condition effect on either co-activation index for other muscle pairs or normalized muscle burst duration for each leg muscle.

4. Discussion

During treadmill walking, people with MMC preferred a shorter normalized SL and a wider normalized SW with increased SW and SL variability compared to healthy people (Table 2). Our results implied that with reduced sensory information from the lower limbs and poor muscle strength to control leg movements, people with MMC showed wider SW, shorter SL, and increased active control in both AP and ML directions.

A wider SW during walking is generally recognized as a sign of poor balance and instability. With the ELS, people with MMC decreased their SW. It suggests that lateral instability impacts on the choice of preferred SW in people with MMC. The 20% decrease in SW and 24.81% decrease in energy costs implied that people with MMC benefited from the ELS with lower costs of step-to-step transitions and increased ML stability. The ELS caused healthy people to prefer a 47% narrower SW and a 5.7% decrease in metabolic cost (Donelan et al., 2004). However, we found a 20% decrease in SW but a 24.81% decrease in energy costs for people with MMC. It suggests that lateral stabilization is a major factor that impacts the metabolic cost of walking in people with MMC.

Our finding that people with MMC show less decrease (20%) in SW while being provided with ELS compared to healthy people (47%) may result from their limited sensory and proprioception information for lower limbs. When walking with reduced sensory information, both healthy people and persons with peripheral neuropathy increase SW (Bauby and Kuo, 2000; Richardson et al., 2004). Limited sensory information may have an impact on SW. While ELS improves gait patterns with narrower SW in people with MMC, reduced sensory information still restricts their preferred SW.

Large COM motions in the frontal plane might induce energetic cost of walking (Saunders et al., 1953; Donelan et al., 2001). In a computer simulation model, redirecting COM during walking costs mechanical work and energy (Kuo et al., 2005). The wider SW in the frontal plane requires more mechanical work to redirect the COM during step-to-step transitions, with a proportional increase in metabolic costs (Donelan et al., 2001, 2002). When people with MMC walk with wider SWs, they might require more energy to redirect their COM in the frontal plane. Increased pelvis motions in the frontal plane (pelvic obliquity) have been shown to be related to increased oxygen cost of walking for people with MMC (Duffy et al., 1996a). We believe that people with MMC increase their energy costs to maintain the stabilities of COM and pelvis in the frontal plane. Our results implied that the ELS improved the stability of walking and decreased energy costs in people with MMC with reductions of COM and pelvis motions in the frontal plane. It robustly emphasized the importance of body stability in the ML direction and provided future clinicians with background information to design rehabilitation techniques for people with MMC.

Our findings indicated SW variability for people with MMC was 44.11% lower than SL variability. Previous researchers showed similar results in other populations. For healthy adults, SW variability was 38.5% lower than SL variability after treadmill speed correction in AP direction (Donelan et al., 2004). For toddlers with typical development and toddlers with Down syndrome at the onset of walking, SW variability was lower than SL variability during overground walking (Looper et al., 2006). However, one study showed SW variability was larger than SL variability in young healthy adults (Bauby and Kuo, 2000).

Our question is why people with MMC showed higher variability in the AP direction compared to the ML direction. It might result from three factors. First, reduced leg muscle strength might make it difficult to control the foot placement in AP direction. Second, insufficient sensory feedback might induce increased variability in the AP direction when the swing leg is moving through the air as well as when a foot is targeting its foot placement on the treadmill belt. Third, stability in the ML direction might be more essential than stability in the AP direction for people with MMC. As people with MMC actively control their walking in ML direction to prevent falling, they sacrifice foot placement variability in AP direction. It implies that walking control in ML direction is the key antidote for people with MMC to independently progress forward in the sagittal plane and maintain balance.

For the effect of ELS on muscle activity, the ELS decreased muscle co-activation of S and QV during

Table 2
Normalized step length (SL), step width (SW), variability of step length (VSL), variability of step width (VSW) for people with MMC and healthy people (Donelan et al, 2004)

	SB				Healthy people (Donelan et al., 2004)			
	SL	SW	VSL	VSW	SL	SW	VSL	VSW
Mean	0.72	0.3	0.0866	0.0484	0.756	0.121	0.026	0.016
SD	0.345	0.097	0.0503	0.0194	0.052	0.029	0.009	0.003

walking for people with MMC. Soleus (S) provides energy needed for trunk forward progression during the single-leg stance (Neptune et al., 2004). The function of QV is to accelerate the trunk in the sagittal plane in the beginning of stance (Neptune et al., 2001). As the ELS decreased muscle co-activation of S and QV, it suggests that reducing mechanical demands of walking in the frontal plane could assist people with MMC in accelerating the trunk and progressing forward in the sagittal plane during the stance phase. While muscle co-activation is generally considered as a factor that increases energy cost, decreasing muscle co-activation of S and QV might contribute to the reduction of energy expenditure. It is consistent with our results that the energy expenditure decreased when applying the ELS.

Our findings indicated that the external lateral stabilization (ELS) improved the treadmill walking of people with MMC in the ML direction. Clearly, there is something to be said for the roles of arm position during human walking. The major function of arm-swinging in human walking has been suggested to balance rotational forces in the human trunk, which may be due to the upper limb moves forward as the opposite lower leg moves forward (Elftman, 1939). As the frequency of arm swing is also associated with the frequency of leg swing (Kubo et al., 2004), the frequency ratio between arm and leg movements is around 2:1 or 1:1 which depends on walking velocity. Due to the interference of the ELS, participants in this study placed their hands on the lateral side of the trunk just above the pelvis or flexed their arms without touching the cords of the ELS. With free arm movements the reduction in energy expenditure and gait parameters by the lateral stabilizer might not have been so high. If arms moved freely the control condition may not have had such high energy cost. However, they selected the arm position that they felt comfortable and maintained the same arm position during both conditions (with and without lateral stabilization) to control the effect of the arm swing. We suggest future researchers to design a portable lateral stabilization walker without interfering regular arm swing during overground walking. Thus, patients can apply this technique in real world.

The results of this ELS study also shed some light on designing better rehabilitation programs for patients with MMC. Traditionally, a wider step width (SW) is associated with tightness in the iliotibial band (IT band) or hip abductors. Thus, clinicians may treat a wide SW with hip abductor stretching exercises. Our results suggest clinicians should consider an alternate explanation for patients with MMC. A wider SW shown in patients with MMC may not be due to muscle tightness or IT band tightness, but related to their solutions to overcome their reduced walking stability in the ML direction.

Overall, our results suggested that ELS decreased the ML control demands in people with MMC. This technique improved their gait patterns by producing smaller SW, larger SL as well as reduced COM and pelvic ranges of motion in the frontal plane. The ELS also assisted in decreasing energy cost and muscle co-activation of S and QV that may help in diminishing the chances of joint pain and fatigue in people with MMC. Exploring the effect of the ELS provided us significant and fundamental information that might be used to increase mobility safety and to develop a superior rehabilitation intervention for people with MMC.

Conflict of interest statement

There is no conflict of interest.

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