Evaluation of Force-Sensing Resistors for Gait Event Detection to Trigger Electrical Stimulation to Improve Walking in the Child With Cerebral Palsy

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***Abstract*—Force-sensing resistors (FSRs) were used to detect the transitions between five main phases of gait for the control of electrical stimulation (ES) while walking with seven children with spastic diplegia, cerebral palsy. The FSR positions within each child's insoles were customized based on plantar pressure profiles determined using a pressure-sensitive membrane array (Tekscan Inc., Boston, MA). The FSRs were placed in the insoles so that pressure transitions coincided with an ipsilateral or contralateral gait event. The transitions between the following gait phases were determined: loading response, mid- and terminal stance, and pre- and initial swing. Following several months of walking on a regular basis with FSR-triggered intramuscular ES to the hip and knee extensors, hip abductors, and ankle dorsi and plantar flexors, the accuracy and reliability of the FSRs to detect gait phase transitions were evaluated. Accuracy was evaluated with four of the subjects by synchronizing the output of the FSR detection scheme with a VICON (Oxford Metrics, U.K.) motion analysis system, which was used as the gait event reference. While mean differences between each FSR-detected gait event and that of the standard (VICON) ranged from +35 ms (indicating that the FSR detection scheme recognized the event before it actually happened) to −55 ms (indicating that the FSR scheme recognized the event after it occurred), the difference data was widely distributed, which appeared to be due in part to both intra-subject (step-to-step) and intersubject variability. Terminal stance exhibited the largest mean difference and standard deviation, while initial swing exhibited the smallest deviation and preswing the smallest mean difference. To determine step-to-step reliability, all seven children walked on a level walkway for at least 50 steps. Of 642 steps, there were no detection errors in 94.5% of the steps. Of the steps that contained a detection error, 80% were due to the failure of the FSR signal to reach the programmed threshold level during the transition to loading response. Recovery from an error always occurred one to three steps later.**

***Index Terms*—Cerebral palsy, children, foot switches, functional electrical stimulation, gait.**

I. Introduction

FOR THE child with spastic diplegia, cerebral palsy (CP), the ability to ambulate is compromised. Walking for these children often requires an assistive device and/or orthoses to provide balance, support, and foot clearance. Children with CP exhibit motion disorders, including spasticity, athetosis, and/or atoxia, resulting in imbalances between agonist and antagonist muscles, apparent muscle weakness, and a loss of selective muscle control. This results in poor stability during the stance phase of gait and inadequate foot clearance during the swing phase. [1], [2]

Electrical stimulation (ES) has been investigated as a means of improving mobility for this population by modulating spasticity, unmasking or improving volitional control, improving joint range of motion, strengthening weak muscles, and stimulation of muscle during an activity, such as walking, in an attempt to ensure adequate recruitment and timing of muscles [3]–[9].

In walking, ES has typically been applied to the ankle plantar flexors to improve tibial control in stance for ankle stability and to the dorsiflexors or the peroneal nerve to improve foot clearance during the swing phase. To time the stimulation of these muscles, differentiation of the swing and stance phases of gait has been accomplished by handswitches controlled by the user or therapist [5], [6] or by changes in plantar pressure using force-sensing resistors (FSRs) [7]–[9]. An FSR is a sensor whose electrical resistance changes in proportion to the pressure applied to it. As it applies to gait event detection, the FSR(s) is(are) placed in the insole of the shoe so that the change in plantar pressure can be directly related to a gait event. Naumann [9] reported successful use of foot switches for swing/stance differentiation to trigger tibialis anterior or peroneal nerve stimulation for two children with CP. Gracanin [7] described a foot-switch-driven two-channel peroneal nerve stimulator. With this system, the peroneal nerve was stimulated using either "heel off" of the foot that was stimulated (ipsilateral mode) or heel strike of the opposite foot (contralateral mode). However, no reference to foot switch performance was made in either study.

While FSRs have been commonly employed for ES control during walking for the population with spinal cord injury [10]–[14], the atypical and variable foot-to-floor contact patterns encountered with children with CP [2] potentially make the use of FSRs particularly challenging [7] even for relatively straightforward swing and stance phase differentiation. Further, to stimulate muscles that are typically active for portions of swing and stance such as the hip and knee extensors, it would be beneficial to further distinguish the transitions within the stance and swing phases to elucidate the benefit of various ES timing schemes.

Efforts at our institution have focused on applying FSR-triggered ES directly while walking to both proximal and distal

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TABLE I
LIST OF STUDY PARTICIPANTS

Subject	Study Child Participated In R- Reliability A-Accuracy	Age	GMFCS	Braces/Assistive Devices	Surgery
AF	R, A	6	II	None	None
AL	R, A	7	IV	Walker	Bilateral Muscle Lengthenings(BML)
CW	R, A	7	III	Walker	BML, Bilateral Osteotomies (BO)
SW	R, A	8	III	Hinged Ankle Foot Orthoses (AFOs) & Walker	None
AM	R	13	II	Supramalleolar Orthoses & 1 Crutch	BML,left split anterior tibialis tendon transfer, BO
DW	R	7	IV	AFOs & Walker	BML, BO
PL	R	11	IV	AFOs & Walker	BML

lower extremity musculature to determine both the carryover and immediate effects on the walking ability of children with CP. The purpose of this paper is to describe quantitatively the performance of a gait event detection scheme using FSRs to trigger ES with seven children with CP.

II. METHODS

A. Subjects

Seven subjects between the ages of 7 and 13 y participated in the study (Table I). All of the subjects met the selection criteria as previously described [15]. All children were diagnosed as spastic diplegia, cerebral palsy, and ambulated with inadequate hip and knee extension in stance with excessive plantar flexion in single limb support. Self-selected walking speed ranged from 0.35 to 1.2 m/s. One of the subjects walked without an assistive device, while the remaining six required either a walker or crutches. Five of the seven subjects wore ankle foot orthoses bilaterally. The subjects with CP fell between a level II and IV on the Gross Motor Function Classification Scale. This scale is based on self-initiated movement with particular emphasis on sitting (trunk control) and walking [16].

For each subject, percutaneous intramuscular electrodes were implanted, using a needle insertion procedure described previously [17]. For all subjects, the gluteus maximus was implanted for hip extension, the gluteus medius for hip abduction, and the vastus lateralis and vastus medialis for knee extension. For four subjects, the tibialis anterior was implanted for dorsiflexion, and for five subjects, the soleus was implanted for plantar flexion. Each electrode was profiled after implantation to determine the stimulation parameters necessary to develop the most forceful muscle contraction that could be tolerated by the child. Using a research-grade stimulator designed in our laboratory, a charge-balanced asymmetrical biphasic stimulation waveform was applied using an amplitude of 20 mA at 20 pulses per second. Stimulated muscle force was modulated using pulse durations of up to 200 μs. Five of seven subjects also had surgical interventions to correct joint contractures for the application of ES (Table I).

B. FSR Characteristics

The FSR[1] is a sensor whose electrical resistance decreases when pressure is applied to the active surface. The rise and fall times of the FSR output are on the order of 1 to 2 ms. The FSR response approximately follows an inverse power law characteristic [18] such that within the range of the weights of the subjects of this study (20 to 55 kg), an approximately linear relationship exists between FSR resistance and the force applied to it. A circular FSR with a 7/8-in diameter and 0.5-mm thickness was used, as its size was best suited to placing multiple sensors given the small size of the children's feet. A 1.5-μF capacitor in parallel with each FSR provided low-pass filtering before digitization, which introduced electrical delays of approximately 5 ms depending on whether the FSR was being weighted or unweighted.

C. FSR Placements

Because children with CP produce atypical plantar pressure patterns while walking, it was not possible to use standard foot-to-floor contact positions [2] for FSR placement for gait event detection. In addition, with our pilot subjects [19], [20], we found it time consuming and thus fatiguing to the child to iteratively move FSRs about the insole to locate areas of greatest plantar pressure corresponding to each gait event. Thus, we decided to first quantify each child's plantar pressure profile using an F-Scan system.[2] With a thin (0.18 mm) insole sensor array that is trimmed to accommodate the entire foot surface, the F-Scan provided a complete plantar pressure map of the foot throughout the gait cycle normalized to the subject's body weight. The F-Scan output provided a color-coded pressure map in 0.5-cm^2 increments at 10-ms intervals. To set FSR locations, two or three areas of maximum pressure (area defined by sensor size) were identified for each foot that corresponded temporally to stance phase events. The initial area of greatest pressure as the foot contacted the floor was used as an FSR location for the transition into loading response for the ipsilateral leg and typically the transition into preswing

[1]Interlink Electronics, Camarillo, CA.
[2]Tekscan, Inc., Boston, MA.

for the contralateral leg. A second FSR location was defined as the last area of greatest pressure just before the foot left the ground. This FSR was used to identify the transition into initial swing on the ipsilateral leg and typically the transition into midstance for the contralateral leg. Alternatively, if a distinct pressure transition from the back of the foot toward the midfoot was present on the ipsilateral leg, then it was used to denote midstance. For terminal stance, if a progression of plantar pressure *off* the back foot and onto the forefoot existed, then the area of increased pressure in the forefoot was used as a terminal stance FSR location. Alternatively, the reduction of hindfoot pressure was also used to identify terminal stance when it was clear. For this analysis, each subject walked under the same conditions (i.e., using the same orthotic and assistive device) that were subsequently used with ES. Two or three FSR locations were identified per foot. Occasionally, one FSR was used when one location could satisfy several of the conditions above.

Once the FSR positions were determined, the sensor locations were traced from the F-Scan printouts onto a transparency and from there were superimposed onto the FSR insoles. The insoles were custom cut to fit each child's shoe snuggly to minimize movement within the shoe. If the child walked with ankle foot orthoses (AFOs), the FSRs were placed between the AFO and the subject's insole. The FSRs were placed on the insoles so that the active surface was on the superior side. Small slits were made so that the solder tabs for each sensor went through the insole. Wiring connections between the FSRs and the common cable were made on the underside of the insole at the arch of the foot. The FSR cable was pathed up the leg and connected to the stimulator at the waist.

D. Gait Event Detection and Stimulation Timing

A 24-channel portable stimulator was used [21] to implement the FSR-driven ES system. Once the FSR insoles were fabricated, a three-step process was employed to program ES-assisted gait. First, the subject walked without ES enabled for approximately 20 steps with the FSR insoles while the portable stimulator collected the FSR signal data. From these data, the FSR threshold levels for determining each gait event were determined. The occurrence of a gait event was determined by whether the FSR or FSRs sensitive to that gait event were ON or OFF. An FSR was ON if the signal exceeded the designated threshold level, and an FSR was considered OFF if the signal went below a designated threshold level. The ON and OFF threshold levels for a given FSR could be set independent from each other. Threshold levels were generally set midway between the minimum and maximum force level to avoid susceptibility to possible spurious deflections at lower threshold levels and to minimize detection delays, which would occur at higher threshold levels. Once determined, the FSR threshold detection levels for each gait event were downloaded to the portable stimulator. Within the portable stimulator, the FSR threshold detection scheme searched sequentially for the FSR conditions that satisfied each gait event in the order in which the gait events naturally occur (i.e., loading response, midstance, terminal stance, preswing, and initial swing). The scheme would not advance until the current FSR condition(s) was(were) satisfied. The FSR conditions were examined sequentially to prevent large swings in the detection output, which could introduce abrupt and potentially unsafe changes in ES.

As the second step in the process, the subject then walked 20 steps with the FSRs and without ES enabled while the stimulator recorded the FSR signals and the gait event detection output to verify that the gait event detection scheme was properly cycling. If needed, threshold detection levels were adjusted and another walk was performed to confirm that the FSRs were deflecting sufficiently and thus that the detection scheme was cycling properly.

In the third step, once the gait event detection scheme was established without ES enabled, muscle stimulation was programmed into the portable stimulator. Stimulation was programmed in the following manner. The hip and knee extensors and hip abductor muscles were activated during loading response into midstance in an attempt to improve stance phase stability. When available, ES of those muscles was initiated during the terminal swing phase by using the terminal stance transition from the opposite leg as the trigger. The plantar flexor muscle was stimulated from midstance to preswing with maximal stimulation during terminal stance in an attempt to improve ankle stability and pushoff force. The ankle dorsiflexors were stimulated from initial swing to loading response.

III. Data Collection and Analysis

Following completion of the ES programming, children walked in the laboratory and at home with FSR-triggered ES. All subjects were at least 1 y postsurgery and electrode implantation and had at least three months' experience walking with their FSR-triggered ES system at the time of data collection. Children were asked to walk with the ES system at least 3 d per week for 1 h. Both accuracy and reliability data on the FSR gait event detection scheme were collected while the subjects walked with their ES systems. The FSR signals were digitized at 100 Hz and stored simultaneously with the output of the FSR gait event detection scheme within the stimulation unit. After each trial, the data were transferred to a desktop computer for analysis.

A. Accuracy

Because three of the subjects were not available at the time these data were collected, the comparison of the occurrence of the FSR-detected gait events to the actual time of the gait events was carried out with four of the seven subjects. To determine the accuracy of gait event detection using FSRs, the algorithm output was compared to actual gait events by collecting the algorithm data in synchrony with joint kinematic data. For this, the six-camera VICON[3] motion-analysis system was used with markers placed bilaterally at the toe, ankle, and knee.

During data collection, the subject was prompted to walk at a comfortable pace. Data were collected for at least eight steps for each subject. A cue switch was interfaced to the VICON system to synchronize the output of the gait event detection

[3]Oxford Metrics, U.K.

scheme and the individual FSR signals with the kinematic data. Once the trial was complete, the actual gait events were determined for each leg from the kinematic data using the following definitions for gait phase transitions. The initiation of loading response was the point where the first ipsilateral foot marker stopped moving upon floor contact. The midstance gait event began when the contralateral toe marker started to move upward. By watching the ipsilateral ankle marker start to elevate, terminal stance was identified. Preswing started when the first contralateral foot marker stopped moving, signifying ground contact. The final gait event, initial swing, began when both of the ipsilateral foot markers first started to move upward. Because the sampling rate of the VICON system was fixed at 60 Hz, the timing of an event had a resolution of 16.7 ms. Because the FSR data were sampled at 100 Hz, this introduced a measurement offset with the VICON system. The same person determined all gait events for all subject data to eliminate interrater variability.

For each child's steps, the time corresponding to each gait event and the time that the FSR gait event detection scheme detected that event were recorded. The difference between the two times was then determined. Then an overall mean difference and the first and second standard deviations were calculated for each gait event.

B. Reliability

The reliability test was designed to determine the consistency of FSR performance while the child walked with ES enabled. To determine step-to-step reliability, each of the seven subjects walked on a level walkway for one to four 50-step trials with all walks for a given subject completed in the same session. As mentioned earlier, the detection scheme was designed to search sequentially for the FSR conditions that satisfied each gait event in the order in which the gait events naturally occur. The scheme would not advance until the FSR condition(s) it was presently waiting for was satisfied. Thus, a detection error was identified by examining the output of the detection scheme concurrently with the individual FSR signals (Fig. 1). If the detection output failed to continue to cycle with changes in the FSR signals, then this was noted as a detection error and the FSR(s) associated with that failure and the reason for that error were tabulated. Also examined was the number of steps required until proper cycling was recovered following an error. The first and last steps taken were not included in the analysis.

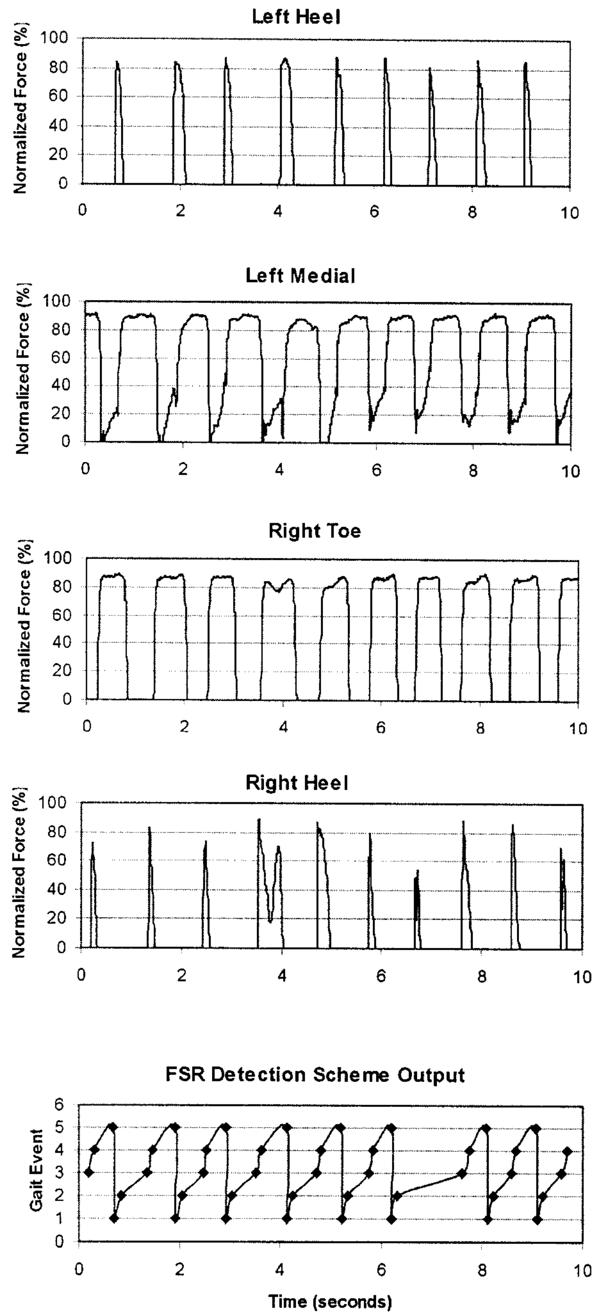

Fig. 1. A 10-s portion of a reliability trial for subject AF showing the four FSR signals used for her detection scheme and the output of the FSR detection scheme. Note that between 6 and 8 s into the trial, an event was missed due to the inability of the right heel to reach its programmed threshold level. The detection scheme recovered on the next step when that condition was satisfied. [Gait Event Key: 1—transition into loading response (left) and preswing (right), 2—midstance (left) and initial swing (right), 3—preswing (left), 4—initial swing (left) and midstance (right), 5—terminal stance (right)].

IV. Results

A. FSR Placement

The FSR conditions used for each programmed gait event for each child are shown in Table II, and a summary of the locations of the FSRs used to trigger the various gait events are shown in Table III. The FSR placements were categorized as being either at the heel, midfoot, or toe region of the fabricated insole. The regions were defined by two lines that transsected the foot in the coronal plane: one line beginning at the head of the first metatarsal and the second line beginning at the most proximal aspect of the medial arch. Seventy-seven percent of the loading response FSRs and 70% of terminal stance FSRs came from the ipsilateral heel or midfoot. Most midstance (79%) and preswing (83%) events were triggered using contralateral FSRs.

B. Accuracy

Using the technique described by Bland and Altman [22] (Fig. 2), for each gait event, the difference between the onset times registered by the FSR detection scheme and the VICON

TABLE II

COMPILATION OF WHICH FSRS WERE USED. THE ARROW SIGNIFIES WHETHER FORCE IS BEING APPLIED (↑) OR REMOVED (↓) FOR THE PURPOSE OF DETECTION. CELLS WITH LINES THROUGH THEM INDICATE GAIT PHASE TRANSITIONS THAT WERE NOT PROGRAMMED. KEY: LH-LEFT HEEL; RH-RIGHT HEEL; LT-LEFT TOE; RT-RIGHT TOE; LM-LEFT MIDSOLE; RM-RIGHT MIDSOLE

Subject	Foot	CW	AL	SW	AF	AM	PL	DW
Weight Acceptance	Left	*LH*↑	*LM*↑	*LH*↑ *or LM*↑	*LH*↑	*LH*↑ *or RM*↓	-----	*LH*↑ *or LT*↑
	Right	-----	*RM*↑	-----	-----	*LH*↓ *or RH*↑	*RH*↑	-----
Mid-Stance	Left	-----	*RT*↓	*RM*↓ *& RT*↓	*RT*↓	*LM*↑ *or LT*↑	*RT*↓	-----
	Right	*LM*↓	*LM*↓	*LT*↓	*LM*↓	*LM*↓ *or RL*↑	*LM*↓	-----
Terminal Stance	Left	*LH*↓	-----	*LH*↓	-----	*LH*↓	*LM*↑	*LH*↓
	Right	*LM*↓	-----	*RH*↓	*LM*↑	-----	*RT*↑	*RM*↑
Pre-Swing	Left	-----	*RM*↑	*RT*↑	*RH*↑	*LH*↓ *or RH*↑	*RH*↑	*RT*↑
	Right	*LH*↑	*LM*↑	*LH*↑ *or LM*↑	*LH*↑	*LH*↑ *or RM*↓	-----	*LH*↑ *or LT*↑
Initial Swing	Left	*LM*↓	*LM*↓	*LT*↓	*LM*↓	*LM*↓ *or RM*↑	*LM*↓	-----
	Right	*LM*↑	-----	*RM*↓ *& RT*↓	*RT*↓	-----	*RT*↓	----

TABLE III

PERCENTAGE DISTRIBUTION OF SENSOR PLACEMENT ACCORDING TO GAIT EVENT TRIGGERED

	Ipsilateral Foot			Contralateral Foot		
	Heel	Mid-foot	Toe	Heel	Mid-foot	Toe
Loading Response	7- 54%	3- 23%	1- 8%	1-8%	1-8%	---
Mid-Stance	---	2- 14%	1 –7%	---	6- 43%	5- 36%
Terminal Stance	5- 50%	2- 20%	1- 10%	---	2- 20%	---
Pre-Swing	1- 6%	1- 6%	---	8- 50%	3- 19%	3- 19%
Initial Swing	---	6- 50%	4- 33%	---	2- 17%	---

system was plotted versus the mean times of each event (measured from the elapsed time since the start of the walking trial). Horizontal lines were then drawn representing the mean of all the differences and $+/-1$ and $+/-2$ standard deviations about the mean. Terminal stance exhibited the largest mean difference and standard deviation between the FSR detected event and that determined by the VICON, while initial swing exhibited the smallest deviation and preswing the smallest mean difference. There was no observable trend in the differences in the FSR and VICON times with respect to the time elapsed since the beginning of each trial (as represented by the x-axis of graphs of Fig. 2).

C. Reliability

Of 642 steps analyzed across the seven study subjects, there were 35 steps (5.5%) that included missed event detections. In all cases, a missed event caused subsequent events for that step to be missed since the program was waiting for the missed event to occur. Correct detection then was always reestablished on one of the next three steps when the conditions for the earlier missed event were satisfied. Approximately 80% of the misses self-corrected on the next step, while the remaining 20% self-corrected on the second or third step. Of the 35 steps involving missed event detections, 28 (80%) occurred when waiting for the loading response FSR condition, six (17%) occurred while waiting for the initial swing condition, and one (3%) occurred while waiting for terminal stance condition.

Subjects AF and AL generated 37% and 23% of all missed detections, respectively. Subjects AM, CW, DW, and SW each generated between 9–11% of all missed detections. Subject PL did not generate any missed detections. Expressed as a percentage of the total steps each child took, missed event detections for four subjects (AF, PL, CW, and SW) involved less than 5% of their total steps. For two subjects (AM, DW), about 7.5% of their steps involved a missed detection, and for one subject (AL), 12% of his steps involved a missed detection. There did not appear to be a correlation between the percentage of missed detections and walking ability according the Gross Motor Function Classification Scale (Table I).

V. DISCUSSION

The purpose of this paper was to report the reliability and accuracy of FSR-controlled ES for children with CP. In terms of reliability, for 94.5% of steps, there were no errors in gait event detection. When an error did occur, recovery was within three steps. In terms of the accuracy data, while mean differences between the FSR-detected events and that of the standard (VICON) were within 55 ms, there were wide distributions in the differences that appeared to be due in part to both intrasubject (step-to-step) and intersubject variability (Fig. 2).

To some degree, inherent differences between the techniques of measure (plantar pressures for the FSRs versus joint motion of the VICON system) may have contributed to the timing differences. In particular, during initial contact and initial swing (foot off), the VICON-measured events may have consistently been visualized before or after plantar pressures were realized, respectively. Data from force plates would perhaps provide a more appropriate comparison. However, because six of the seven children in this study used assistive devices, in our experience it would have been difficult if not impossible to collect foot contact data without the assistive device's contacting the plates.

The transition into terminal stance was the most difficult to accurately detect with FSRs. The mean difference and standard deviation were largest for this event (Fig. 2). For one subject,

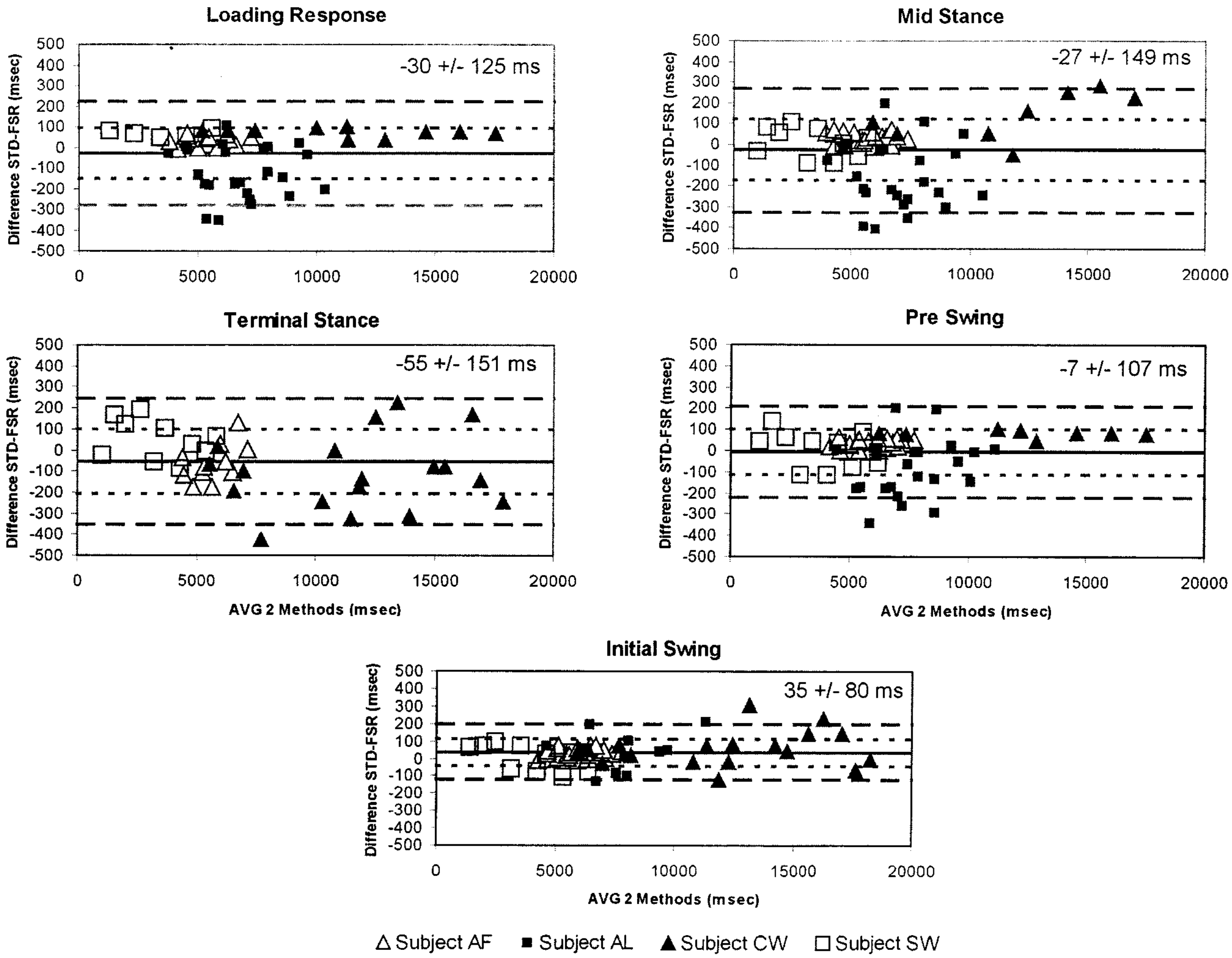

Fig. 2. For each gait event, the differences between the time the FSR detection scheme and the VICON system registered an event are plotted verses the mean times of each event (as elapsed since the start of the walking trial and computed from the values obtained with both measurement systems). All data points from the four subjects are included. The solid horizontal line denotes the mean difference and the short and long dashed horizontal lines denote the first and second standard deviations, respectively. The mean difference and first standard deviation for each gait event are also annotated in the upper right corner of each graph. For subject AL, terminal stance was not programmed because corresponding plantar pressure transitions were not apparent.

terminal stance was not programmed because corresponding plantar pressure transitions were not apparent. These difficulties were due to the absence of heel rise that marks the beginning of terminal stance. This affected the ability both to determine the actual event from the VICON system and to find an FSR location that would isolate either a decreased pressure on the back foot or an increased pressure toward the front of the foot. Since the opposite leg is in terminal swing at that time, there is no contralateral plantar pressure available to aid in terminal stance detection. Terminal stance is not only important for timing plantar flexion stimulation to improve pushoff but also could be critical to identify terminal swing of the opposite leg to trigger ES of the knee and hip extensors to prepare for loading response. In two cases, terminal stance could not be identified and loading response of the opposite leg was used. This timing was our only viable alternative but was less than desirable due to the inherent delay it introduces in the initiation of ES to the hip and knee extensor muscles.

Throughout the study, no subject fell or appeared to be in danger of falling when walking with ES even when gait events went undetected and stimulation was unchanged for one or two steps. Six of the seven subjects used assistive devices that may have aided them in adjusting to ill-timed stimulation due to a missed or delayed gait event detection.

The plantar pressure map of both feet over the entire gait cycle was very useful to determine FSR locations, as plantar pressures were spatially discrete and unique to each subject. During double limb support, it allowed us to link changes in plantar pressures of one foot to gait events of the other leg. Forty two percent of gait events were detected with plantar pressure of the opposite foot. In these instances, standard definitions relating the simultaneous events of each leg for typical gait were applied [2]. Principally, the loading response and initial swing transitions of one leg were used to predict the preswing and midstance transitions of the other leg, respectively. Contralateral FSR control provided a reference for these preswing and midstance transitions that otherwise would be difficult if not impossible to identify. Since transitions into loading response and initial swing are characterized by initial foot contact and when the foot first leaves the floor, respectively, the corresponding FSR locations are typically identifiable even with atypical plantar pressures. Conversely, to use ipsilateral FSRs to detect transitions into midstance and preswing, one would need to identify the typical plantar pressure transition from the hind to forefoot and an increased pressure on the metacarpal joints, respectively [2]. These ipsilateral transitions were not consistently identifiable from the plantar pressure analyses.

In terms of triggering ES for children with CP, the use of the contralateral transitions into loading response and initial swing of one leg to determine preswing and midstance of the other leg, respectively, seems appropriate. Initial contact portends the transition of weight onto the control limb (loading response) so that the opposite, stimulated limb must be preparing for swing phase. At this time, ankle dorsiflexor stimulation is initiated and remaining ankle plantar flexor stimulation is terminated. Similarly, Gracanin [7] used initial contact of one foot to initiate peroneal nerve stimulation of the other foot to improve limb clearance. Likewise, initial swing is initiated as the foot is just lifted from the floor so that the opposite leg has just entered full single limb support. At this moment, the transition into ankle plantar flexor stimulation can be initiated and knee extension stimulation can be terminated for the support limb.

The most common detection error (80%) was due to the failure of an FSR signal to reach the programmed threshold detection level during the transition into loading response. This event corresponds to initial contact of the foot with the floor. This error is likely a reflection of the step-to-step variability in the landing position assumed by the foot. Depending upon the spatial variability in the landing area, this detection error could be reduced by using an FSR with a larger surface area or by identifying two or three likely landing positions, each covered with an FSR. In the case of several FSRs, a logic OR condition would be used to determine the first FSR to sense pressure.

We could find no other studies that formally examined FSR reliability and accuracy for ES control in the population with CP with or without ES enabled. However, several studies have evaluated the use of FSRs for gait event detection in the population with spinal cord injury (SCI). Skelly [23] used FSRs applied to a fuzzy logic rule base for gait event detection. Using video to determine actual gait events, errors of $\pm 12\%$ or less of the gait cycle duration were reported for heel strike and toe off. Pappas and colleagues [24] used three FSRs located on the heel and medial and lateral forefoot together with a miniature gyroscope attached to the shoe heel to measure angular acceleration of the foot. Using their system with six subjects with impaired walking ability, including incomplete SCI, they reported average time differences of 70 ms or less in the detection of stance, heel off, swing and heel strike gait phase transitions using the VICON system as the measurement reference and found the detection system to be robust to different walking surfaces.

A better understanding of when to apply stimulation while walking would help to focus the requirements for an ES control system for this population. With typical walking, the initiation and relaxation of muscle activity is closely related to the standard gait events [2]. For the child with CP, muscle activity can be prolonged, premature, continuous, absent, or diminished [2]. Stimulation to aid walking then may be beneficial earlier than is typical to prompt voluntary muscle contractions, later than the initiation of voluntary activity to supplement force, or simultaneous with voluntary effort. Thus, there may be intermediate points between the standard gait events that may be important to identify for ES control. To date, stimulation timing has been applied to mimic EMG activity of an immature gait pattern [5], adjusted based on the degree of spasticity [7] and superimposed at times that would be considered "typical" [9], [15], [20].

Knowledge of the contraction and relaxation times of CP muscle in the presence of stimulation will also help to determine the timing and sensitivity of muscle stimulation during gait. There is evidence to suggest that the structure of skeletal muscle in children with CP is different from typically developing muscle [25]. Specifically, several studies suggest that for children with CP who are ambulatory (like those of this study), there is a predominance of Type I muscle fibers (slow twitch) [26], [27]. If so, one would expect that the times to both achieve increased muscle forces and to relax the muscle following stimulation may be relatively long. This would suggest that the anticipation of gait events may be critical to achieve timely stimulation.

More sophisticated techniques of rule-based event detection such as machine learning [28] or fuzzy logic [29] may reduce the difference between the FSR-detected and actual gait events realized using the simple threshold detection method of this study. Such techniques could reduce the sensitivity to noise or to small variations in the FSR signal from one step to another. Ultimately, it would be advantageous to have a state controller that is akin to a gait *cycle* detector that would provide a continuous output proportional to the actual gait cycle.

VI. Conclusion

In this paper, we applied intramuscular ES to seven children with spastic diplegia, CP, using FSRs to identify the transitions between five gait phases and thus trigger electrical stimulation. The accuracy and reliability of the FSR gait event detection schemes were examined. Accuracy was evaluated with four of the subjects by synchronizing the output of the FSR detection scheme with a VICON motion-analysis system, which was used as the gait event reference. While mean differences ranged from +35 ms (indicating that the FSR detection scheme recognized the event before it actually happened) to −55 ms (indicating that the FSR scheme recognized the event after it occurred), the wide distributions in the data reflect both intrasubject (step-to-step) and intersubject variability. To determine step-to-step reliability, each of the seven subjects walked on a level walkway for at least 50 steps. Of 642 steps, there were 35 steps that involved detection errors (5.5%). Eighty percent of detection errors were due to the failure of the FSR signal to reach the programmed threshold level during the transition to loading response.

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