Functional Electrical Stimulation Changes Dynamic Resources in Children With Spastic Cerebral Palsy

Background and Purpose. Children with cerebral palsy (CP) often are faced with difficulty in walking. The purpose of this experiment was to determine the effects of functional electrical stimulation (FES) applied to the gastrocnemius-soleus muscle complex on the ability to produce appropriately timed force and reduce stiffness (elastic property of the body) and on stride length and stride frequency during walking. Subjects and Methods. Thirteen children with spastic CP (including 4 children who were dropped from the study due to their inability to cooperate) and 6 children who were developing typically participated in the study. A crossover study design was implemented. The children with spastic CP were randomly assigned to either a group that received FES for 15 trials followed by no FES for 15 trials or a group that received no FES for 15 trials followed by FES for 15 trials. The children who were having typical development walked without FES. Kinematic data were collected for the children with CP in each walking condition and for the children who were developing typically. Impulse (force-producing ability) and stiffness were estimated from an escapement-driven pendulum and spring system model of human walking. Stride length and stride frequency also were measured. To compare between walking conditions and between the children with CP and the children who were developing typically, dimensional analysis and speed normalization procedures were used. Results. Nonparametric statistics showed that there was no significant difference between the children with CP in the no-FES condition and the children who were developing typically on speed-normalized dimensionless impulse. In contrast, the children with CP in the FES condition had a significantly higher median value than the children who were developing typically. The FES significantly increased speed-normalized dimensionless impulse from 10.02 to 16.32 when comparing walking conditions for the children with CP. No significant differences were found between walking conditions for stiffness, stride length, and stride frequency. Discussion and Conclusion. The results suggest that FES is effective in increasing impulse during walking but not in decreasing stiffness. The effect on increasing impulse does not result in more typical spatiotemporal gait parameters. [Ho CL, Holt KG, Saltzman E, Wagenaar RC. Functional electrical stimulation changes dynamic resources in children with spastic cerebral palsy. Phys Ther. 2006;86:987–1000.]

Key Words: Cerebral palsy, Dynamic systems, Electrical stimulation, Locomotion.

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Children with cerebral palsy (CP) are a commonly treated group seen by pediatric physical therapists. Spasticity is the most prevalent form of motor dystonia.1 Attributing spasticity as a major contributor to movement dysfunction may be a dated concept in the literature.2 However, the concept is still relevant to most clinicians and is the premise of some treatment methods (eg, posterior selective rhizotomies). Various walking problems can accompany spasticity such as poor power generation during the push-off period,3 decreased peak force production during the late stance phase,4 decreased maximal voluntary contractions of the ankle plantar flexors,5 changes in timing and sequencing of muscle activation,6 and decreased walking speed with higher cadence and shorter stride length.7 Kinematic gait changes such as toe-walking and a running-like gait pattern are commonly observed in children with spastic CP.8 Changing the gait patterns of children with spastic CP without invasive surgeries and neurochannel blockage is challenging. Even with these interventions, prolonged positive effects on gait have not been shown.

The evidence available to support current physical therapy practices for improving locomotor function in children with spastic CP is limited and equivocal.9,10 In recent developments, one type of intervention, functional electrical stimulation (FES), has been more widely used in clinical settings and has received more attention in the research literature.11–14 Clinical research and clinical practice have focused on correcting atypical gait patterns by using electrical stimulation. For example, attempts to increase ankle dorsiflexion by applying neuromuscular electrical stimulation to the tibialis anterior muscle have met with limited success.15–18 In contrast, the findings of a study by Carmick11 suggested that gait improvement such as decreased excessive ankle plantar flexion occurred in 3 children with spastic CP when electrical stimulation was applied only to the gastrocnemius-soleus muscle complex (G-S) during walking. In a study of the application of FES on the G-S with a larger sample (N=14), Comeaux et al13 showed positive effects such as increased ankle range of motion and ankle dorsiflexion at initial foot contact when compared with walking without FES. It seems paradoxical that stimulation of the ankle plantar flexors during the period they would normally be active results in an improvement in dorsiflexion at initial contact. However, the results must be accepted with caution because neither study systematically evaluated the effects of FES on gait parameters using randomized control trials.

One possible reason that application of FES to the G-S may be effective in improving the gait patterns is related to the dynamic resources available to children with spastic CP for locomotion. Dynamic resources refer to an individual’s capability to generate, dissipate, or conserve the energy required to locomote.19–25 The elements include the active contractile component of muscles that are capable of producing forces, elastic elements of soft tissues such as muscles and tendons that are capable of storing and returning elastic potential energy, viscous...
tissue properties for passively dissipated energy, rigid bony structures and body masses that are capable of storing gravitational potential energy and exchanging potential energy (elastic and gravitational) and kinetic energy during walking, and neural circuitry that delivers graded and timed muscle input. Dynamic resources of locomotion include the capability to actively produce appropriately timed muscle forces and to store elastic energy through soft tissues. Dynamic resources have been operationalized using abstract biomechanical models that treat the body as a force-driven or escapement-driven global pendulum and spring. Total forces during the push-off phase (impulse) and stiffness of the tissues were estimated from the escapement-driven pendulum and spring model of locomotion.

Children with spastic hemiplegic CP show an increase in stiffness and a decrease in force during push-off for the affected limbs, while the nonaffected limbs show greater force during push-off and the same stiffness compared with children who were developing typically. Holt et al. argued that children with spastic hemiplegic CP adapt their walking patterns to optimize the utilization of the different dynamic resources available to them, for example, by adopting a running-like pattern that maximizes the use of increased musculoskeletal stiffness. Thus, children with spastic hemiplegic CP have a shorter step length on the affected side due to the decrease in appropriately timed force production and an increase in step (and stride) frequency due to the increased stiffness when compared with children with typical development.

In addition, Fonseca et al. showed that children with hemiplegic CP showed an asymmetric gait pattern in which the nonaffected limb raises the center of mass (COM) in a pendulum-like fashion and “drops” it onto the affected limb that absorbs and returns elastic energy in a similar way as a pogo stick. Furthermore, equinus gait, in which the foot is plantar flexed at initial contact, is viewed as an adaptation to the weakness of the G-S. The pattern allows children with spastic CP to take advantage of the stiffness of the soft tissues in the leg to store elastic energy. Thus, the atypical pattern may be adaptive and “normal” in the sense that it takes advantage of the available resources (see Latash and Anson). Holt and colleagues have claimed that the relatively more consistent positive effects of FES applied to the G-S compared with the equivocal results of attempts to correct gait patterns by FES applied to the Tibialis anterior muscle are because FES applied to the G-S addresses the decrease in the dynamic resource, namely the appropriately timed force production of the G-S during gait. Holt and colleagues proposed that, by providing the child with appropriately timed stimulation, the need for gait pattern adaptations such as a plantar-flexed foot or running-like gait that facilitate the use of greater stiffness may no longer be necessary.

The purpose of this study was to investigate the immediate effects of FES applied to the G-S. We hypothesized that FES applied to the G-S mimics the appropriate grading and timing of neural dynamic resources, an escapement-forcing function that is similar to that used by children who were developing typically, with stride parameters such as stride length and stride frequency changing accordingly. Specifically, we predicted that FES would increase the force production, as measured by impulse, of the limbs during the push-off phase and that there would be an associated decrease in the system stiffness. We predicted that the changes in resource utilization would be accompanied by an increase in stride length and a decrease in stride frequency, respectively, at any particular speed. We also predicted that the FES intervention would lead to values of impulse, stiffness, stride length, and stride frequency that are similar to those found in children who were developing typically.

Method

Subjects

Thirteen children between 3 and 11 years of age were recruited from the patient population at the outpatient clinic of the Physical Medicine and Rehabilitation Department of New England Medical Center (NEMC) in Boston, Mass. The following inclusion criteria were met: (1) age range of 3 to 12 years, (2) physician’s diagnosis of spastic-type CP, (3) mildly involved CP with a Modified Ashworth Scale score of 3 or less, (4) able to ambulate independently without an assistive device or orthoses, (5) unable to achieve heel-strike at initial foot contact at a comfortable or fast walking speed, (6) no cardiovascular diseases, (7) no surgery within the previous 24 months, (8) no sensory defensiveness, and (9) ability to follow instructions. Four subjects with CP were subsequently dropped from the study due to their inability to cooperate; thus, data for 9 children with CP were included in the final analysis. Six children who were developing typically were recruited from the Boston University community as a control group. The inclusion criteria for this group were: (1) age range of 3 to 12 years and (2) no known musculoskeletal, neurological, or cardiopulmonary diseases. The parents or guardians signed informed consent forms. Children above the age of 7 years also signed an assent form.

Study Design

A crossover design with a single factor (FES) was implemented. Children with CP were randomly assigned to either a group that walked with FES for 15 trials followed by no FES for 15 trials or a group that walked without...
FES for 15 trials followed by FES for 15 trials. The children who were developing typically walked without FES. The control group walked 30 trials without FES. All trials were conducted in one experimental session.

Instrumentation
We used a Respond II Select* unit to deliver the electrical stimulation. The unit has 2 channels that allow stimulation of the G-S bilaterally in a reciprocal manner for children with diplegia. Footswitches composed of 3 capacitive pressure sensors‡ triggered the electrical stimulation by a customized program written in LABVIEW, version 5.† Reusable adhesive electrodes were used. Electrodes of a variety of sizes were used to accommodate the individual child’s limb size.

Three-dimensional (3-D) kinematic data were collected by the OptoTrak 3020 system§ and processed using custom-written programs in MATLAB, version 5. The calibrated viewing volume of 3 sets of serially connected cameras was 3 m in length. Each bank with 3 individual cameras had root mean square accuracy to 0.1 mm and resolution to 0.01 mm. The subjects in all 3 groups walked approximately 3 m prior to entering the viewing volume. Up to 5 strides of data were collected as the subjects walked toward the cameras. An OptoTrak Data Acquisition Unit† was used to collect the footswitch analog data and synchronize the data with the position data.

Testing Procedure
After introducing the subjects and their parents to the experimental environment, a physical therapist measured the subjects’ anthropometric characteristics, including body height, weight, and leg length; estimated the involvement level of the lower-extremity muscles using the Modified Ashworth Scale28; and documented the involvement level of the lower-extremity muscles.

Children walked overground through the viewing volume at their self-selected speed for 30 trials. The number of trials for 2 children with CP was decreased due to physical tiredness and an unwillingness to continue. The FES was triggered by the foot’s initial contact with the ground by any footswitch sensor, remained on when the foot was in contact with the ground, and turned off when all the sensors lost contact with the ground. Thus, FES provided an escapement during the stance phase of the gait cycle, a posturally state-dependent source of muscle force. The kinematic data were collected at 100 Hz. Data of 2 to 6 strides were collected during each trial, which lasted 5 to 10 seconds depending on each participant’s preferred walking speed. The total data collection period was 20 to 30 minutes for each subject.

Data Reduction
Kinematic data were interpolated for consecutively missing position data of less than 15 frames and filtered through a fourth-order Butterworth filter with a cutoff frequency of 5 Hz. This interpolation procedure was previously validated for children with CP.24 Two dependent variables, impulse and stiffness, and spatiotemporal parameters (stride length, cadence, and walking speed) were measured.

In this study, the effects of FES were quantified using the escapement-driven inverted pendulum with spring and viscous damping (EDIPS) model25–25 (Fig. 1). The equa-

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* Medtronic Inc, 710 Medtronic Plsw, Minneapolis, MN 55432-5604.
† Interlink Electronics Inc, 546 Flynn Rd, Camarillo, CA 93012.
‡ National Instruments Co, 11500 N Mopac Expwy, Austin, TX 78759-3504.
§ Northern Digital Inc, 103 Randall Dr, Waterloo, Ontario, Canada N2V 1C5.
† The MathWorks Inc, 3 Apple Hill Dr, Natick, MA 01760-2098.
A complete stride is defined as consecutive initial foot contacts of the same foot. Stride length is defined as the distance traveled by the body’s COM during a complete stride. Initial foot contacts were estimated by a modified linear kinematic algorithm that used the position data from the foot-segment COM but not from the heel. Measurement error for initial contact was estimated in a pilot study by comparing the algorithm results with force-plate measurements. Mean error was 0.04 second. Stride length was measured from time plots of the forward displacement of the right ankle marker, and stride duration was determined as the duration of a complete stride. Cadence, measured as steps per minute, was estimated from the stride frequency (ie, inverse of the stride duration). Walking speed was calculated by multiplying the stride frequency by the stride length. Stride length and stride duration were measured for each stride within each trial. Intrasubject means of each stride parameter were obtained by averaging over at least 20 strides for each FES and no-FES condition.

**Dimensional Analysis and Speed Normalization**

Gait parameters are very much influenced by a person’s limb lengths and body mass. Therefore, to compare the walking dynamics among subjects, the effects of these anthropometric differences must be removed by a method called dimensional analysis. Dimensional analysis was applied to the dynamic resources and spatiotemporal parameters. The dimensionless form of impulse is calculated based on the method proposed by Hof:

\[ I = \frac{m}{L \sqrt{l/g}} \]

where \( I \) is the impulse, \( m \) is body mass minus stance foot mass, \( L \) is pendulum equivalent length, and \( g \) is gravity. The dimensionless form of stiffness is calculated based on Holt and colleagues’ model:

\[ k = \frac{m}{L \sqrt{l/g}} \]

where \( k \) is stiffness coefficient. We calculated dimensionless forms of walking speed, stride length, and stride frequency based on the methods of Alexander and Wagenaar and Beek:

\[ u = \frac{u}{\sqrt{g}} \]

\[ \dot{f} = f \frac{1}{\sqrt{g}} \]

where \( u \) is dimensionless walking speed, \( \dot{f} \) is dimensionless stride frequency, and \( l \) is leg length.

Impulse and stiffness increase in a linear fashion with increase in walking speed. Therefore, in order to compare between conditions for children with CP and to compare children with CP and children who were developing typically at an equivalent speed, we normalized dimensionless forms of impulse and stiffness by dividing them by dimensionless walking speed (\( \dot{u} \)) and obtained the speed-normalized dimensionless forms of impulse and stiffness. The speed-normalized dimensionless forms of stride length and stride frequency were calcu-
related by dividing each by square root of dimensionless speed \((\dot{u})^{0.5}\). This procedure was validated in 68 children who were developing typically (mean age = 7.14 years, SD = 2.58, range 3–12). We found significant linear regression models between the dimensionless stride length and the square root of dimensionless speed \((\dot{u})^{0.5}, P < .05, r^2 = .52\) and between the dimensionless stride frequency and the square root of dimensionless speed \((\dot{f} = 0.39(\dot{u})^{0.5}, P < .05, r^2 = .28\). Wagenaar and Beek\(^38\) reported similar relationships among dimensionless stride length, stride frequency, and walking speed for different populations such as adults who were healthy and patients with stroke. To quantify the change of dimensionless stride length relative to dimensionless stride frequency, the ratio between speed-normalized dimensionless stride length and speed-normalized dimensionless stride frequency was calculated. The ratio represents a relationship between stride length and frequency at an equivalent speed and thus enables comparisons between FES conditions and between subjects with CP and subjects who are developing typically.

### Data Analysis

Dependent variables (ie, nonadjusted, dimensionless form and speed-normalized dimensionless form of impulse, stiffness, stride length, stride frequency, and speed) were measured for at least 20 strides for each condition for each individual. Descriptive statistics, including intrasubject mean and standard deviation, were obtained for the subjects who were developing typically and for the no-FES and FES conditions of the children with CP. Eighteen measurements that fell outside of 6 standard deviations of the mean were considered outliers and were excluded from further analysis. Outlying data points resulted from sudden loss of the relevant markers from bony landmarks or unexpected blockage of the markers from cameras. Intrasubject means were used to generate group medians for the subjects who were developing typically and for the no-FES and FES conditions of the children with CP. Eighteen measurements that fell outside of 6 standard deviations of the mean were considered outliers and were excluded from further analysis.

### Table 1.

Subject Characteristics

<table>
<thead>
<tr>
<th>Group</th>
<th>Subject No.</th>
<th>Age (y)</th>
<th>Sex</th>
<th>Body Weight (kg)</th>
<th>Body Height (m)</th>
<th>Distribution of Involvement</th>
<th>Lower-Extremity Ashworth Scale Score</th>
<th>GMFCS Level</th>
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<tbody>
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<td>TD</td>
<td>N1</td>
<td>4</td>
<td>Female</td>
<td>16.34</td>
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<td>N2</td>
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<td>Male</td>
<td>22.70</td>
<td>1.24</td>
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<td>1.16</td>
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<td>N4</td>
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<td>Male</td>
<td>23.40</td>
<td>1.23</td>
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<td>0</td>
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<tr>
<td></td>
<td>N5</td>
<td>10</td>
<td>Female</td>
<td>43.42</td>
<td>1.40</td>
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<td>0</td>
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<tr>
<td></td>
<td>N6</td>
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<td>Male</td>
<td>48.57</td>
<td>1.49</td>
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<tr>
<td>OX</td>
<td>1</td>
<td>7</td>
<td>Male</td>
<td>18.59</td>
<td>1.13</td>
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<td>L: 3</td>
<td>I</td>
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<tr>
<td></td>
<td>2</td>
<td>4</td>
<td>Female</td>
<td>14.74</td>
<td>0.99</td>
<td>Diplegia</td>
<td>L: 2</td>
<td>I</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>7</td>
<td>Female</td>
<td>22.22</td>
<td>1.16</td>
<td>Hemiplegia</td>
<td>L: 0</td>
<td>I</td>
</tr>
<tr>
<td></td>
<td>4</td>
<td>7</td>
<td>Male</td>
<td>22.68</td>
<td>1.16</td>
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<tr>
<td></td>
<td>5</td>
<td>11</td>
<td>Male</td>
<td>24.94</td>
<td>1.29</td>
<td>Hemiplegia</td>
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<tr>
<td>XO</td>
<td>6</td>
<td>11</td>
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<td>39.91</td>
<td>1.49</td>
<td>Diplegia</td>
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<td>I</td>
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<tr>
<td></td>
<td>7</td>
<td>6</td>
<td>Male</td>
<td>18.00</td>
<td>0.96</td>
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<td>I</td>
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<tr>
<td></td>
<td>8</td>
<td>7</td>
<td>Female</td>
<td>30.87</td>
<td>1.20</td>
<td>Diplegia</td>
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<td>I</td>
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<tr>
<td></td>
<td>9</td>
<td>8</td>
<td>Male</td>
<td>22.25</td>
<td>1.21</td>
<td>Hemiplegia</td>
<td>L: 0</td>
<td>I</td>
</tr>
</tbody>
</table>

\(^a\) TD = subjects who were developing typically, OX = subjects without spastic cerebral palsy who walked with functional electrical stimulation (FES) for 15 trials followed by no FES for 15 trials, XO = subjects with spastic cerebral palsy who walked with FES for 15 trials followed by no FES for 15 trials, GMFCS = Gross Motor Function Classification System, N/A = not applicable, L = left, R = right.
Table 2.
Descriptive Statistics of the Dynamic Resources and the Gait Spatiotemporal Parameters for the Children Who Were Developing Typically and the Children With Spastic Cerebral Palsy (CP) in the No-Functional Electrical Stimulation (FES) and FES Conditions

<table>
<thead>
<tr>
<th></th>
<th>Children With CP (n=9)</th>
<th></th>
<th>FES</th>
<th></th>
<th>Children Developing Typically (n=6)</th>
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<tr>
<td></td>
<td>No-FES</td>
<td>FES</td>
<td></td>
<td></td>
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<tr>
<td></td>
<td>Nonadjusted variables</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Impulse (N×s)</td>
<td>79.53 (44.25)</td>
<td>65.08 (31.31)</td>
<td>79.69 (44.25)</td>
<td></td>
<td>79.69 (51.92)</td>
<td>31.09–177.81</td>
</tr>
<tr>
<td>Stiffness (N×m)</td>
<td>735.92 (370.81)</td>
<td>718.02 (375.83)</td>
<td>1,009.79 (664.39)</td>
<td></td>
<td>1,229.69 (664.39)</td>
<td>536.24–2,209.38</td>
</tr>
<tr>
<td>Stride length (m)</td>
<td>0.81 (0.30)</td>
<td>0.76 (0.21)</td>
<td>1.09 (0.20)</td>
<td></td>
<td>1.10 (0.20)</td>
<td>0.90–1.46</td>
</tr>
<tr>
<td>Cadence (steps/min)</td>
<td>120.89 (14.01)</td>
<td>118.61 (12.46)</td>
<td>121.20 (12.46)</td>
<td></td>
<td>113.79–136.87</td>
<td></td>
</tr>
<tr>
<td>Speed (m/s)</td>
<td>0.73 (0.30)</td>
<td>0.76 (0.21)</td>
<td>1.10 (0.20)</td>
<td></td>
<td>1.15 (0.20)</td>
<td>0.92–1.47</td>
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<td></td>
<td>Dimensionless variables</td>
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<tr>
<td>Impulse (N×s)</td>
<td>3.04 (0.61)</td>
<td>5.18 (2.79)</td>
<td>2.98 (1.42)</td>
<td></td>
<td>2.22–5.40</td>
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<tr>
<td>Stiffness (N×m)</td>
<td>4.29 (0.47)</td>
<td>4.02 (0.48)</td>
<td>4.76 (0.45)</td>
<td></td>
<td>4.15–5.31</td>
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<tr>
<td>Stride length (m)</td>
<td>1.28 (0.33)</td>
<td>1.32 (0.22)</td>
<td>1.70 (0.22)</td>
<td></td>
<td>1.38–1.94</td>
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<tr>
<td>Stride frequency</td>
<td>0.25 (0.02)</td>
<td>0.24 (0.02)</td>
<td>0.27 (0.02)</td>
<td></td>
<td>0.24–0.30</td>
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<tr>
<td>Speed (m/s)</td>
<td>0.31 (0.10)</td>
<td>0.32 (0.07)</td>
<td>0.45 (0.06)</td>
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<td>0.37–0.54</td>
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<tr>
<td>Impulse</td>
<td>10.02 (3.83)</td>
<td>16.50 (7.19)</td>
<td>7.49 (2.18)</td>
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<td>5.14–10.37</td>
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<tr>
<td>Stiffness</td>
<td>9.74 (1.95)</td>
<td>9.74 (1.52)</td>
<td>8.15 (0.98)</td>
<td></td>
<td>7.28–9.93</td>
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</tr>
<tr>
<td>Stride length</td>
<td>2.26 (0.21)</td>
<td>2.35 (0.15)</td>
<td>2.51 (0.20)</td>
<td></td>
<td>2.21–2.80</td>
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<tr>
<td>Stride frequency</td>
<td>0.45 (0.03)</td>
<td>0.42 (0.02)</td>
<td>0.40 (0.03)</td>
<td></td>
<td>0.36–0.45</td>
<td></td>
</tr>
<tr>
<td>Stride length and stride frequency ratio</td>
<td>5.17 (1.01)</td>
<td>5.57 (0.74)</td>
<td>6.28 (1.02)</td>
<td></td>
<td>4.90–7.86</td>
<td></td>
</tr>
</tbody>
</table>
Values Obtained by Wilcoxon Signed Rank Tests

developing typically had significantly lower median speed-
condition (P=.37). In contrast, the children who were
devolving typically showed significantly
higher median dimensionless stiffness than the children
with CP compared with the no-FES condition in
FES condition (P=.05). Functional electrical
stimulation did not significantly reduce stiffness of the
children with CP compared with the no-FES condition in
either nonadjusted or adjusted (dimensionless, speed-
normalized dimensionless) data (Tab. 3).

**Stiffness**

There were no significant differences in nonadjusted data for stiffness
between the children who were developing typically and the children with CP in either the FES condition (P=.11) or the no-FES condition (P=.14) The children who were developing typically showed significantly higher median dimensionless stiffness than the children with CP in the FES condition (P=.02). There was no significant difference in either dimensionless stiffness (P=.08) or speed-normalized dimensionless stiffness (P=.14) when the children who were developing typically were compared with the children with CP in the
no-FES condition. In contrast, the children who were
developing typically had significantly lower median speed-
normalized dimensionless stiffness than the children with CP in the no-FES condition (P=.05). Functional electrical
stimulation did not significantly reduce stiffness of the
children with CP compared with the no-FES condition in
either nonadjusted or adjusted (dimensionless, speed-
normalized dimensionless) data (Tab. 3).

**Stride Length**

The children who were developing typically showed significantly longer median stride length than the children with CP in either the FES condition (P=.02) or the no-FES condition (P=.03) in the nonadjusted data. Similar results were found in either the FES condition (P=.02) or the no-FES condition (P=.05) in the dimensionless data. There was no significant difference in speed-normalized dimensionless stride length between the children who were developing typically and the children with CP in either the FES condition (P=.11) or the no-FES condition (P=.37). There was no significant

**Table 3.**

<table>
<thead>
<tr>
<th></th>
<th>Median</th>
<th>Range</th>
<th>P*</th>
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</thead>
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<tr>
<td>Nonadjusted variables</td>
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<td></td>
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<tr>
<td>Impulse (N·s)</td>
<td>−6.16</td>
<td>−37.54−9.66</td>
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</tr>
<tr>
<td>Stiffness [N·m]</td>
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<td>−67.61−35.34</td>
<td>.12</td>
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<tr>
<td>Stride length (m)</td>
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<td>−0.29−0.09</td>
<td>.33</td>
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<td>Cadence (steps/min)</td>
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<td>−13.97−4.32</td>
<td>.20</td>
</tr>
<tr>
<td>Speed (m/s)</td>
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<td>−0.29−0.12</td>
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<td>Dimensionless variables</td>
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<tr>
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<tr>
<td>Stride frequency</td>
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<tr>
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<td>Stride frequency</td>
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<td>1.00</td>
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<td>Stride length and stride frequency ratio</td>
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<td>−1.3−0.57</td>
<td>.57</td>
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</table>

*P* values smaller than alpha value are presented in bold type.

with those of the subjects who were developing typically.
To assess the effects of FES, the Wilcoxon signed rank
test was used to compare the dependent variable
values of the children with CP between walking
conditions. Alpha level was set at .05 for all the analyses.
Statistical analysis was performed by using SAS soft-
ware, version 5.*

**Results**

Subjects’ characteristics are presented in Table 1. All
children with CP were able to walk comfortably in the
FES condition. Table 2 summarizes the measurements
for impulse, stiffness, and the spatiotemporal parameters
for the children with CP in both walking conditions and
the children who were developing typically.

**Impulse**

There were no significant differences in impulse
between the children who were developing typically and the children with CP in either walking condition, in either
the nonadjusted data (FES, P=.68; no-FES, P=.85) or the
dimensionless data (FES, P=.21; no-FES, P=.85). There
was no significant difference in speed-normalized dimen-
sionless impulse between the children who were
developing typically and the children with CP in the no-FES
condition (P=.37). In contrast, the children who were
developing typically had significantly lower median speed-

* SAS Institute Inc., SAS Campus Dr, Cary, NC 27513-2414.
difference in stride length between walking conditions in either nonadjusted or adjusted data (Tab. 3).

**Stride Frequency**

There were no significant differences in cadence between the children who were developing typically and the children with CP in either the FES condition ($P=.21$) or the no-FES condition ($P=.51$). The children who were developing typically showed significantly higher median dimensionless stride frequency than the children with CP in the FES condition ($P=.05$). There was no significant difference when the children who were developing typically were compared with the children with CP in the no-FES condition ($P=.08$). No significant difference in speed-normalized dimensionless stride frequency was found between the children who were developing typically and the children with CP in either the FES condition ($P=.12$) or the no-FES condition.

**Figure 2.**

Dimensionless variables of (A) impulse, (B) stiffness, (C) stride length, (D) stride frequency, and (E) walking speed for the 6 children who were typically developing (TD) and each of the 9 children with spastic cerebral palsy CP in the no-functional electrical stimulation (no-FES) condition and the FES condition.
condition ($P=0.20$). There was no significant difference in stride frequency between walking conditions in either nonadjusted or adjusted data (Tab. 3).

Walking Speed
The children who were developing typically had significantly higher median walking speed than the children with CP in either the FES condition ($P=0.02$) or the no-FES condition ($P=0.04$). Similar results were found...
for dimensionless walking speed in either the FES condition ($P=.01$) or the no-FES condition ($P=.05$). There was no significant difference in walking speed between walking conditions in either nonadjusted or dimensionless data (Tab. 3).

**Stride Length:Stride Frequency Ratio**
Data points of dimensionless stride length and dimensionless stride frequency are shown in Fig. 4A for each participant. Data points of speed-normalized dimensionless stride length and speed-normalized dimensionless stride frequency were shown in Fig. 4B.

There was no significant difference in the ratio of stride length and stride frequency between the children who were developing typically and the children with CP in either the FES condition ($P=.11$) or the no-FES condition ($P=.37$). Ratios of the children with CP in either walking condition were not significantly different (Tab. 3).

**Discussion and Conclusions**
The results of this study indicate that there are discernable effects of FES on impulse of children with CP. The major finding is that FES successfully increases the impulse generated during the push-off phase of the gait cycle. However, translating that energy into increased speed and stride length and decreasing the adapted stiffness probably may require a longer period of training with FES than was used in this study. Nevertheless, the fact that immediate benefits of increased impulse, decreased stiffness, increased stride length, and decreased stride frequency were obtained in 3 out of 9 children (participants 2, 4, and 7) suggests that the use of FES in this manner warrants further investigation.

Predictions for an increase in impulse with FES were supported, but the expected decrease in stiffness was not supported. The largest median impulse was observed for children with CP in the FES condition in which the speed-normalized dimensionless values were more than twice of those of the children who were developing typically and more than 150% of those of the children with CP in the no-FES condition. This finding suggests that to walk at speeds comparable to those of children who were developing typically, children with CP require a much greater impulse. From a mechanical perspective, given that the stiffness does not change, this finding would be expected. Stiffer systems, by definition, require greater amounts of force to produce an equivalent amplitude of oscillation (displacement). Thus, the greater impulse observed in the FES condition does not translate into greater stride lengths (as a correlate of amplitudes of the motion of the COM). However, given the large difference between impulse (16.32) and stiffness (9.50) for the FES condition (Tab. 2) compared with the no-FES condition (10.02 and 9.74 for impulse and stiffness,
respectively), the failure to see any differences in stride length suggests that much of the increased impulse fails to transfer into increased motion of the COM. The possibility exists that the impulse is lost or dissipated in other ways through the body.

The predicted immediate effects of FES on decreased stiffness were not observed. In retrospect, this finding might have been expected. Some of the mechanisms responsible for increased stiffness, such as increased reflex gain, morphological changes resulting in shorter muscle bellies and longer tendons, and muscle fibrosis, are slowly developing adaptations. System stiffness of children with spastic CP that results from the underlying pathophysiology adapts gradually over time according to the demands of locomotion in the gravitational force field. Morphological changes that increase stiffness are particularly resilient to change, and probably take much longer to adapt. The observation that older children do not respond as well to stimulation of the G-S as children around 3 years of age may be due to the fact that the morphological changes are well established in this older population. Our children ranged in age from 3 to 12 years, and the individual differences of mean stiffness between the no-FES and FES conditions were quite different based on their age (Fig. 5). We suggest that age may be an important factor in the immediate responses to FES, particularly with respect to the stiffness parameter. Younger children may show greater change in stiffness, although we cannot speculate on why the youngest participant (participant 2) did not show the greatest change among all participants (Fig. 5). We suggest that future studies involving treatment sessions that last for weeks or even months may be needed in order to bring about significant decreases in the stiffness of children using FES.

Reliability of the anthropometric measurements (body weight, height, and segment lengths) did not affect our study results of the FES effects and thus was not tested. Our study used a repeated-measures design, and each child with spastic CP was his or her own control. Same values of measurements were used to normalize the nonadjusted variables. In addition, only one therapist did the measurements. Measurement errors are assumed to be the same for children who are developing typically and children with spastic CP. Besides, segment lengths used to calculate the pendulum equivalent length were derived from kinematic data. Measurement errors related with placement of makers are assumed to be consistent for children who are developing typically and children with spastic CP.

The small sample size of the current study and the large variability of the responses resulted in low levels of power. The power was 8.8%, 25.3%, 25.3%, and 9.8%, respectively, in the analyses of speed-normalized dimensionless stiffness, stride length, stride frequency, and stride length–to–stride frequency ratio. For example, 5 out of the 9 children with CP showed a decrease in speed-normalized dimensionless stiffness. Sample size estimation would suggest the need for 500 subjects to achieve a statistical power of 80%. The fact that impulse showed a significant increase despite the small number of participants points to the importance of the finding. The significance of FES intervention may be greater than suggested by the nonsignificant statistical outcome because of the lack of power in the analyses.

Our sample represented a cross-section of children with spastic CP. Besides the same gross motor function level as measured with the Gross Motor Function Classification System, their individual characteristics were quite different, including age, previous and current therapies, degree of involvements as measured by the Modified Ashworth Scale, and the number of limbs involved. One resolution to this issue of intersubject variation is the recruitment of homogeneous study groups based on the clinical presentation of the disorder, as has been suggested by Hur. The problem with this approach,
however, lies in the practical difficulty of recruiting large enough samples of homogeneous groups, and even in defining homogeneity. Another limitation is that such results could not be generalized beyond the specific groups; therefore, the research endeavor would have limited clinical value. Physical therapists must be able to identify those children who are most likely to benefit from this form of intervention and those who will not. To this end, our plan for future study is to include more children and to determine by cluster analysis the characteristics of those children who benefit from FES. With a larger sample size, we may identify what characteristics are likely indicative of positive FES effects. In particular, we are interested in the kinematic characteristics of gait patterns that would be easily observable by a trained clinician (eg, ankle plantar flexion, knee flexion). Wagenaar and Beek applied a similar approach to evaluate gait deficits in patients with stroke.

In order to progress during gait from a biomechanical perspective, the COM has to move toward a “goal.” To attain this goal, the body must supply the appropriate forces at the right time in the cycle that drives the COM in the appropriate direction. In walking, these forces can be generated by active muscle contraction and conserved by transferring forces between limb segments and between the body and the environment. Forces can be conserved in 2 ways: (1) through the passive elastic energy return of soft tissues such as tendons and the elastic elements in muscle and (2) by the ability of the individual to exchange energy from gravitational potential energy to kinetic energy. The fundamental claim is that a change in these dynamic resources due to disease (such as upper motor neuron disease) leads to an adaptation of movement patterns that takes advantage of the available resources. Conversely, if we were able through our interventions to enhance a limited resource, a more normal pattern might emerge that is energy efficient, stable, and perhaps more cosmetically appealing.

References
Appendix.

Impulse and Stiffness Estimation

A 7-segment model (head-arms-trunk, 2 thighs, 2 lower legs, and 2 feet) was used to calculate the three-dimensional coordinates of the total body center of mass (COM) from segment masses and the location of the segment COMs. Those parameters were obtained by Jensen’s regression equations.\(^*\)

The behavior of the escapement-driven inverted pendulum with spring and viscous damping (EDIPS) model is described by the following Newtonian equation of motion:

\[
mL^2 \ddot{\theta} = -k\dot{\theta} - c\dot{\theta} + mg\sin\theta + T(\theta, \dot{\theta})
\]

where \( \theta \) is the angular displacement (the angle is formed by an ankle joint to COM and the vertical reference vector), \( m \) is body mass minus stance foot mass, \( L \) is pendulum equivalent length,\(^2\) \( k \) is the torsional stiffness coefficient, \( c \) is the viscous damping coefficient, \( g \approx 9.8 \text{ m/s}^2 \) is gravitational acceleration, \( mL^2 \dot{\theta} \) is the inertia torque, \( k\dot{\theta} \) is torque due to the elastic property of the system, \( c\dot{\theta} \) is the damping effect within the system, \( mg\sin\theta \) when linearized around an equilibrium angle of upward vertical \( 0^\circ \) is the torque that the gravitational force field exerts on the pendulum, and \( T(\theta, \dot{\theta}) \) is the forcing function in the form of torque. The stiffness coefficient of this autonomous system can be estimated from the natural frequency \( (\omega_0) \) of the system:

\[
k = mgL + mL^2(\omega_0)^2
\]

The natural frequency was estimated by the inverse of the stride duration \( (\omega_0 = 2\pi/\tau \text{ as stride duration}) \). The damping coefficient \( (c) \) was computed by assuming critical damping and estimated as:

\[
c = 2mL^2 \sqrt{\frac{k-mgL}{mL^2}}
\]

Finally, forcing impulse \( (I) \) is estimated as:

\[
I = \frac{\sum_{\theta \in \theta_{\text{contra}}} T(\theta, \dot{\theta}) \Delta t}{L} = \frac{\sum_{\theta \in \theta_{\text{contra}}} mL^2 \dot{\theta} \Delta t + \sum_{\theta \in \theta_{\text{contra}}} c\dot{\theta} \Delta t + \sum_{\theta \in \theta_{\text{contra}}} (k-mgL) \theta \Delta t}{L}
\]

where \( \theta_{\text{contra}} \) represents the frame of initial contact of the ipsilateral leg and \( T_{\text{contra}} \) represents the toe-off of the contralateral leg. \( \Delta t \) represents the period between samples.