Use of cluster analysis for gait pattern classification of patients in the early and late recovery phases following stroke

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Abstract

The mixture of gait deviations seen in patients following a stroke is remarkably variable. An objective system for classification of gait patterns for this population could be used to guide treatment planning. Quantitated gait analysis was conducted for 47 individuals at admission to in-patient rehabilitation and again at 6 months post-stroke for 42 subjects. Non-hierarchical cluster analysis was used to classify the gait patterns of patients based on the temporal–spatial and kinematic parameters of walking. Four clusters of patients were identified at both assessment intervals. At the admission test walking velocity, peak knee extension in mid-stance and peak dorsiflexion in swing were the three factors that best characterized the groups. At 6 months the explanatory variables were velocity, knee extension in terminal stance, and knee flexion in pre-swing. Differences in muscle strength and muscle activation patterns during walking were identified between groups.

Keywords: Gait; Stroke

1. Introduction

The degree to which walking is impaired following a stroke can vary widely and relates to the severity of the patient’s lower extremity motor impairment [1]. Clinicians have recognized that even among those patients with markedly decreased ambulatory function, the combination of gait deviations can differ markedly. Identification of patterns of gait dysfunction for patients following stroke could be used to construct a classification system to guide intervention. Relation of the gait patterns to the underlying impairments would allow clinicians to target rehabilitation strategies to the individual’s needs.

Selection of the key parameters on which to base a classification system of walking after stroke has been subjective and generally based on observation of patient’s function or by visual inspection of gait parameters [2–5]. This practice leads to a classification system that is dependent on a particular individual’s opinion of which gait parameters are most critical to function. Classification using a single category of parameters has resulted in functionally heterogeneous groups. Stratification by stride characteristics alone produced groups of patients with similar velocities, but with widely variable kinematic and electromyographic (EMG) patterns [4]. Similarly, groups based on EMG patterns resulted in large variability in kinematic patterns and stride characteristics within groups [2,3].

The clinical phenomenon of partial recovery of function within the first 6 months following a stroke is widely recognized. It is not known whether patients make improvements towards normal while retaining the same basic walking pattern or whether the gait pattern changes completely during recovery. Only Shiavi et al. examined the change in patterns of walking in individuals following a stroke over the course of recovery [3].
They classified patients based on patterns of EMG activation. Nine of ten patients with an abnormal EMG activation pattern at the early test (4 weeks) demonstrated a different pattern in the chronic phase (57 weeks); two demonstrated a normal pattern and the remaining seven displayed a different abnormal pattern at the second test. They did not relate the muscle activation patterns to kinematics and the EMG patterns did not correlate with velocity.

To overcome the limitations of classification by a single category of parameters, de Quervain et al. grouped patients using a combination of velocity as an index of functional ability with lower extremity joint motion to represent the compensatory response in walking to the neuromotor deficits [5]. Based solely on visual inspection of the kinematic data, de Quervain et al. identified four patterns of gait kinematics in a group of patients in the early recovery period following a stroke. Three of the patterns were displayed by patients with a very slow velocity, (less than 20% of free speed walking for able-bodied individuals). The authors were unable to find any differences in EMG activity or clinical motor control measures that determined which pattern of walking was adopted.

Cluster analysis is a statistical technique that provides an objective, quantitative classification system by separating individuals into homogenous groups based on selected input parameters. This statistical analysis, used primarily in the social sciences, has been adopted for classification of patients into groups to guide medical treatment choices [6]. It has been used to classify patients with complaints of shoulder pain [7], to identify patterns of gait deviations in children with cerebral palsy [8], to distinguish the walking parameters of young from elderly subjects [9], to differentiate normal from abnormal EMG patterns during gait [10], and to determine if more than one pattern of walking existed for individuals without pathology [11].

The purpose of this study was 2-fold: first to form a classification system of gait patterns using a statistical analysis of the temporal–spatial and kinematic data for patients in both the early and late recovery phases after a cerebrovascular accident (CVA) and second to determine whether any measures of muscle strength, spasticity, or EMG activity during walking, were related to the patients’ walking pattern. The temporal–spatial and kinematic variables were selected as input variables for the classification system to facilitate its clinical applicability once the critical variables have been identified. A patient’s kinematics can be observed visually and basic stride characteristics recorded with minimal equipment. The EMG variables provide insight into possible causes of the patient’s gait, but cannot be observed outside of a gait laboratory.

2. Methodology

2.1. Subjects

The subjects who volunteered for this study included 52 individuals (29 men, 23 women) who were admitted for inpatient rehabilitation an average of 9.4 days following a first stroke (22 left CVA, 30 right CVA). Average age was 57.4 ± 8.7 years (range 40–75 years). Each subject signed an informed consent and was provided with the Bill of Rights of Human Subjects. All testing was conducted at the Pathokinesiology Laboratory at Rancho Los Amigos National Rehabilitation Center, Downey CA. The initial gait evaluation was conducted at an average of 14.2 days after admission. If patients were able to walk without an orthosis for at least 6 m with no more than moderate assistance of one person, the initial quantitated gait analysis was conducted within 5 days of admission. If patients were not able to ambulate at admission, the gait test was conducted within 5 days of attaining the ability to walk. Five of the 52 subjects were unable to walk without an ankle foot orthosis prior to discharge and were only included in the 6-month analysis. Walking function was reassessed at 6 months post-stroke in 42 subjects. Ten subjects were lost to follow-up. Thirty-seven individuals were tested at both admission and 6 months post-stroke.

2.2. Procedures

Gait analysis was conducted on a 10-m walkway with the middle 6 m delineated for data collection by photoelectric beams. Simultaneous recordings were made of foot–floor contacts, lower extremity kinematics, and EMG activity as subjects traversed the walkway at a self-selected velocity with customary assistive devices and shoes. Foot–floor contact patterns were recorded with a Stride Analyzer System (B & L Engineering, Tustin, CA) using compression-closing footswitch insoles taped to the bottom of the subject’s shoes. Three-dimensional kinematics of the subject’s hemiplegic lower extremity were documented with the VICON motion analysis system (Oxford Metrics, Oxford, UK). Six infrared, 50 Hz cameras recorded the location of fourteen retro-reflective markers taped onto the skin overlying the bony landmarks of the pelvis, thigh, shank and foot. Motion data were acquired on a DEC PDP 11/83 computer (Digital Equipment Corporation, Maynard, MA).

Intramuscular EMG was recorded with indwelling, fine-wire electrodes inserted into the belly of eight lower extremity muscles including gluteus maximus, semimembranosus, biceps femoris long head, adductor longus, rectus femoris, vastus intermedius, soleus, and anterior tibialis using the technique of Basmajian and Stecko [12]. Electrode placement was confirmed by palpating...
tion in the tendon or muscle belly during mild electrical stimulation through the inserted wires. EMG signals were transmitted by FM–FM telemetry (Model 2600, Biosentry Telemetry Inc., Torrance, CA), band-pass filtered through an analog filter (150–1000 Hz) and sampled and digitized at a 2500 Hz rate. The overall signal gain was 1000. Prior to the walking trials, EMG was recorded for each muscle at rest to determine the baseline threshold for myoelectric activity, during a 5-s isometric maximal voluntary contraction and during a 5-s quick stretch test for spasticity.

Maximal isometric torques were recorded with a LIDO Active dynamometer (Loredan Biomedical Inc., Davis CA) for the ankle plantar flexors, dorsiflexors and knee extensors. A tensiometer recorded the maximal isometric force produced by the hip extensors. The lever arm utilized during the effort was measured manually and used to calculate hip extension torque values. Torque values were expressed as a percentage of gender—matched normal using a laboratory database of maximal isometric torques in able-bodied individuals 40–65 years of age.

2.3. Data management and analysis

Velocity, cadence and stride length were calculated from the footswitch data and expressed as a percentage of normal (%N) for the subject’s gender and age. The timing of the gait cycle was identified from footswitch data. Each stride was time normalized with initial contact defined as 0% of the gait cycle and the end of swing as 100%. EMG data were normalized to a stance duration of 65% of the gait cycle (the mean duration for all groups) to allow for comparison across subjects.

VICON motion data were processed with ADTECH motion analysis Software (Adtech, Adelphi, MD) to produce 3-D trajectories for each marker. The position and orientation of each lower extremity segment were obtained and lower extremity joint angles for each percent of the gait cycle were determined with computer algorithms using Euler embedded coordinates. An ensemble average of all strides (typically four to six strides) was determined for each subject.

EMG signals were full-wave rectified and integrated over intervals of 0.01-s. A moving window was used to identify the greatest 1-s interval of the EMG signal recorded during the 5-s maximal muscle test. The average 0.01-s interval was calculated from the maximal 1-s period. If that value was at least 25 digitized units (61 mV), it served as the normalization value for the EMG recorded during walking. If the average 0.01-s interval of the manual muscle test was less than 25 digitized units, the normalization value for the walking trials was set at 25. Use of this minimum normalization value, which was approximately 20% of a full interference pattern, prevented inflation of EMG signals during walking in muscles where the subject lacked sufficient volitional control to produce a significant signal during manual muscle testing. EMG intensity was expressed as a percentage of maximal voluntary contraction (%MAX).

Phasing of the EMG activity during walking was determined with the EMG analyzer Software (B & L Engineering, Tustin, CA) [13]. The EMG analyzer identified the onset and cessation times (as a percentage of the gait cycle) for each package of muscle activity that had an intensity of at least 5% MAX and a duration of at least 5% of the gait cycle. With the minimum normalization value of 25 digitized units, the 5% of maximum threshold for significant EMG activity would correspond to 1.25 digitized units or 3 mV. Any signal less than this was not considered functionally significant. Packets of EMG separated by quiescent intervals of less than 5% of the gait cycle were combined. Mean onset and cessation times were calculated and a time-adjusted mean profile obtained for each muscle [13]. For statistical comparison, muscle onsets or cessations that occurred in late swing in some subjects and after initial contact in others were recorded as 100 + the percentage of the gait cycle for those values that occurred after initial contact. For instance, the soleus onset might occur at 95% of the gait cycle (terminal swing) in one subject and at 5% of the gait cycle (loading response) in another. The second value would be recorded as 105% indicating a 10% gait cycle difference rather than 90%.

The EMG analyzer program also calculated the mean intensity of EMG activity for each muscle during its period of activity in the gait cycle.

The duration of EMG activity that exceeded 5% MAX following the quick stretch test was recorded in seconds for each muscle.

2.4. Statistics

Non-hierarchical cluster analysis was used to classify the gait patterns of patients based on the temporal–spatial characteristics and the peaks of the sagittal plane kinematic parameters for each phase of the gait cycle. The optimal number of groups was determined by evaluating the resultant clinical characteristics of a practical range of numbers of groups and by maximizing the R ratio (reduction in within group differences) [10]. If the number of groups selected were too small, variability in kinematic and stride parameters within groups would be high. If too many groups were selected, differences between groups would be less or the number of subjects in one or more groups would be small. Once the clusters were formed, step-wise discriminant analysis was used to determine the variables that best determined group placement and robustness of group placement was analyzed using a cross-validation Jack–knife pro-
Ankle and thigh positions in mid and terminal stance also were significantly different between groups, but were correlated with the position of the knee in stance (Figs. 1–3). Once knee extension in mid stance entered the discriminant analysis as a predictor variable, ankle and thigh position in stance no longer added any significant information to explain group placement. Ankle dorsiflexion in terminal stance was inadequate in Group 4 (Extended) at −2°, but mildly excessive in the other groups (14°–15°). Peak thigh extension in terminal stance was markedly decreased at 12° flexion in Group 3 (Flexed), while the other groups showed reduced arcs of hyperextension compared with the 20° seen in able-bodied subjects. Group 1 (Fast) had the least reduction at 12° hyperextension while Groups 2 and 4 (Moderate and Extended) had only 2 and 4° hyperextension, respectively. Hip position in stance also was significantly different between groups but, did not discriminate as well as thigh position and was not related to the knee posture owing to variability in pelvic tilt.

Muscle strength also varied among the four groups. Isometric torques for the ankle plantar flexors, knee extensors and hip extensors were significantly reduced in the three slower groups (17–36%N) compared to the fastest group (53–70%N; Table 2). Hip extension torque was lowest in the Flexed group but, because of high variability, the difference was only statistically significant when compared with the Fast group. One subject in the Flexed group had relatively strong hip extensors (59%N) while the remainder had very low hip extension torques. Ankle dorsiflexors were statistically weaker only for the Extended group (13%N) when compared with the Fast group (52%N).

The duration of EMG in response to quick stretch was not different between subject groups for any of the muscles tested. The median duration of the quick stretch response ranged from 0.06-s for anterior tibialis to 4.24-s for adductor longus. Only adductor longus and semimembranosus had a median quick stretch response of greater than 1-s.

### Table 1
Velocity and lower extremity kinematics at admission; mean and (standard deviation)

<table>
<thead>
<tr>
<th></th>
<th>Group 1 FAST n = 11</th>
<th>Group 2 MOD n = 14</th>
<th>Group 3 FLEX n = 7</th>
<th>Group 4 EXTEND n = 15</th>
</tr>
</thead>
<tbody>
<tr>
<td>Velocity (%N)a</td>
<td>43.8 (7.9)</td>
<td>20.7 (7.4)</td>
<td>9.7 (3.9)</td>
<td>11.1 (5.3)</td>
</tr>
<tr>
<td>Ankle Dorsiflexion Terminal Stance (°)</td>
<td>13.7 (5.4)</td>
<td>14.6 (5.1)</td>
<td>13.9 (3.8)</td>
<td>−2.4 (6.2)</td>
</tr>
<tr>
<td>Ankle Dorsiflexion Mid Swing (°)</td>
<td>0.6 (5.0)</td>
<td>2.8 (5.2)</td>
<td>−7.6 (6.5)</td>
<td>−11.1 (6.2)</td>
</tr>
<tr>
<td>Knee Flexion Mid Stance (°)</td>
<td>7.0 (7.0)</td>
<td>14.3 (5.5)</td>
<td>23.0 (2.9)</td>
<td>−5.5 (7.6)</td>
</tr>
<tr>
<td>Knee Flexion Terminal Stance (°)</td>
<td>6.0 (7.1)</td>
<td>14.4 (5.2)</td>
<td>25.9 (4.7)</td>
<td>−5.1 (7.8)</td>
</tr>
<tr>
<td>Knee Flexion Pre-Swing (°)</td>
<td>41.1 (5.6)</td>
<td>35.4 (6.0)</td>
<td>40.3 (8.5)</td>
<td>36.8 (11.5)</td>
</tr>
<tr>
<td>Knee Flexion Initial Swing (°)</td>
<td>46.9 (6.3)</td>
<td>38.2 (7.4)</td>
<td>40.7 (5.7)</td>
<td>36.4 (12.8)</td>
</tr>
<tr>
<td>Thigh Extension Terminal Stance (°)</td>
<td>−12.2 (4.5)</td>
<td>−2.4 (2.8)</td>
<td>12.2 (6.4)</td>
<td>−4.0 (5.3)</td>
</tr>
</tbody>
</table>

a Variable identified by discriminant analysis as determinant of cluster placement.
During walking the mean intensity of soleus EMG was greater in the Fast group (37% MAX) than in the Flexed and Extended groups (14 and 15% MAX) (Table 3). The onset of soleus activity was delayed in the Extended group at 11% of the gait cycle compared with the Fast group at 99% of the gait cycle. Mean intensity

Fig. 1. Sagittal plane ankle joint motion during walking for the four clusters of patients at the admission test. Dorsiflexion is positive.

Fig. 2. Sagittal plane knee joint motion during walking for the four clusters of patients at the admission test. Flexion is positive.
of anterior tibialis EMG was significantly greater in the Moderate group (27% MAX) than in the Flexed and Extended groups (8% MAX). The intensity of anterior tibialis activity in the Fast group at 24% MAX was greater than in the two slower groups with a trend for statistical significance ($P < 0.1$).

At the knee, vastus intermedius had a lower mean intensity in the Extended group (13% MAX) than in the Moderate and the Fast groups (28 and 26% MAX), although the comparison only reached statistical significance with the Moderate group. Onset of vastus intermedius was significantly later in the Extended group (5% gait cycle) than in the Fast (87% gait cycle) and the Moderate groups (93% gait cycle). The intensity of semimembranosus was greater in the Fast group (21% MAX) than in the Flexed and Extended groups (10% MAX) with a trend for the Moderate group at (13% MAX). The onset of semimembranosus was earlier in the Fast group (76% gait cycle) than in the Flexed (95% gait cycle) and Extended (90% gait cycle) groups. Biceps femoris activation was similar to that of semimembranosus with greater intensity in the Fast group (19% MAX) than in the Flexed (11% MAX) and Extended groups (7% MAX). The onset of biceps femoris also was earlier in the Fast group (80% gait cycle) than in the Moderate (94% gait cycle), Flexed (0% gait cycle) and Extended (98% gait cycle) groups.

### 3.2. Six-month test

Four clusters of patients also were identified in the 6-month test data but, the determining gait parameters differed from those identified for the initial test. The three gait parameters that best determined group placement were velocity, knee flexion in terminal stance and knee flexion in preswing. Group placement using only these three parameters was 98% accurate. The Jackknife procedure estimated that new patients would be classified correctly 93% of the time.

#### Table 2

Maximal isometric torque at admission; mean and (S.D.)

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Group 1 FAST $n = 11$</th>
<th>Group 2 MOD $n = 14$</th>
<th>Group 3 FLEX $n = 7$</th>
<th>Group 4 EXTEND $n = 15$</th>
<th>$P$ value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ankle Plantar Flexion (%N)</td>
<td>53.2 (15.1)</td>
<td>22.6 (17.2)*</td>
<td>16.8 (16.1)*</td>
<td>17.8 (11.2)*</td>
<td>0.0000</td>
</tr>
<tr>
<td>Ankle Dorsiflexion (%N)</td>
<td>52.4 (27.0)</td>
<td>38.5 (28.0)</td>
<td>23.1 (19.1)</td>
<td>13.0 (14.5)*</td>
<td>0.0096</td>
</tr>
<tr>
<td>Knee Extension (%N)</td>
<td>56.6 (17.5)</td>
<td>27.8 (17.1)*</td>
<td>27.4 (34.9)*</td>
<td>22.8 (20.4)*</td>
<td>0.002</td>
</tr>
<tr>
<td>Hip Extension (%N)</td>
<td>69.5 (23.3)</td>
<td>33.9 (22.0)*</td>
<td>23.5 (17.6)*</td>
<td>36.3 (22.5)*</td>
<td>0.0002</td>
</tr>
</tbody>
</table>

*Significantly different than Fast Group.

Fig. 3. Sagittal plane thigh joint motion during walking for the four clusters of patients at the admission test. Flexion is positive.
Table 3
EMG timing and intensity during walking at admission; mean and (S.D.)

<table>
<thead>
<tr>
<th>Muscles</th>
<th>Group 1 FAST n = 11</th>
<th>Group 2 MOD n = 14</th>
<th>Group 3 FLEX n = 7</th>
<th>Group 4 EXTEND n = 15</th>
<th>P value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Soleus intensity (% MAX)</td>
<td>36.5 (22.6)</td>
<td>21.3 (12.6)</td>
<td>14.3 (4.6)</td>
<td>15.0 (15.0)</td>
<td>0.012</td>
</tr>
<tr>
<td>Soleus onset (% Gait Cycle)</td>
<td>99.1 (7.9)</td>
<td>3.7 (8.1)</td>
<td>0.3 (5.3)</td>
<td>11.1 (15.3)</td>
<td>0.044</td>
</tr>
<tr>
<td>Anterior tibialis intensity (% MAX)</td>
<td>24.6 (8.4)</td>
<td>27.1 (8.6)</td>
<td>7.8 (3.1)</td>
<td>8.4 (6.8)</td>
<td>0.011</td>
</tr>
<tr>
<td>Vastus intermedius intensity (% MAX)</td>
<td>25.7 (7.3)</td>
<td>28.0 (16.8)</td>
<td>21.5 (10.4)</td>
<td>12.5 (8.6)</td>
<td>0.025</td>
</tr>
<tr>
<td>Semimembranosus onset (% Gait Cycle)</td>
<td>75.5 (6.8)</td>
<td>81.9 (7.4)</td>
<td>95.2 (5.0)</td>
<td>90.3 (13.6)</td>
<td>0.020</td>
</tr>
<tr>
<td>Semimembranosus intensity (% MAX)</td>
<td>20.7 (6.7)</td>
<td>12.7 (7.6)</td>
<td>9.6 (9.3)</td>
<td>10.0 (8.6)</td>
<td>0.035</td>
</tr>
<tr>
<td>Vastus intermedius onset (% Gait Cycle)</td>
<td>87.0 (8.5)</td>
<td>93.0 (7.2)</td>
<td>98.6 (8.4)</td>
<td>4.6 (13.3)</td>
<td>0.0013</td>
</tr>
</tbody>
</table>

% MAX, percentage of EMG during maximum isometric contraction.
a Significantly different than Fast Group.
b Significantly different than Moderate Group.

The subjects were relatively evenly distributed between the four groups with 12 in the first group, and ten in the other three. Group 1 (Fast) had the fastest velocity at 67% of normal with 4° of knee flexion in terminal stance and normal knee flexion in preswing at 48° (Table 4, Fig. 5). The second group (Moderate/Slight Extension) had an intermediate velocity of 41%N with 2° of hyperextension in terminal stance and 37° knee flexion in preswing. Group 3 (Flexed) had a slow velocity (27%N) with excessive knee flexion in terminal stance at 18° and 39° of knee flexion in preswing. Group 4 (Extended) also had a very slow velocity (20%N) with 7° of knee hyperextension in terminal stance and markedly limited knee flexion in preswing at 18°.

Similar to the results at the admission test, the ankle and thigh positions in mid and terminal stance were significantly different between groups, but were correlated with the position of the knee in stance. Once knee extension in terminal stance entered the discriminant analysis as a predictor variable, ankle and thigh position in stance no longer added any significant information to explain group placement. Ankle dorsiflexors in terminal stance was inadequate in Group 4 (Extended) at 3°, normal in Group 2 (Moderate) at 10°, and excessive in Groups 1 and 3 (Fast and Flexed) at 19 and 16°, respectively (Fig. 4). Peak thigh extension in terminal stance was moderately decreased at 0° in Group 3 (Flexed) and mildly reduced in Groups 1, 2 and 4 (Fast, Moderate and Extended) at 16, 14 and 11° of hyperextension, respectively (Fig. 6).

Isometric torques for the ankle plantar flexors and ankle dorsiflexors were significantly weaker in both of the slow groups (Flexed and Extended; 23–35%N) than in the fastest group (50–66%N; Table 5). Hip extensors were weaker only for the Flexed group at 27% of normal compared with the Fast group at 69%N. The knee extensors, however, were weaker only in the Extended group when compared with the Fast group (25 and 50%N, respectively).

The duration of EMG in response to quick stretch again, was not different between subject groups for any of the muscles tested. Most of the muscles demonstrated little change in the duration of the response from that seen at admission with a low for anterior tibialis at 0.04-s and a high of 4.2-s for adductor longus.

During walking, mean intensity of soleus EMG was greater in the Fast group (43% MAX) than in the other three groups (22–24% MAX; Table 6). The onset of vastus intermedius was significantly later in the Flexed group (96% gait cycle) than in the Fast and Moderate groups (83% gait cycle). Cessation of vastus intermedius was significantly later in the Flexed group (52% gait...
cycle) than in the Extended (29% gait cycle) and the Moderate groups (30% gait cycle) while cessation of biceps femoris was later in the Extended group (44% gait cycle) than in the Fast group (25% gait cycle). Adductor longus intensity was greater in the Fast group (28% MAX) than in the Extended group (13% MAX). Cessation of gluteus maximus was delayed in the Flexed (39% gait cycle) and Extended groups (41% gait cycle).

Fig. 5. Sagittal plane knee joint motion during walking for the four clusters of patients at the 6-month test. Flexion is positive.

Fig. 4. Sagittal plane ankle joint motion during walking for the four clusters of patients at the 6-month test. Dorsiflexion is positive.
compared with the Fast group (21% gait cycle) but only reached statistical significance for the Extended group.

Thirty-seven subjects were tested during walking without an ankle-foot orthosis both at admission and at 6 months post-stroke. Group assignment from admission to 6 months was most stable for the Fast group where seven out eight individuals who were in the Fast group at admission remained so at 6 months (Table 7). One subject switched from the Fast group at admission to the Moderate (slightly extended) group at 6 months. The Extended group also was somewhat stable with the eight out of 12 either remaining in the Extended group at 6 months (n = 4) or switching to the Moderate group (n = 4) which had a similar, but less extreme, pattern of extension. One individual who was in the Extended group at admission moved into the Fast group at 6 months. Three subjects changed patterns from Extended at admission to Flexed at 6 months.

The Flexed and Moderate (slightly flexed) groups demonstrated more change in walking patterns from the early to the 6-month test. Of the six subjects who were in the Flexed group at admission, all but one had a different pattern at 6 months with four changing to the Extended group and one to the Fast group. There was no equivalent pattern of the Moderate (slightly flexed) group at the 6-month test. The Moderate group with an intermediate velocity at 6 months had a slightly extended motion pattern at the ankle and knee. The motion pattern of the Flexed group at 6 months was most similar to that of the Moderate group at admission. Of the 11 patients with the Moderate pattern at admission, five were classified in the Flexed group, three

### Table 5

<table>
<thead>
<tr>
<th></th>
<th>Group 1 FAST n = 12</th>
<th>Group 2 MOD n = 10</th>
<th>Group 3 FLEX n = 10</th>
<th>Group 4 EXTEND n = 10</th>
<th>P value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ankle plantar flexion (%N)</td>
<td>49.8 (13.7)</td>
<td>34.8 (14.0)</td>
<td>22.7 (12.9)*</td>
<td>22.6 (9.8)*</td>
<td>0.0000</td>
</tr>
<tr>
<td>Ankle dorsiflexion (%N)</td>
<td>66.4 (16.7)</td>
<td>48.9 (14.8)</td>
<td>34.9 (22.2)*</td>
<td>26.9 (14.6)*</td>
<td>0.0001</td>
</tr>
<tr>
<td>Knee extension (%N)</td>
<td>49.8 (19.9)</td>
<td>33.1 (13.9)</td>
<td>30.5 (13.7)</td>
<td>25.3 (15.7)*</td>
<td>0.0099</td>
</tr>
<tr>
<td>Hip extension (%N)</td>
<td>69.1 (15.0)</td>
<td>46.5 (28.2)</td>
<td>27.3 (20.7)*</td>
<td>42.5 (25.3)</td>
<td>0.0017</td>
</tr>
</tbody>
</table>

* Significantly different than Fast Group.
in the Fast group, two in the Moderate (slightly extended) group, and one was placed in the Extended group at the 6-month test.

4. Discussion

The walking pattern in the early period following a stroke was characterized by decreased muscle activation. Those patients with greater walking impairment demonstrated decreased intensity and duration of muscle activity in several critical muscles including soleus, anterior tibialis, quadriceps and the hamstrings. This decreased muscle activation resulted in limitations to both stance and swing phase movement patterns. In the stance phase, insufficient activity of the ankle plantar flexors resulted in poor control of the tibia. The pattern of walking compensation adopted was reflected in the knee position in mid stance. In the two slower groups without restraint of the tibia from the calf muscles, the knee either collapsed into flexion with the demand shifted to the quadriceps or thrust into hyperextension with reliance on passive knee stability and the hip extensor muscles. As volitional lower extremity strength increased in the two faster groups, improved tibial control resulted in less residual knee flexion with normal extension in the Fast group.

The swing phase movement pattern was characterized by the amount of dorsiflexion achieved in mid swing. The decreased activation of anterior tibialis in the two slower groups produced inadequate dorsiflexion in mid swing. While volitional ankle dorsiflexion strength was less than normal in the two faster groups, the muscle response was adequate to achieve at least a neutral ankle in swing.

The finding of decreased muscle activation in the early post-stroke period agrees with the results of Shia et al. who identified reduced activation in two or more muscles as the most common EMG pattern in the acute phase [3]. They did not specify, however, which muscles were most often affected.

The kinematic patterns identified by cluster analysis in the early post-stroke period in this study were similar, but not identical, to those documented by De Quervain et al. [5]. They described four patterns in 18 subjects with three of the four patterns occurring in subjects who had a very slow velocity (6–17%N). The Extension thrust pattern described in their study was very similar to the individuals in our Group 4 (Extended) with knee hyperextension in stance and inadequate knee flexion in swing. Their Buckling-knee pattern was characteristic of

<table>
<thead>
<tr>
<th>Group at admission</th>
<th>Group at 6 months</th>
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<tbody>
<tr>
<td>FAST n = 11</td>
<td>MOD slightly extended n = 10</td>
</tr>
<tr>
<td>FAST n = 11</td>
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<tr>
<td>MODERATE Slightly flexed n = 14</td>
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<tr>
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<td>EXTENDED n = 15</td>
<td>1</td>
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<td>AFO Only n = 5</td>
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</tbody>
</table>

%MAX, percentage of EMG during maximum isometric contraction.

* Significantly different than Fast Group.

** Significantly different than Flexed Group.
those patients in our Group 3 (Flexed). We found no group analogous to the Stiff-Knee group described by De Quervain in their slow subjects. Their subjects with a ‘Stiff Knee’ pattern had a combination of 30° of knee flexion sustained throughout the gait cycle with decreased stance phase ankle dorsiflexion. This pattern of sustained knee flexion was not seen in any of our subjects.

The gait pattern of the subjects with an intermediate velocity described by De Quervain was one of mildly increased knee flexion and ankle dorsiflexion in stance with a velocity of 38%N and was similar to that of the individuals in our Group 1 (Fast) [5]. The De Quervain study did not report a pattern analogous to that of our Group 2 (Moderate). These subjects likely were included in their Intermediate group as this group was slightly slower and had more stance phase knee flexion than our Fast group.

At the 6-month test, velocity had improved in all groups, but the stance phase pattern was still dominated by the amount of knee extension attained in single limb stance. Many subjects continued to display inadequate tibial stability from poor calf function even after the natural recovery period and rehabilitation program focused on increasing lower extremity strength and motor control. Patients with only mild calf and quadriceps weakness demonstrated the fastest velocity with excessive dorsiflexion in stance to maximize forward progression. While their borderline plantar flexion strength permitted excessive dorsiflexion without collapse at the knee, it was not strong enough to produce a heel rise in single limb stance. The excessive dorsiflexion helped to compensate for the lost stride length from the lack of heel rise.

For those patients with severe ankle plantar flexor weakness, velocity was markedly reduced and the pattern of gait adopted depended on the relative strength of the knee and hip extensors. Patients in the Flexed group with greater weakness in the hip extensors than in the knee extensors displayed a pattern of excessive flexion at all joints in stance with prolonged activity of the quadriceps to support the flexed knee posture. Rehabilitation efforts for this group should include an ankle–foot orthosis with a dorsiflexion stop and strengthening exercises particularly for the hip extensors. Those patients with stronger hip extensors than knee extensors (Extended group) were excessively extended at the ankle and knee in stance with prolonged activity of biceps femoris in response to the knee hyperextension.

Ankle dorsiflexion in swing improved in the two slower groups by the 6 months test, although the Extended group still did not average a neutral ankle angle. This improvement in swing phase dorsiflexion reflected increased intensity of anterior tibialis in the two slower groups from the muscle activity levels seen at admission. The dominant characteristic of the swing phase at 6 months shifted from the ankle dorsiflexion position to the amount knee flexion attained. Preswing knee flexion angle provided better statistical discrimination between groups than the peak value in initial swing. Swing phase knee flexion showed slight improvements in all groups except in the Extended group which had a 12° decline from the admission test. Their average peak knee flexion angle of 24° represents a severe impediment to swing limb advancement.

Inadequate knee flexion in swing has been associated with several causes including prolonged activity of rectus femoris, inadequate hip flexion and decreased ankle plantar flexion in preswing [14—16]. Rectus femoris activity was prolonged in all groups of subjects with no significant differences in timing among groups. Additionally, no differences between groups were seen in the duration of rectus femoris activity elicited by a quick stretch in supine. The Extended group of patients did show decreased activation of adductor longus, however, (13% MAX compared with the Fast group at 28% MAX which contributed to a weak flexion pattern and resulted in markedly inadequate knee flexion in swing. Those patients with more adequate activation of adductor longus were able to attain a greater knee flexion angle despite prolonged rectus femoris activity by increasing the hip flexion effort, which secondarily increases knee flexion through inertia. Patients in the Flexed group also had low Adductor Longus intensity at 17% MAX. Their arc of knee flexion in preswing and initial swing was significantly less than normal, but their starting position was more flexed owing to their collapsed position in single limb stance. Subjects in this group may develop decreased peak knee flexion in swing if a more vertical posture in stance is achieved through strengthening or orthotic control.

Rehabilitation for patients with the extended walking pattern should focus on strengthening of the quadriceps and the entire flexor pattern. For patients whose swing phase knee flexion did not improve with strengthening of the hip flexors, assessment of rectus femoris by dynamic EMG could identify activation patterns amenable to surgical intervention [14]. Orthotic support would be appropriate to provide ankle dorsiflexion to neutral in swing.

Average walking velocity was the strongest determinant of group placement at both the early and 6-month gait evaluations. This partly reflects the wide range of walking ability in the study population. Velocity alone, however, was insufficient to characterize the functional groups especially among the slower groups. Onley and Richards were unable to discriminate between the flexed and extended postures when subjects were grouped by velocity alone [4]. They described no pattern that involved knee hyperextension or inadequate dorsiflexion in stance. The ranges of velocities overlapped in all...
groups in this study, but the Flexed and Extended groups had particularly similar values. The kinematic patterns adopted by the subjects in these two groups were strikingly different. Differences in muscle strength and activation during gait that explained the substitution patterns employed were identified. This finding is consistent with a previous study in which velocity alone successfully classified subjects into various levels of community ambulation ability following a stroke but was unable to differentiate distinct functional levels for individuals who were limited to household ambulation [1].

The limitations of this study relate both to its exploratory nature and inclusion of multiple variables as well as the exclusion of other gait data. The variables chosen to form the patient clusters, kinematics and temporal–spatial parameters, were limited to sagittal plane motion of the hemiparetic limb only and the easily measured stride characteristics (velocity, cadence, stride length). EMG and isometric torque measurements also were restricted to those with primarily sagittal function. Quantification of hip flexion strength would have assisted with identification of the possible causes of inadequate knee flexion in swing. The significant decrease in adductor longus activity documented in the Extended group suggests that poor hip flexion activation is an important cause of this deviation. This corresponds with normal function when the primary determinant of swing phase knee flexion originates from the hip. Knee flexion strength is less critical to swing limb advancement [17].

Characterizing muscle activity patterns during walking is complex, involving multiple variables even for a single muscle (onset, cessation, intensity, timing and magnitude of peak activity). Since this study was not designed with a priori hypotheses, the number of EMG variables compared across the patient clusters was large (n = 24). This number of comparisons increases the likelihood of type I statistical error and thus the results must be viewed with caution. Adjusting for the multiple comparisons in this type of study, however, would be overly conservative. A logical, next step would be to group a new population of subjects using the variables identified as critical to cluster formation (velocity, knee extension in stance and ankle dorsiflexion in swing for the early recovery period or knee flexion in swing for the later period) and test only selected a prior hypotheses about EMG patterns.

The cluster analysis was successful in objectively identifying gait patterns that reflected homogenous levels of function. Clinicians can use the critical parameters identified to categorize patients following CVA. The intervention program can then be targeted more specifically to the underlying impairments. The qualitatively pattern of walking changed markedly in many patients from the early evaluation to the 6 months test. Continued follow-up care after discharge from inpatient rehabilitation is critical to determine if the therapeutic interventions prescribed initially such as orthoses, assistive devices and exercise programs are still appropriate later in the recovery course.

Acknowledgements

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References