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Clinical Study

A Low-Cost Biofeedback System for Electromyogram-Triggered Functional Electrical Stimulation Therapy: An Indo-German Feasibility Study

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Functional electrical stimulation (FES) facilitates ambulatory function after paralysis by activating the muscles of the lower extremities. The FES-assisted stepping can either be triggered by a heel-swich, or by an electromyogram-(EMG-) based gait event detector. A group of six chronic (>6 months poststroke) hemiplegic stroke survivors underwent transcutaneous FES-assisted training for 1 hour on stepping task with EMG biofeedback from paretic tibialis anterior (TA) and medial gastrocnemius (GM) muscles, where the stimulation of the paretic TA or GM was triggered with surface EMG from the same muscle. During the baseline, postintervention, and 2-day-postintervention assessments, a total of 5 minutes of surface EMG was recorded from paretic GM and TA muscles during volitional treadmill walking. Two-way ANOVA showed significant effects in terms of P values for the 6 stroke subjects, 0.002, the 3 assessments, 0, and the interaction between subjects and assessments, $6.21E-19$. The study showed a significant improvement from baseline in paretic GM and TA muscles coordination during volitional treadmill walking. Moreover, it was found that the EMG-triggered FES-assisted therapy for stand-to-walk transition helped in convergence of the deviation in centroidal angular momentum from the normative value to a quasi-steady state during the double-support phase of the nonparetic. Also, the observational gait analysis showed improvement in ankle plantarflexion during late stance, knee flexion, and ground clearance of the foot during swing phase of the gait.

1. Introduction

Stroke is caused when an artery carrying blood from heart to an area in the brain bursts or a clot obstructs the blood flow thereby preventing delivery of oxygen and nutrients. Global Burden of Disease Study estimated a population-based annual stroke incidence in India to be 89/100,000 in 2005, which is projected to increase to 91/100,000 in 2015 and to 98/100,000 in 2030 [1]. Foot drop is a common symptom in stroke survivors that inhibits the sufferer from being able to raise their foot during the swing phase of gait. The ability to walk is important for independent performance of activities of daily living and therefore determines the quality of life [2]. Reduced walking for activities of daily living further affects their cardiovascular health which can make them susceptible to another stroke. Functional electrical stimulation (FES) involves electrical stimulation of nerves and muscles with continuous short pulses of electrical current at a certain pulse rate (or frequency) in a coordinated fashion to improve functional movement of limbs during walking [3]. FES has been shown in studies to enhance walking abilities in stroke survivors, increase gait speed while lowering effort, increase confidence during walking due to reduced fear of tripping, reduce spasticity in the paretic leg while increasing the range

of motion at the ankle, and has recently developed into a therapeutic intervention for poststroke gait rehabilitation [4–11].

Gait is a complex biomechanical task that requires coordination across multiple limb segments of human locomotor apparatus to maintain balance. The initiation of gait following quiet standing requires volitional transition from a state of static stability to steady state walking involving repetitive leg motor pattern and emergence of dynamic stability. Understanding multisegment coordination of the locomotor apparatus during the transient state between standing and steady state walking—stand-to-walk transition—is necessary for effective interventions following neuromuscular disorders. Moreover, stand-to-walk transition can be an important tool for diagnosing pathological gait [12] where a pattern of muscle activation is necessary for normal walking [13]. Here, a modular organization has been shown across different walking speeds and body weight support [13–15] where these modules representing coordination among multiple muscles remain relatively consistent indicating basic elements of neural control [15–17]. Moreover, these modules have been shown to have functional relevance in gait biomechanics [18, 19] where healthy modules can be merged to predict poststroke reduced locomotor performance and muscle coordination complexity [20].

In this study, we investigated the flexor-extensor coordination in the paretic ankle of stroke survivors suffering from foot drop and how that affected poststroke stand-to-walk transition. In fact, Neptune and colleagues [19] have found synergistic action of soleus and medial gastrocnemius which provided body support and forward propulsion in late stance and then synergistic action of rectus femoris, tibialis anterior, and hamstrings, which coordinated leg swing by acting to accelerate the leg into swing in early stance and decelerate the leg in late swing in preparation for foot contact. Our prior work [21] on coordinated muscle action found clusters of surface EMG patterns from the lateral gastrocnemius (GL), medial gastrocnemius (GM), peroneus longus (PL), biceps femoris (BF), rectus femoris (RF), tibialis anterior (TA), gluteus medius (GD), vastus lateralis (VL), vastus medialis (VM), and adductor longus (AD) based on their cross correlation coefficients where 4 distinctly separate groups or "synergies" ((1): GL+PL+GM; (2): BF+TA; (3): GD+RF+VL; and (4): AD) were found [22]. The "synergies" were modulated bilaterally during able-body gait cycle (GC) from heel strike (HS) to HS [22]. Therefore, these "synergies" may provide a "minimal" set of muscle coordination to be targeted during gait rehabilitation where the stroke survivor can learn to volitionally merge them into a normal pattern using EMG biofeedback of the deficits.

The objective of this study was to investigate the improvement in the stepping biomechanics as well as paretic TA-GM coordination (preliminary findings presented in a conference [22]) following EMG-triggered FES-assisted training of weight-transfer and forward propulsion with the paretic limb during stepping action, targeting "synergies" (1) and (2). Here, biomechanical studies have found that the aggregate angular momentum of the body referred to its center of mass—centroidal angular momentum—is highly regulated [23]. Moreover, the rate of change of the centroidal angular momentum has been shown to contain gait stability information [24]. This makes centroidal angular momentum an interesting biomechanical parameter to be investigated during stepping action where the stand-to-walk transition was initiated in this study with the paretic leg. It was hypothesized that stepping action is a controlled fall where the changes in angular momentum are regulated with appropriate muscle activity "synergies" generating joint moments and appropriate foot placement. In fact, it has been shown during able-bodied walking that adjacent leg-segment momenta are balanced in the mediolateral direction (left foot momentum cancels right foot momentum, etc.) [23]. In accordance, a low-cost system for EMG-triggered functional electrical stimulation therapy was developed that was used for poststroke EMG biofeedback training of stepping action as well as to capture the consequent regulation of angular momentum during the gait training sessions [25].

2. Methods

2.1. Subjects. Four able-bodied subjects (age: 24–28 years) and six ambulatory chronic (>6 months poststroke) stroke survivors (age: 52–78 years) suffering from unilateral foot drop volunteered for this study after informed consent. The Indo-German clinical study was registered with the Clinical Trials Registry, India, on 09/02/2012 (CTRI/2012/02/002412). All the stroke survivors who participated in this study suffered unilateral infarct in the territory supplied by the middle cerebral artery, excluding basal ganglia. The stroke survivors received two weeks (3 days a week) of treadmill gait training with heel switch triggered (ODFS Pace, Odstock, UK) FES system before they volunteered for this EMG-triggered FES therapy where they could comfortably walk at a gait speed of more than 0.6 m/sec for more than 6 minutes on a treadmill without any assistance.

2.2. Experimental Setup. The experimental setup for EMG-triggered FES therapy is described in Dutta, Khattar and Banerjee [22] and is shown in Figures 1(a) and 1(b). A single-channel transcutaneous FES device (ODFS Pace, Odstock, UK) delivered electrical stimulation to activate medial gastrocnemius (GM) or tibialis anterior (TA) muscles of the paretic leg. Surface EMG was collected from the muscles with 2 cm interelectrode distance following SENIAM guidelines [26]. The EMG signals were amplified and low-pass filtered (anti-aliasing, $\text{frequency}_{\text{cutoff}} = 1000$ Hz) with custom-made amplifier before being sampled at 2400 Hz by 16-bit data acquisition system (NI USB-6215, National Instruments, USA). The gain of the amplifier was set to prevent saturation at the maximum volitional contraction. The linear envelope (LE) of EMG was used as the control signal which was computed after a digital band-pass filtering (5th order Butterworth, 3 dB bandwidth = 10–500 HZ), then full-wave rectification, and then low-pass filtering (5th order Butterworth, 3 dB $\text{frequency}_{\text{cutoff}} = 3$ Hz) the sampled EMG signal. Visual biofeedback was provided to the subject on a computer monitor with a sliding window (averaged over 0.1 sec) of

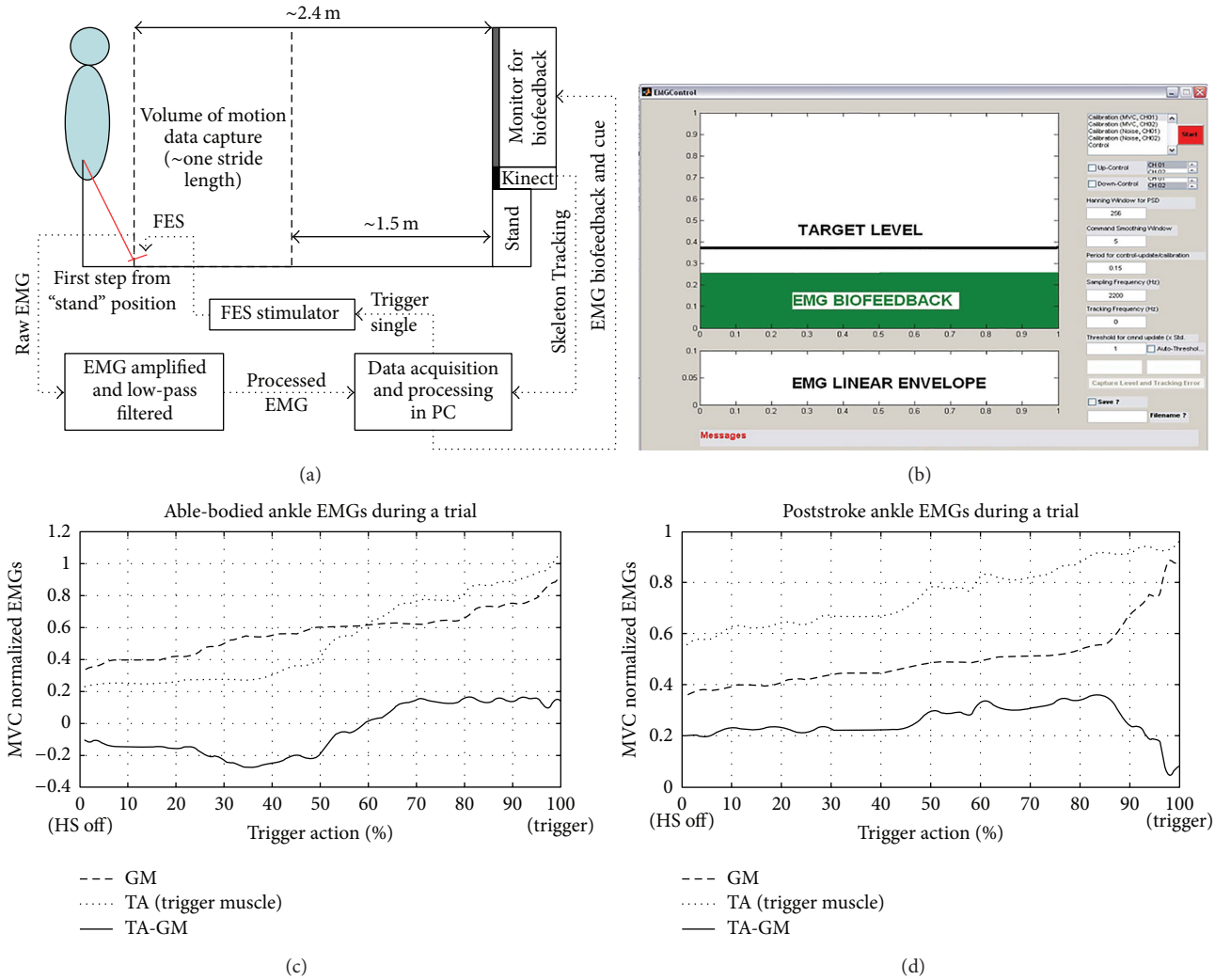


Figure 1: (a) Experimental setup for Electromyogram- (EMG)-triggered functional electrical stimulation (FES) training (PC: personal computer). The paretic leg (shown in red) is the initiator leg for stand-to-walk transition. (b) EMG biofeedback presented during deficient gait phase identified with "Skeleton Tracking" data from [21]. (c) Illustrative example of able-bodied EMG from medial gastrocnemius (GM) and tibialis anterior (TA) normalized by their maximum voluntary contraction (MVC) during stepping action following heel switch (HS) release (trigger at 90% MVC) for EMG-triggered FES training. (d) Illustrative example of poststroke EMG from GM and TA normalized by their MVC during stepping action (trigger at 90% MVC) for EMG-triggered FES training.

the control signal normalized by the maximum voluntary contraction (Figure 1(d)) while the subjects learned to trigger the stimulation with EMG from their paretic GM or TA muscle. An illustrative example of GM and TA EMGs during a stepping action is shown in Figure 1. The threshold was set at 90% maximum voluntary contraction (MVC) so that the subject could comfortably trigger "FIXED TIMING" stimulation of 300 ms duration. The stimulation was delivered once either to GM or TA muscle, during weight-transfer from the paretic limb to the unaffected limb and the subsequent stepping action. The PC-based data acquisition system (NI USB-6215, National Instruments, USA) activated a switching circuit (clamp) by a trigger pulse that disconnected the EMG electrode inputs from the amplifier and connected them to common ground electrode while delivering FES to the same muscle, by activating the FES device with a relay that shorted a rheostat (~M Ω) in series with the force sensitive resistor (heel switch).

The MS Kinect (Microsoft, USA) was used to capture the kinematics of the subject using the Microsoft software development kit (SDK) (Microsoft, USA) [28] where it relayed that information to the PC as "Skeleton Tracking" data, as shown in Figure 2(a). The capture volume was fixed roughly from 1.5 m to 2.4 m from the MS Kinect sensor that provided just enough length to capture one gait cycle (GC) from foot strike to foot strike of the paretic leg (nondominant leg for able-body), the initiator leg for stand-to-walk transition. The body kinematics parameters such as the center of mass (CoM) were estimated from the "Skeleton Tracking" data which is a spatial location where all of the mass of the system could be considered to be located in space. The CoM depends on the pose of the body and it is

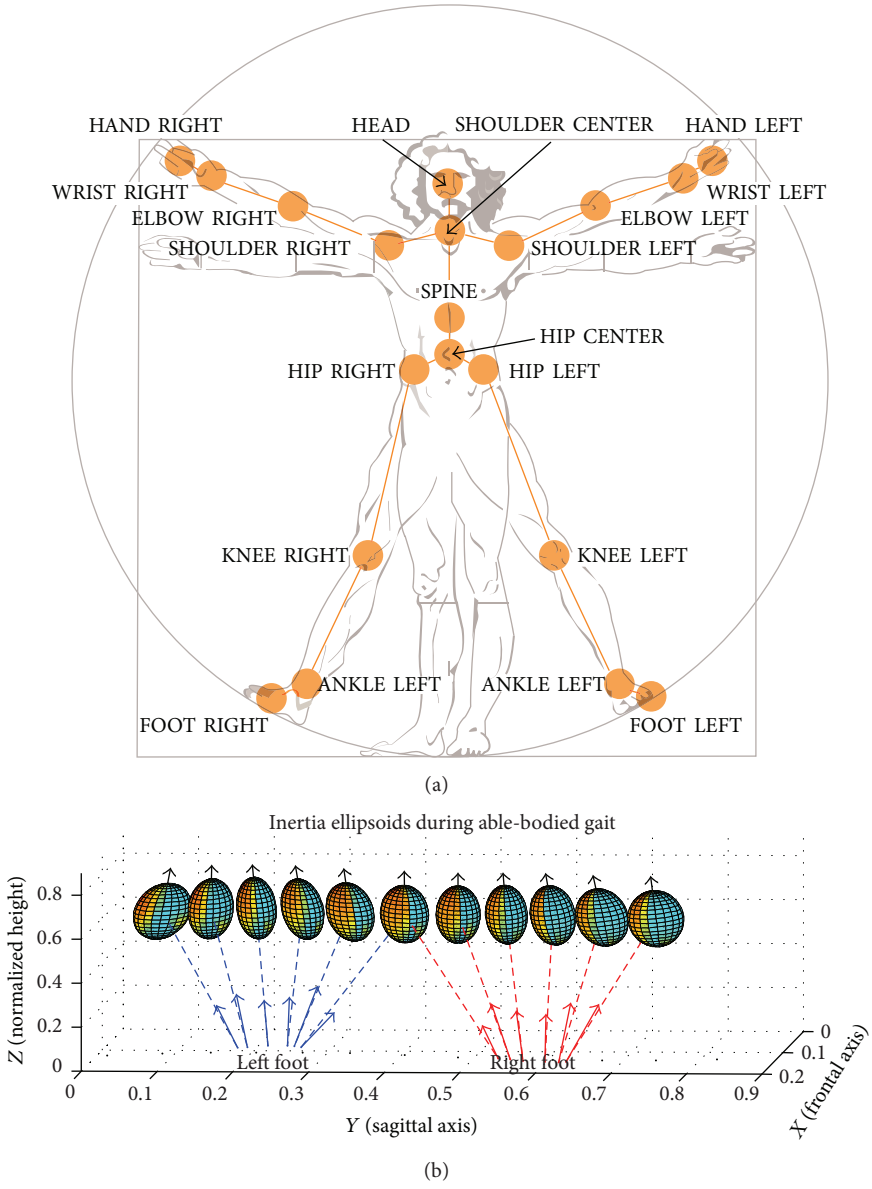


Figure 2: (a) Joint labels for the skeleton model data from Microsoft Kinect. (b) A reduced dimension biped model for capturing the posture. The dimensions of the ellipsoid changes based on the rotational inertia of the body, as discussed in Dutta and Goswami [27].

possible to have the CoM outside the body. The body can maintain static balance as long as the CoM is maintained within the support area (i.e., base of support), which is the area between the feet. For computational purposes, it is often possible to replace the entire mass of the body with a point mass which is equal to body's mass in magnitude and is located at the CoM. Human body can be considered to consist of several solid links (i.e., body segments) and joints. The anthropometric data of human body segments with their proximal joint endpoint, distal joint endpoint, fractional body mass, and their CoM were found from literature [29]. The body was therefore divided into following 7 segments: foot, shank, thigh, hand and forearm (lumped into forearm), upper arm, and head and trunk (lumped into head + trunk). The CoM of head, hand, and foot were taken as HEAD, HAND RIGHT and HAND LEFT, and FOOT RIGHT and FOOT LEFT, respectively, from the "Skeleton Tracking" data (see Figure 2). In the "Skeleton Tracking" data, the location of two points per segment for shank, thigh, forearm, and upper arm was available. Therefore, the CoM for each of these segments was on the line joining the two end-points based on anthropometric data [29]. The CoM for the trunk was on the line joining HIP_CENTER and SHOULDER_CENTER as obtained from anthropometry [29]. The CoM of the whole body was determined by taking a weighted average of the CoM of body segments, which were weighted by their fractional body mass. Thereafter, the angular momentum of all the segments about the whole

body CoM was summed together to compute the centroidal angular momentum (CAM) as given below,

$$L = \sum_{i=1}^{7} \left[\left(r_i^{\text{com}} - r^{\text{com}} \right) \times m_i \left(v_i^{\text{com}} - v^{\text{com}} \right) + I_i^{\text{com}} \omega_i \right], \quad (1)$$

where L represents CAM, r_i^{com}, v_i^{com}, and ω_i are the position, linear velocity, and angular velocity, respectively, of the CoM of the ith body segment of mass m_i, and I_i^{com} is the rotational inertia about CoM. It should be noted that since only two points per segment were monitored for most body segments any rotation (to compute ω_i) of those body segments along the line joining the two end points could not be registered. The massive trunk had 7 points in the "Skeleton Tracking" data where its ω_i was captured more accurately, especially the rotation about the vertical direction (i.e., in transverse plane). The r^{com} and v^{com} are the position and linear velocity, respectively, of the whole body CoM. The change in CAM about CoM due to the sum of moments acting on the subject is given by

$$\frac{dL}{dt} = M_{\text{GR}} + \left(r^{\text{com}} - r^{\text{cop}} \right) \times F_{\text{GR}}, \quad (2)$$

where the r^{com} and r^{cop} are the position of the whole body CoM and center of pressure (CoP), respectively, M_{GR} is the ground reaction moment, and F_{GR} is the ground reaction force (GRF). Equations (1) and (2) show that the CAM can be regulated with CoP location (i.e., foot placement) and GRF (i.e., joint moments) as well as by modulating the rotational inertia about CoM (i.e., configuration of the locomotor apparatus). During biped walking, the CAM needs to maintain a tightly regulated oscillation about zero for dynamical stability [23, 24] when the body undergoes controlled fall during the single-support phase of gait where reduced muscle function and joint moments in hemiplegia may lead to compensatory movements [27]. Further, it has been shown in able-bodied walking that the pelvis and abdomen momenta are balanced by leg, chest, and head momenta in the anterior-posterior direction, and leg momentum is balanced by upper-body momentum in the vertical direction [23]. In fact, Dutta and Goswami have shown that it is necessary to account for the centroidal moment that is generated by the GRF about the CoM to regulate CAM during human walking [27]. Here, the centroidal moment was significantly different from zero not only during pathological gait but also during the double support phase of the able-bodied gait. In our prior work [27], we had collected the force plate data to investigate the role of ground reaction forces (GRF) where the centroidal moment (CM) generated by the GRF vector about the CoM in able-bodied gait is shown in (a) and (b) (inset) of Figure 3. Here, the GRF generated an anticlockwise CM (labeled positive) at foot strike and a clockwise CM (labeled negative) before foot off. Figure 3(b) also shows that able-bodied subjects during overground walking placed the foot such that CAM could be adequately regulated with CoP location and GRF [27, 30] where ankle moment (shown in Figure 3(c)) was postulated to play an important role [30]. Moreover, the rotational inertia about CoM was shown to be modulated during overground walking [27] where the rotational inertia of the whole body is a property of the distributed masses of the limbs, and, by ignoring it, unnatural constraints, such as zero angular momentum at the CoM and resultant GRF collinear with the lean line, are forced on to the model. The reaction mass pendulum (RMP) model augments the traditional point-mass pendulum model by capturing the shape, size, and orientation of the aggregate rotational centroidal inertia [27], as shown in Figure 2(b).

2.3. Baseline, Postintervention, and 2-Day-Postintervention Assessments. The subjects practiced overground FES-assisted stepping with EMG biofeedback during stand-to-walk transition for 1 hour with sufficient rest inbetween (roughly 15 trials each subject), where the stimulation to either the paretic TA or GM was triggered for 300 ms with LE from the same muscle during stepping action. The LE was monitored during the deficient gait phase (roughly 45%–60% of able-bodied gait cycle) found from the kinematics data (from MS Kinect) where the subject was asked to activate the muscle (TA or GM) using EMG biofeedback (Figure 1(d)) during the deficient gait phase. For comparison between the speed matched (but not age matched) able bodied and pathological gait, the gait cycle (GC) was time normalized to 0–100 percent from heel strike to heel strike of the paretic (nondominant for able body) leg which was also the initiator leg for stand-to-walk transition. Here, 0–15% was the double-support phase following the foot strike of the paretic (or nondominant for able bodied) leg, 15–45% was the single-support phase of the paretic leg, 45–60% was the double-support phase of the nonparetic (or, dominant for able bodied) leg, and 60–100% was the single-support phase of the nonparetic leg. The foot strikes and foot offs were manually verified from the video data for each trial for all the participants before segmentation. The angular momentum was normalized by the product of each participant's mass, CoM height and self-selected gait speed [23]. Since the EMG biofeedback was provided from 45–60% of the GC we hypothesized that the stroke survivors will be able to regulate CAM during this deficient gait phase where a plot of the deviation in CAM from the normative value, ΔCAM, at Nth% GC versus $(N + 1)$th% GC during 45–60% GC may elucidate training effects.

During the baseline (preintervention), postintervention, and 2-day-postintervention EMG assessments, a total of 5 minutes of surface EMG was recorded from bilateral GM and TA muscles while all the subjects walked without FES-assistance on a treadmill at ~1 m/sec (comfortable considering their self-selected gait speed of 0.84 ± 0.18 m/sec). Also, speed-matched (but not age matched) EMG data was collected from four able-body volunteers for comparison. Kinematics data could not be collected with MS Kinect during treadmill walking due to issues with "Skeleton Tracking" at leg crossings. We are currently improving the "Skeleton Tracking" to capture kinematics data during treadmill walking using MS Kinect.

The gait cycle (GC) was divided into 100 equal segments from heel strike to heel strike of the paretic limb (nondominant limb for able body) and 250 of such GCs were collected

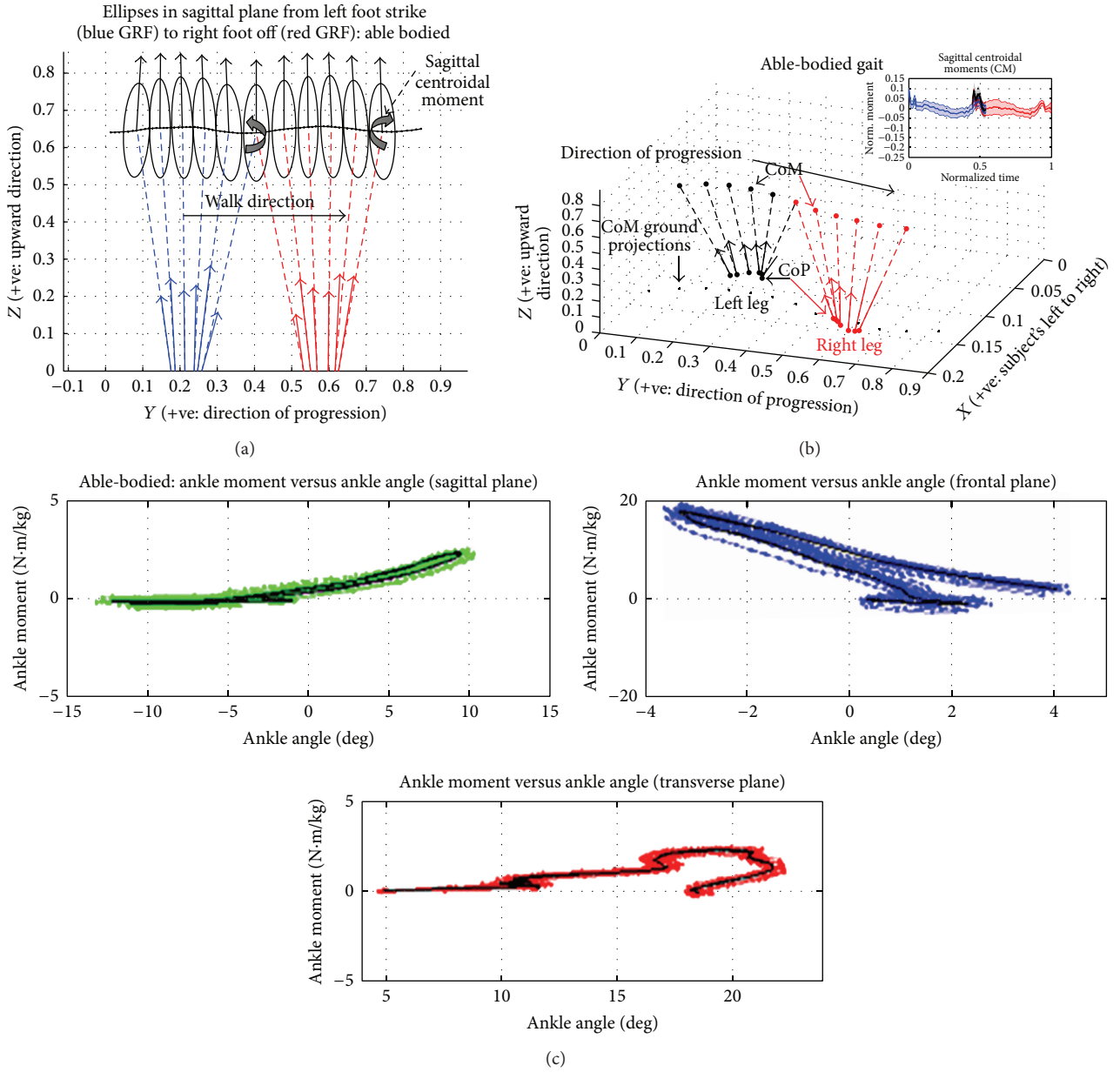


FIGURE 3: (a) Evolution of Reaction Mass Pendulum model in sagittal plane for able-bodied along with the centroidal moment created by the ground reaction forces (GRF). (b) Typical able-bodied gait with center of mass (CoM) and center of pressure (CoP) trajectories during a gait cycle. (c) Ankle moment versus ankle angle plots during able-bodied gait cycle.

for each subject during 5 minutes of treadmill walking. The LEs from the paretic ankle muscles (GM and TA) were plotted over a GC against the corresponding LE from the nondominant leg of able-body subjects (called stroke-able Cyclogram henceforth). If the Cyclogram data points lie entirely on a 45° straight line passing through the origin (called symmetry line henceforth) then the corresponding LEs from the stroke and the able-body subjects are symmetric and synchronized [31]. The coordination between paretic TA and GM was evaluated with flexor-extensor (TA-GM) Cyclogram where the LEs from the GM and TA were plotted against each other [32]. The TA-GM Cyclogram was compared segment-by-segment of the gait cycle over all the GCs between stroke and able-body subjects using the following distance measure:

$$\text{dist} = \sqrt{E\left[\left(LE_{GM,stroke} - LE^{mean}_{GM,able}\right)^2 + \left(LE_{TA,stroke} - LE^{mean}_{TA,able}\right)^2\right]}, \quad (3)$$

where the symbol E denotes expectation (or mean), LE is the linear envelope of EMG from GM or TA of the stroke subjects, and LE^{mean} is the ensemble average ($N = 250 \times 4$) of LEs from GM or TA of the able-body subjects. Two-way (6 stroke

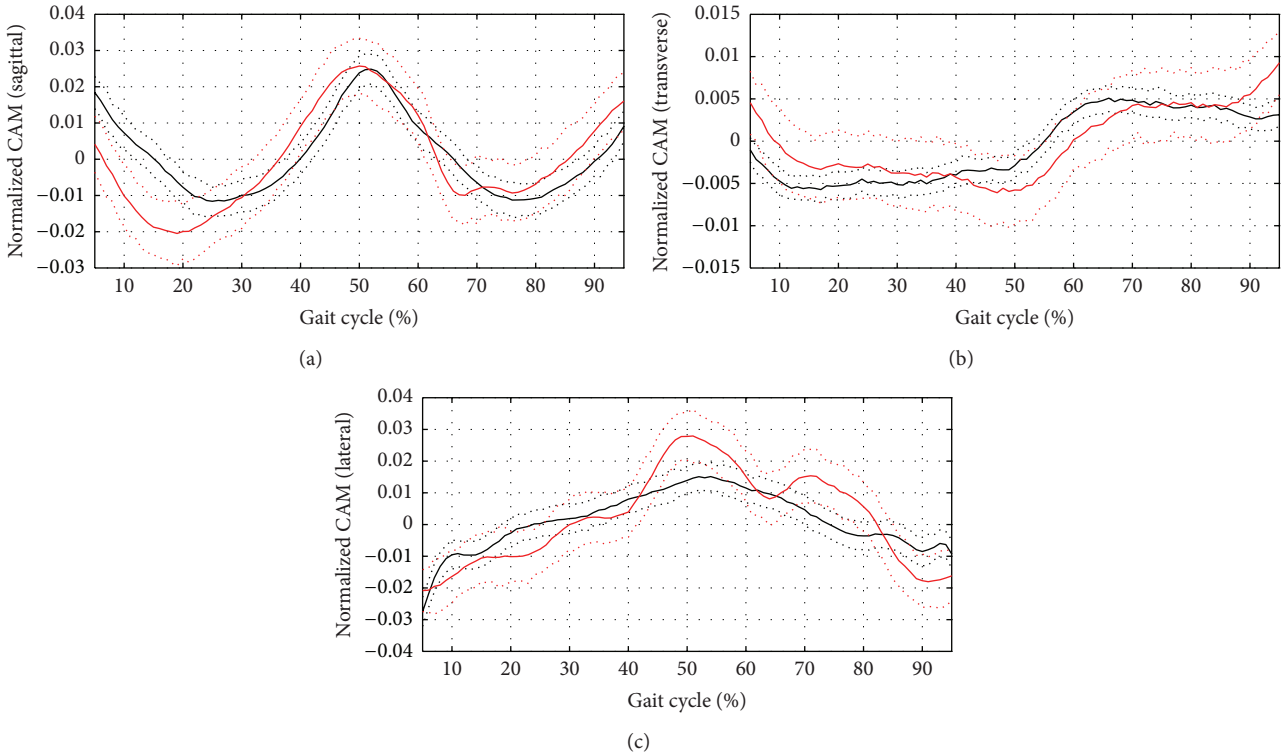


Figure 4: Whole body normalized centroidal angular momentum (CAM) during poststroke (red color) and able-bodied (black color) gait cycle during stand-to-walk transition in the sagittal (a), transverse (b), and lateral (c) planes of gait. Dotted lines show ±1 standard deviation.

subjects × 3 assessments) analysis of variance (ANOVA) was performed on dist measure to evaluate the hypothesis that the subjects, assessments, and interaction effects are all the same, against the alternative that they are not all the same.

Also, improvement in volitional overground gait was determined after comparing baseline with post- and 2-day-postintervention using video-based observational gait analysis (OGA) [33–35]. Since interrater agreement is much higher with a binary scale [34, 35] a trained physiotherapist rated improvement at the ankle, knee, hip, pelvis, and trunk as "present" or "absent."

3. Results

An illustrative whole body normalized centroidal angular momentum (CAM) during poststroke (red color) and able-bodied (black color) gait cycle (GC) during stand-to-walk transition in the sagittal (a), transverse (b), and lateral (c) planes of gait is shown in Figure 4. Figure 4 shows only 5–95% GC since manual examination of the gait cycle in the video data indicated that MS Kinect could not consistently capture the whole body "Skeleton Tracking" data at the boundaries of the capture volume for some trials. The dotted lines show ±1 standard deviation which was found reasonable for 5–95% GC. Although the whole body CAM looked similar between the poststroke and able-bodied subjects but video-based OGA indicated abnormal double-support gait phase due to compensatory movements. Therefore, to capture this abnormality due to compensatory movements of the nonparetic side, we analyzed the CAM of the paretic side (i.e., initiator limb side), the nonparetic side (i.e., follower limb side), and the head and trunk (i.e., head + trunk) separately. As shown in the Figure 5, the normalized CAM for the initiator limb side (a) included that summed from paretic foot, shank, thigh, forearm, upper arm, and the normalized CAM for the follower limb side, (b) included that summed from nonparetic foot, shank, thigh, forearm, upper arm, and the normalized CAM for the head + trunk, and (c) included that summed from head and trunk. In Figure 5, the dissimilarity between the poststroke and able-bodied subjects is more evident during double-support gait phases, that is, 0–15% and 45–60% GC. We then investigated if the EMG biofeedback provided from 45–60% of the GC during EMG-triggered FES-assisted training for stand-to-walk transition helped the stroke survivors to regulate CAM during this deficient gait phase where a plot of the deviation in CAM from the normative value (i.e., mean of able-bodied data), ΔCAM, at Nth% GC versus $(N + 1)$th% GC during 45–60% GC is shown in Figure 6.

Figure 7 shows the mean linear envelope (LE) from TA and GM muscles normalized by their maximum voluntary contraction, ensemble averaged over 250 gait cycles during treadmill walking. Different colors represent the 6 stroke subjects and different markers represent the 4 able-body subjects. Figure 7(a) shows the baseline (preintervention) TA-GM Cyclogram for able-body and stroke subjects in the top panel and baseline stroke-able Cyclogram for TA and GM in the bottom panel. Figures 7(b) and 7(c) show the same for

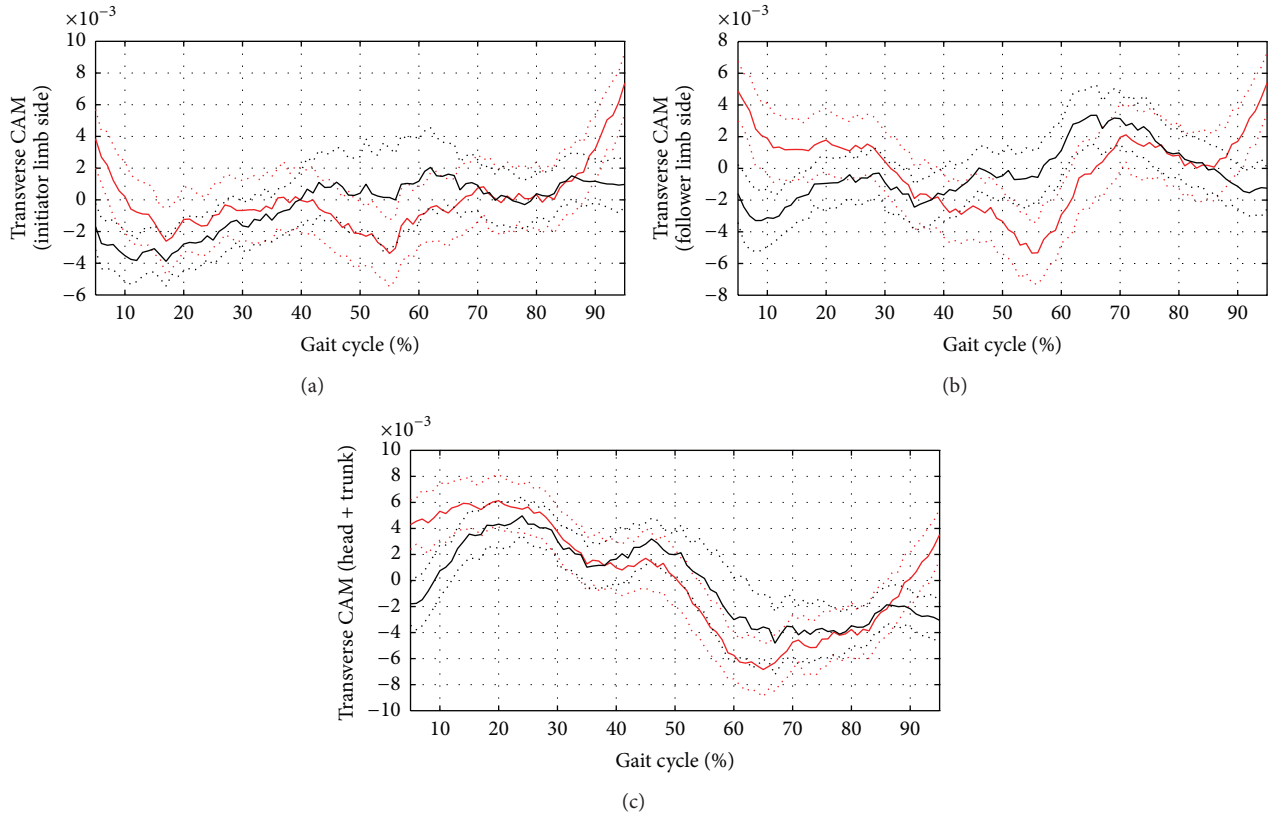


Figure 5: (a) Summed normalized centroidal angular momentum (CAM) from initiator limb side (ILS), that is, paretic foot, shank, thigh, forearm, and upper arm. (b) Summed normalized CAM from follower limb side (FLS), that is, nonparetic foot, shank, thigh, forearm, and upper arm. (c) Summed normalized CAM from head + trunk, head and trunk. Dotted lines show ±1 standard deviation.

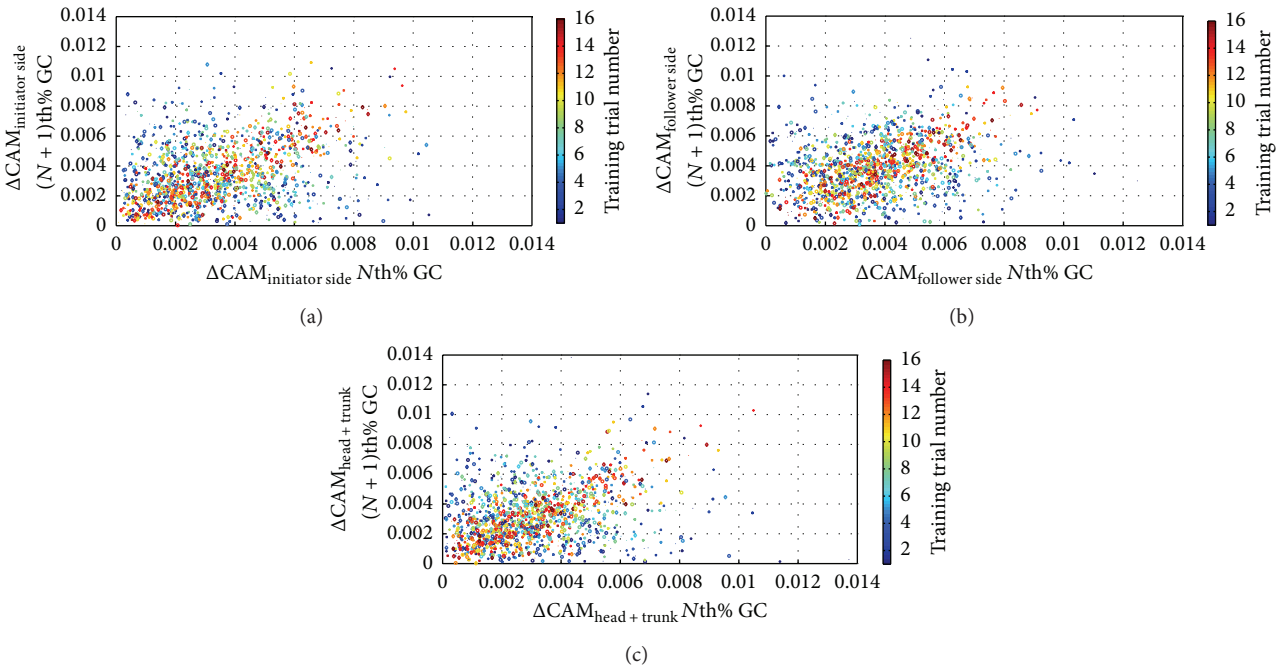
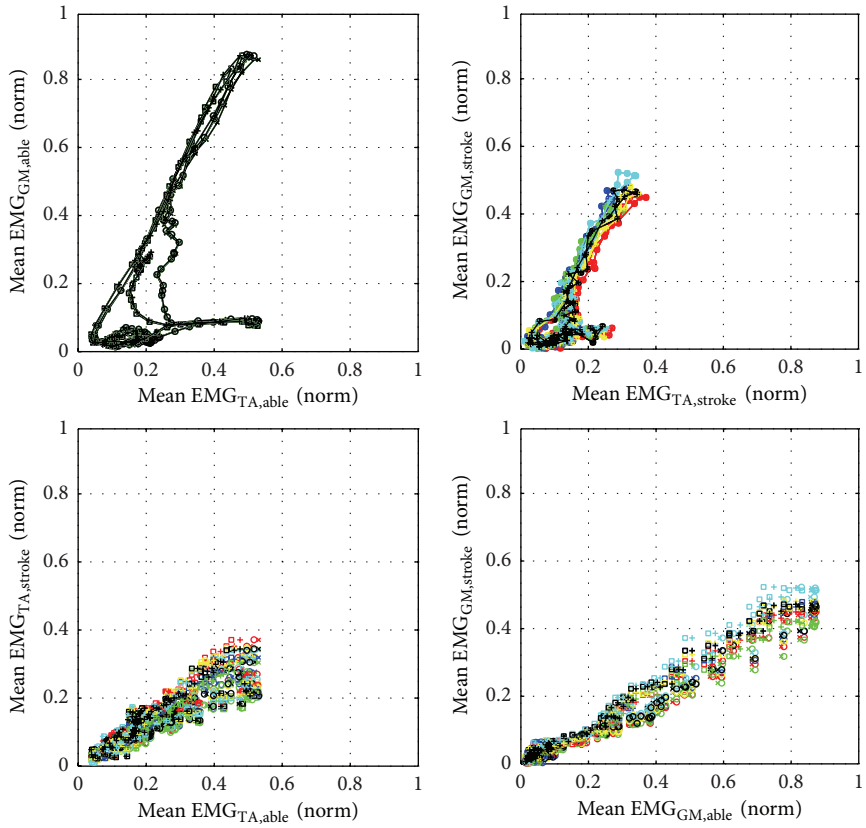
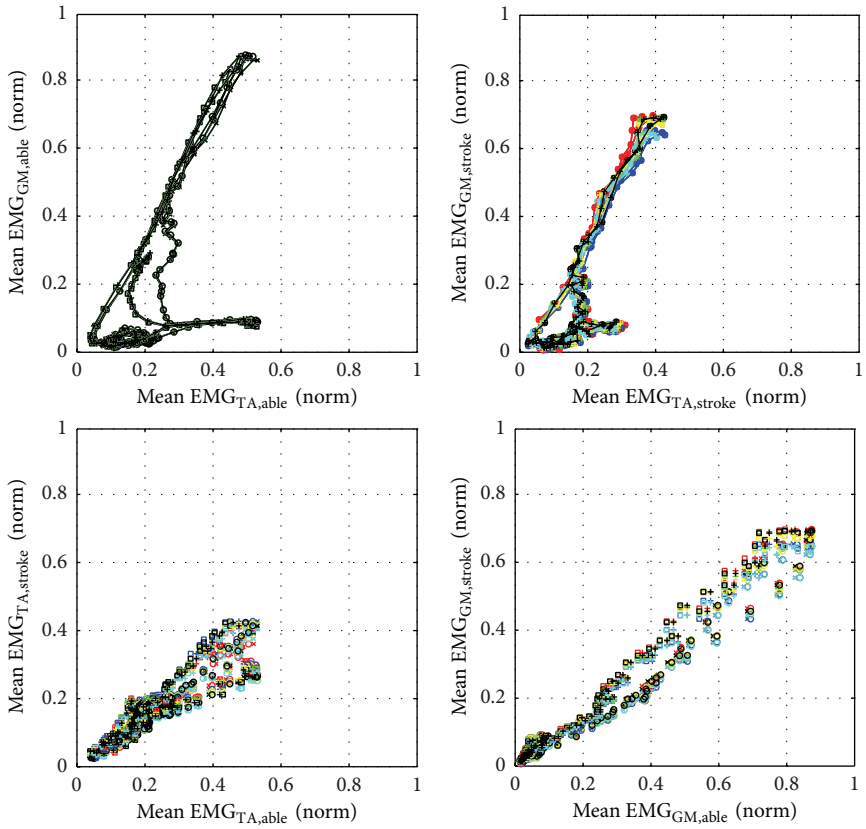


Figure 6: Plot of the deviation in normalized CAM from the normative value, ΔCAM, at Nth% GC versus (N + 1)th% GC during 45–60% GC for the paretic initiator side (a), nonparetic follower side (b), and head + trunk (c). The color map shows the training trial number where higher trial number indicates more training. Poststroke subjects are represented with different markers in the plot.

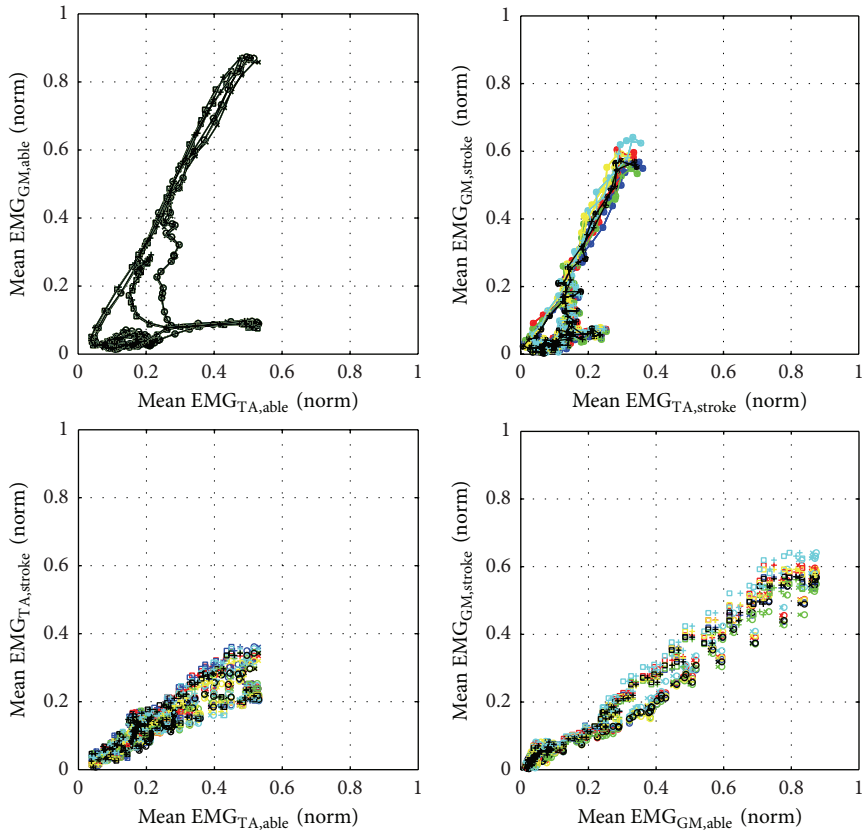


(a)



(b)

Figure 7: Continued.



(c)

Figure 7: Ensemble averaged linear envelope of EMG (mean EMG), normalized (norm) by the maximum voluntary contraction from tibialis anterior (TA) and medial gastrocnemius (GM) muscles over 250 gait cycles during treadmill walking (1 m/s) for able-body (able) and stroke (stroke) subjects. Colors denote 6 stroke subjects and the markers denote 4 able-body subjects. (a) Baseline flexor-extensor (TA-GM) Cyclograms, (b) postintervention flexor-extensor (TA-GM) Cyclograms, and (c) 2-day postintervention flexor-extensor (TA-GM) Cyclograms. [22].

Table 1: ANOVA table (source: source of variability, SS: sum of squares, df: degrees of freedom, MS: mean squares, F: F-statistics).

Source	SS	df	MS	F	Prob > F
Subjects	0.01	5	0.001	3.76	0.002
Assessments	9.39	2	4.697	15739.6	0
Interaction	0.03	10	0.003	11.093	$6.21E-19$
Error	1.34	4482	$3E-04$		
Total	10.8	4499			

postintervention and 2-day-postintervention assessments. The stroke-able Cyclograms show deviations from the diagonal symmetry line, especially at higher magnitudes of LE for both the TA and the GM muscles. The distance measure (dist) for the TA-GM Cyclogram was 0.31 ± 0.02 at baseline, 0.19 ± 0.02 at postintervention, and 0.27 ± 0.02 at 2-day-postintervation. Two-way ANOVA ("anova2", Matlab, The Mathworks Inc.) of dist showed effects in terms of P values, as tabulated in Table 1. These values indicate that both subjects and assessments paretic the distance measure, and there was also evidence of a synergistic (interaction) effect of the two. The video-based OGA showed visible improvement at ankle and knee during overground walking during postintervention assessments when compared to baseline, with increased knee flexion during swing phase and increased plantar flexion at foot off resulting in improved ground clearance of the foot. The improvements however were "absent" during 2-day-postintervention assessments when compared to baseline.

4. Discussion

Stand-to-walk transition can be initiated with foot off of any one of the limbs, which is called the initiator limb [36]. The transient state from standing to steady-state walking is defined differently by different researchers [37–39]. Miller and Verstraete showed that steady state in terms of total mechanical energy of the body was reached by the end of three full steps [36]. Dutta et al. investigated the periodicity of kinematic joint trajectories and showed that the quasiperiodic behavior found during able-bodied steady state walking was not attained during first five steps of functional electrical stimulation- (FES)-assisted walking following partial paralysis [12]. In this study, we investigated

the centroidal angular momentum (CAM) during the first gait cycle of stand-to-walk transition with the paretic side as the initiator limb. Based on our prior work [12], it can be postulated that hemiplegic poststroke subjects with muscle weakness and coordination deficits will take it longer to establish quasiperiodic behavior of CAM during stand-to-walk transition when compared to age-matched able-bodied subjects where CAM can be regulated with CoP location (i.e., foot placement) as well as by modulating GRF (i.e., joint moments). In fact, recent findings characterize a component of walking motor control, describe how typical foot roll-over contributes to postural control, and provide a rationale for the increased fall risk observed in individuals with atypical ankle muscle function [40]. Here, appropriate foot placement (called capture point) following the stepping action is necessary such that CAM can be regulated from a significant nonzero value during (controlled) freefall single-support phase towards zero during double-support phase [41]. We postulate that the ankle-foot complex plays an important role in modulating the GRF during able-bodied gait [27] by absorbing/dissipating energy during foot strike phase and injecting energy during push off phase of walking, as it transforms from a relatively pliable state to a more rigid structure [42], as shown with sagittal ankle moment versus ankle angle plots for able-bodied subjects in Figure 3(c). This change in quasistiffness of the ankle-foot complex and the propulsive work by the ankle moment in the stance phase of walking, as elucidated by Shamaei et al. [42], is primarily regulated with muscle activation patterns that need to be targeted during poststroke gait rehabilitation. In fact, in order to recover from fall (or controlled fall during single-support phase), the foot should be placed at a capture point [41] with the ankle-foot complex having appropriate quasistiffness [42] such that the ankle moment can generate sufficient impulsive CM at the CoM using the GRF to bring the CAM close to zero.

Figure 4 showed that the whole body normalized CAM during poststroke and able-bodied gait cycle during stand-to-walk transition did not differ as much as the summed normalized CAM for the paretic initiator limb side (ILS), nonparetic follower limb side (FLS), and the head + trunk, as shown in Figure 5. It should be noted that the post-stroke participants in this study were community ambulatory where they had abnormal double-support gait phase due to compensatory movements that were identified from video OGA. This abnormality was identified in Figure 5 when we computed normalized CAM for the paretic side and the nonparetic side separately. The goal of the EMG-triggered FES-assisted therapy for stand-to-walk transition was to alleviate this abnormality which was more pronounced in the video OGA during the acceleration phase of the overground gait. Here, we investigated this training effect with the plot of the deviation in CAM from the normative value (i.e., mean of able-bodied data), ΔCAM, at Nth% GC versus $(N+1)$th% GC during 45–60% GC (i.e., double support phase of the paretic limb) during stand-to-walk transition, as shown in Figure 6. From Figure 6, it was found that the EMG-triggered FES-assisted therapy for stand-to-walk transition helped in bringing the data points close to 45° line with more training trials, which indicated convergence to a quasisteady state ΔCAM. A quasisteady state ΔCAM can be interpreted as a necessary deviation in the CAM from a normative value due to poststroke muscle weaknesses. Nevertheless, there was some convergence in the CAM towards zero with more training trials, especially in the cases of paretic initiator side (a) and head + trunk (c) which may be subject specific. Here, the EMG biofeedback for EMG-triggered FES-assisted training during stand-to-walk transition focused on the weight transfer and forward propulsion with the paretic limb during stepping action, targeting "synergies" 1 and 2, where the normalized CAM during that deficient gait phase (~45–60% GC) improved slightly in some subjects. We are currently investigating the sensitivity of CAM to disturbing perturbations due to improper (from an ideal capture point [41]) foot placement where hemiplegic subjects also suffered from poor coordination of muscle activity in the rectus femoris, tibialis anterior, and hamstrings (i.e., "synergies" 3 and 4 [22]) leading to "uncontrolled" foot placement following the swing phase, in addition to poor synergistic action of soleus and medial gastrocnemius (i.e., "synergies" 1 and 2 [22]) to generate "appropriate" quasistiffness and propulsive energy with the ankle-foot complex during the stance phase of the paretic side. Here, the ideal capture point [41] can be computed in real time from CAM changes that can be monitored with MS Kinect where a minimum step length/width can be provided as a biofeedback to the subject for training during EMG-triggered FES-assisted gait therapy.

The most significant result of this study was short-term improvement in the paretic ankle flexor-extensor (TA-GM) coordination with EMG-triggered FES-assisted therapy, which is shown in Figure 7 and Table 1. We found immediate improvement in paretic ankle flexor-extensor (TA-GM) coordination during volitional treadmill walking following task-specific EMG-triggered FES training that targeted TA and GM during weight transfer and forward propulsion with the paretic limb. During video-based OGA, we found that an improved volitional plantar flexion at the terminal stance for push off improved the propulsion of the paretic limb forward during the swing phase with improved knee flexion. In fact, our prior study on two weeks of EMG-triggered FES training showed neurophysiological changes in the plantarflexor (GM) muscle which was used to trigger FES [21]. Therefore, this study showed that just one session of EMG-triggered FES training could improve TA-GM coordination which was however lost after 2 days. We also found that flexor-extensor Cyclogram was sensitive in capturing changes in muscle coordination which can be used for real-time biofeedback for training flexor-extensor coordination during EMG-triggered FES training. Moreover, stroke-able Cyclogram was found to be a good visual tool to assess the normality of the paretic LE patterns based on their deviations from the symmetry line. We are currently exploring "dosing" of EMG-triggered FES-assisted gait therapy with a home-based rehabilitation model where stroke survivors can use a low-cost FES device (such as ODFS Pace) with our low-cost biofeedback system providing visual feedback of normative flexor-extensor EMG Cyclogram and appropriate capture point during treadmill walking. Such home-based low-cost FES-therapy system for

chronic poststroke gait rehabilitation is postulated to be more economical and feasible in a low-to-medium resource country like India where well-equipped rehabilitation clinics are expensive, far, and few.

In closing, we would summarize our research hypothesis for stroke neurorehabilitation following Hallet [28], who have presented lessons from animal models where manipulation of environmental, behavioural, and pharmacologic contexts influenced cerebral reorganization and consequently the process of recovery of function following stroke. We propose that early EMG-triggered FES-assisted therapy targeting a "minimal" set of muscle coordination (or, "synergies") exercises may drive functionally relevant neuroplastic changes to effectively merge them to generate a normal volitional gait pattern of the locomotor apparatus.

Conflict of Interests

The authors have no conflict of interests to declare that might raise the question of bias in the work reported or in the conclusions, implications, or opinions stated.

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