

BIOMECHANICAL OPTIMIZATION OF LOWER LIMB ORTHOSES USING DYNAMIC SURFACE POLYELECTROMYOGRAPHY

Mukul Talaty, Rahamim Seliktar, Ph. D.
Biomedical Engineering & Science Institute
Drexel University 32nd & Chestnut Streets; Philadelphia, PA 19104

Alberto Esquenazi, M.D., Barbara Hirai
Gait & Motion Analysis Laboratory
Moss Rehabilitation Hospital; 1200 W. Tabor Road
Philadelphia, PA 19141

ABSTRACT

In the current study, preliminary testing has indicated that a suitably processed electromyography (EMG) signal is sensitive enough to display changes in muscle activity in a normal population due to changes in orthotic alignment. Anterior and posterior motion stops were used to vary alignments (range of motion) of an ankle foot orthosis with which the subjects walked. Integrated values of linear enveloped EMG profiles (as a measure of total muscle activity) of five gait related muscles influenced by the orthosis were statistically compared for various alignments. In four of the five muscles, these profiles were found to differ in the same consistent manner for each of the four subjects. Changes in muscle activity were justified by biomechanical changes in gait. More specifically, mechanically restricted joint motion changes demands placed on gait muscles, and may also alter their activation onset and duration. As a simple example, during early stance the tibia rotates over the foot as the ankle goes from $\sim 15^\circ$ plantarflexion to $\sim 15^\circ$ dorsiflexion. The soleus muscle normally checks this rotation via an eccentric contraction, so a brace which restricts dorsiflexion would reduce this soleus activity. Preliminary results support prior clinical observations that surface EMG is sensitive enough to discriminate between relatively minor changes in muscle activity due to different orthotic alignments. Based on this conclusion, it is believed that EMG may be used to fine tune or optimize orthotic alignment. Further work relating muscle activity changes and overall energy expenditure to minor variations in orthotic alignment is currently underway.

INTRODUCTION

Since EMG is a measure of muscle activity and one function of orthotics is to change muscle force demands, subtle changes in muscle activity due to different orthotic alignments may be detected in the EMG signal. It may be possible to both optimize alignment and make it a more repeatable and standardized procedure. In order to optimize any variable, there must be an optimization function. In our study, this function was based on amount of muscle activity as well as safety and stability of walking. Muscle activity is a function of the duration through which a muscle is "on" as

well as the strength of contraction for that period. It is to be minimized while safety and stability are to be maximized. Although attempts have been made [1] to quantify safety and stability of walking, these characteristics of gait are still largely determined by the clinician's judgment. The proposed optimization technique can in no way remove all subjectivity from the alignment process; however it should add some much needed objectivity. Optimization rules for orthotic alignment may be adapted to many populations with specific deficiencies; this optimization rule has been chosen to suit a particular patient population — Post Polio Syndrome (PPS) patients.

In the decades following recovery from Polio, many survivors experience muscular weakness, neuromuscular fatigue, and pain. These symptoms are often attributed to motor neuron death and muscle atrophy as well as the associated compensatory muscle overuse. In normals, a given muscle may be steadily contracted while many motor units cycle through periods of activity. This phenomenon helps to avoid fatigue in normals for sub-maximal force contractions — like those required for normal walking. But by the same token, it is the lack of this mechanism that leads to "premature" fatigue in populations with reduced motor function. Reduced motor function in the lower limbs engenders muscle weakness and fatigue and thus often compromises an individual's ability to ambulate. The chosen optimization function of minimum muscle activity is directly attributed to this neuromuscular deficiency. Effective treatment in such cases often consists of using an orthotic device to alter weight distribution, provide support, and restrict motion of joints which can thereby decrease demands on supporting and stabilizing muscles. An orthosis can be configured to match the individual's needs and abilities. In order to suitably decrease muscle activity without compromising walking safety and efficiency, the orthosis must be properly aligned.

There are many aspects to alignment, such as position of the anatomical joint with respect to the mechanical joint, amount of foot inversion/eversion or heel height. In this study only the ankle joint range of motion (plantar/dorsi-flexion) is varied since it can be easily modified as needed in a double upright orthosis. Current clinical practice to determine final orthotic alignment is a trial and error process which relies largely on a clinician's ability to apply principles of biomechanics and to integrate a pseudo-quantitative gait analysis and patient feedback. Gait analysis, which may consist of measurement and interpretation of kinematic and kinetic walking parameters as well as dynamic EMG, is often needed to ensure acceptability of the overall gait pattern. A raw or minimally processed EMG profile may thus be used only indirectly, if at all, to determine final alignment of a brace. The method is considerably subjective and qualitative, and so orthotic alignment may vary substantially from one location to another. Patient care is not as uniform as it may be, perhaps because there is no standard repeatable procedure for alignment. A measure of objectivity, such as the quantitative assessment of EMG, in the orthotic alignment protocol may be beneficial, but the EMG signal must be suitably processed to reflect appropriate changes in muscle activity caused by an orthosis.

A processed EMG signal which identifies when muscle activity begins/ends and which pro-

vides some indication of the level of effort at the motor unit or muscle level is needed to make a comparison of activity levels [2]. The technique should yield stable, signature EMG patterns by reducing the variability of EMG. EMG is inherently variable, in part, because (a) motor units stimulate an average number of muscle fibers, (b) there is uncertainty in spatial and temporal activation patterns of muscle fibers and (c) muscle action potentials have a limited spatial range [3]. There are many currently used EMG processing techniques. Integration provides a measure of total EMG activity, but masks phasic and amplitude information [4,5]. Integration with time reset may provide some phasic information, but the choice of reset time is critical and subjective [6]. Both techniques circumvent the arbitrary choice of a threshold to gauge periods of muscle activation, although advances in thresholding have made it more objective [7]. Moving window averaging techniques are simple to implement and produce no phase shifts in the signal, but they have not gained the popularity of the digital filter, perhaps because of the filter's history of reliability and well defined easily tailored parameters. The linear envelope (rectifying followed by low pass filtering) generates an EMG profile which closely follows the phasic nature of the raw EMG [7,8,9] while preserving the relative amplitude relationships. Bodem et. al [3] claim, "...suitable averaging of EMG activity ... in gait may be able to show influence of various conditions on muscle activity." In ensemble averaging, analogous segments of EMG signals are combined to create an average profile; this facilitates comparison of EMGs by reducing signal variability and by de-emphasizing singular, erroneous information — such as motion artifacts or random muscle twitch — which may be misleading in a single EMG trace. Periodic similarities of EMG are accentuated; however activity which is not physiological in origin, but still occurs regularly will also be highlighted. Using the ensemble averaged EMG profile in combination with a statistical comparison of integrated linear envelopes of EMGs from different test conditions allows for judicious comparison of muscle activity changes caused by different orthotic alignments.

The effect of varying alignments of an ankle foot orthosis on parameters of gait is not new [10,11]. Furthermore, the energy cost changes with and without orthotics for normal and pathological populations as well as that of orthoses in various conditions have been documented [10,12,13, 14]. But integrating these ideas — i.e. using physiological data to determine alignment — has not been documented. To reiterate the basis for this technique is that since EMG is a measure of muscle activity, it may be used to minimize muscular demands.

METHODS

Preliminary experimental work was performed in the Gait & Motion Analysis Laboratory at Moss Rehabilitation Hospital. Four normal subjects (i.e. those with no known pathological conditions affecting gait) were fitted with a unilateral double upright ankle-foot orthosis with anterior/

posterior motion stops. They walked in each of four different test conditions (orthotic alignments) which consisted of varying only amounts of motion restriction.

Kinematic data and surface EMG from tibialis anterior, soleus, vastus medialis, vastus lateralis, and rectus femoris muscles were collected from each subject. EMG electrodes and motion analysis markers were affixed, after suitable cleaning and abrading of the necessary areas, on the leg which was braced. Amplifier gain was subjectively chosen by making the real time EMG trace full-field on the oscilloscope. EMG and footswitch data were sampled at 1000Hz. A tachometer was engaged to record walking speed. Prior to data collection subjects were instructed to walk for 5-10 minutes, for each alignment of the orthosis, to get accustomed to the orthosis. Then subjects were asked to walk as naturally as possible for 4-6 runs along the 30 foot walkway. Data collection was manually begun after the subject had taken several strides so that the transient phase of gait initiation was excluded from the recorded data. Duration of collection was set appropriately to terminate before the transient phase of gait termination was reached. The subject repeated the trials until data from about 10 steady state strides had been recorded.

In the control condition the orthosis provided no motion restriction and subjects could plantarflex and dorsiflex the ankle as much as required for their usual gait. In subsequent trials, ankle motion was limited to plantarflexion/dorsiflexion (PF/DF) pairings of 5/5, 5/0, 10/-5 (in degrees). Static range of motion angles of the brace were determined prior to each different test condition using four markers affixed on the brace and a manual goniometer.

DATA PROCESSING

EMG recordings were rectified, filtered, interpolated, and averaged to create ensemble averaged profiles. The following procedure was used: Sections of EMG from heel strike to heel strike were selected from the recorded data. The linear envelopes (i.e. full wave rectification followed by low pass filtering; digital filter specifications: Butterworth 2nd order filter with cutoff frequency, $f_c = 5$ Hz and sampling frequency, $f_s = 1000$ Hz) of these stride EMGs were then generated. The filtered stride EMGs were juxtaposed and unmistakably errant strides were removed; remaining strides were each normalized on a time basis to represent 100% of stride. Normalization was done by a linear interpolation routine which generated 101 (0 to 100%) values of the EMG profile based on the roughly 1000 EMG data values for a single stride. For example, if a stride EMG consisted of 1130 values, the first sample would represent the 0% value, and the $1130/100 = 11.3$ rd sample would represent the 1% value. Then all the 0% values (for a given muscle, test condition, subject) were averaged and stored in an array; this was repeated for the 1%, 2%, ... 100% values from all the strides until there were only 101 values which was the final ensemble averaged profile. Ensemble averaged profiles were generated from the data of only one subject, one muscle, and one test condition at a time. No intersubject profiles were generated.

For statistical analysis, the linear envelopes of all strides of each muscle for an individual in one test condition were numerically integrated using Simpson's rule with initial condition $X(-1) = X(0)$ and final condition $X(102) = X(101)$. These boundary condition assumptions were justified since the EMG linear envelope is not an independent curve — i.e. adjacent values $\{X(t) \text{ and } X(t+1)\}$ are dependent.

RESULTS AND ANALYSIS

Because of the reduced motor capacity of the target population, the optimality criterion of minimum muscle activity was chosen. The less time a muscle was activated would imply less neural input. Also, the lower force (indicated by smaller EMG amplitudes) demand placed on a muscle would suggest less neural input. These two criteria were used to gauge amount of muscle activity. For analysis, it was necessary to be able to evaluate both of these criteria. Integrating the linear enveloped EMG waveforms provided a measure of total activity. But theoretically, it is possible for low-level, long duration muscle activity to produce a lower activity rating than short burst-type muscle activity. The former condition may actually require more neural activity and thus would be less beneficial. In order to exclude this rather extreme scenario and any more realistic variations of it, it was important to be able to assess the phasic and amplitude characteristics of the EMGs as well. Ensemble averaged profiles were suitable for this reason. An actual profile, shown below in figure 1, provides an example of a more typical and more ideal case. It shows clear delay in activation as well as reduced amplitude for test Condition 2. Statistical comparison of total activity (integrated waveform comparison) yielded a value of $p = 0.003$ which indicates a significant reduction in activity for Condition 2.

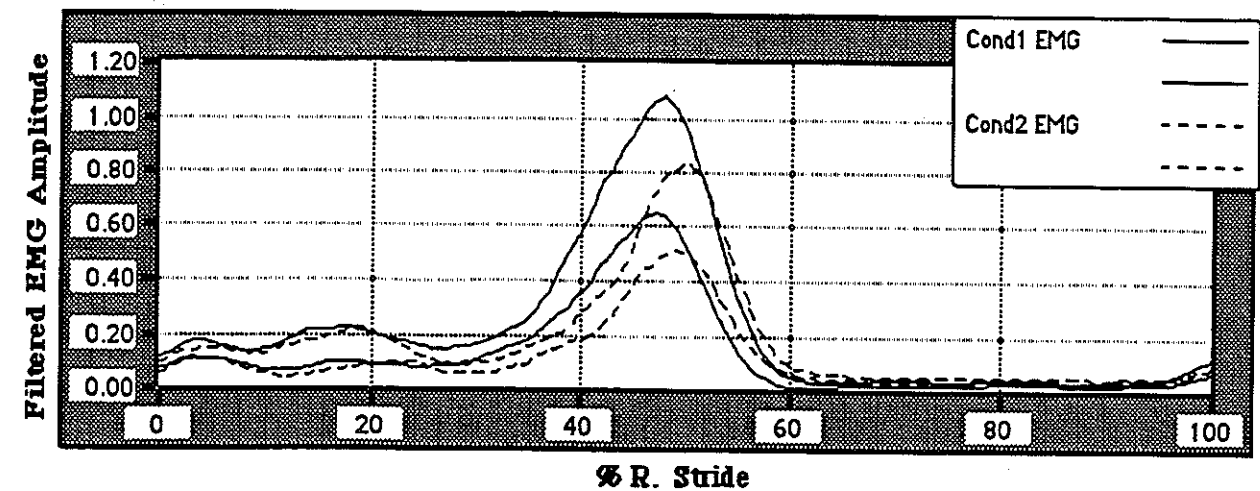


Figure 1. ± 1 Standard Deviation Bands of the Ensemble Averaged EMG from Two Test Conditions

Comparisons were made between the ± 1 standard deviation band of the ensemble averaged EMG profiles for each muscle in the various test conditions. The linear envelope of a single stride EMG and even the ensemble averaged EMG trace (which for a given test condition is represented by a single curve from 0 to 100%) would invariably differ between two test conditions. In fact, it would most likely differ even for two different profiles from the same test condition simply due to normal variations inherent in EMG (4,10). Comparing the ± 1 standard deviation band allows for a more confident statement of changes in activity levels between two test conditions (3).

Since each subject was given nearly (within limits of manual setting of ankle angle restriction of the orthosis) the same alignments, EMG activity was expected to change in a consistent manner for all the test subjects. "Consistent" is used to indicate that the direction of change in the magnitude of the integrated linear enveloped traces (i.e. muscle activity) was the same for all the subjects. Statistical comparisons (t-test) of integrated EMGs from two test conditions showed generally consistent intersubject changes. Intrasubject walking velocities did not change significantly as alignment conditions changed. In comparing Condition D [range of motion : $10^\circ/-5^\circ$ (PF/DF)] and the control, Condition A [range of motion : unrestricted/unrestricted (PF/DF)], nearly the same (all statistically significant or not significant) muscle activity changes could be seen for all subjects in rectus femoris, vastus lateralis, vastus medialis, and soleus muscles. For tibialis anterior, all subjects showed a decrease in activity (in condition D), but only two of the four subjects showed a statistically significant change. The results for this case are summarized in Table 1 below.

For example, all subjects showed a decrease in the soleus activity during approximately 35 - 55% stride in test condition D in which only -5° of dorsiflexion was allowed. Normally the ankle

dorsiflexes to ~15° during midstance, and soleus checks this by an eccentric contraction, but the orthosis inhibited this motion. So, the decrease in soleus activity is justified by the physical limitations. Rectus femoris is used to check knee flexion, and a more plantarflexed (less dorsiflexed) brace has been shown to reduce knee flexion moments [15], so decreased rectus femoris activity may be expected. Tibialis anterior activity may be explained by the reduction in motion permitted from heel strike to foot flat — when it is active to ease the foot on to the floor.

TABLE 1. p VALUES FOR t-TEST COMPARISONS OF TOTAL MUSCLE ACTIVITY IN CONDITION A AND CONDITION D

Muscle Name ⇒ Patient Code ↓	Tibialis Anterior	Soleus	Vastus Medialis	Vastus Lateralis	Rectus Femoris
AE	0.1000 ^a	0.0030	0.0610 ^a	0.3840	0.4410
TW	0.0094	4.3E-5 ^c	0.4370	0.1640	0.0130 ^g
MB	0.0890	1.9E-4	0.1470	0.4130	0.0050
SD	2.0E-5 ^{b,c}	0.0540 ^{b,d}	0.3070 ^{b,t}	0.0640 ^b	1.1E-5 ^{b,h}

** SHADED BOXES WERE NOT JUDGED STATISTICALLY DIFFERENT

KEY TO TABLE 1 NOTES:

- a - used 0-20% stride only due to uneven baseline levels for rest of stride
- b - normalized conditions before comparison
- c - used 0-60% stride only
- d - used 25-58% stride only due to uneven baseline levels for rest of stride
- e - used 45-70% stride only due to uneven baseline levels for rest of stride
- f - used 0-25% stride only due to uneven baseline levels for rest of stride
- g - used 0-25% stride only due to unusual compensatory mechanism from 30-100% stride
- h - possible loss of electrode coupling may have caused decreased amplitudes in Condition D

CONCLUSIONS

This preliminary study has shown changes in muscle activation patterns due to orthotic alignment can be detected by suitably processed EMG. Statistical analysis of integrated linear envelopes of stride EMGs is used to make quantitative assessments and ±1 standard deviation band of ensemble averaged EMG profiles is used to make visual assessments of muscle activity changes caused by varying orthotic alignments. Furthermore, there was agreement between the biomechanically predicted changes in muscular activity due to motion restriction and the actual muscular activity measurements. For example in the case of limited (-5o) dorsiflexion, t-tests of linear envelopes of soleus

EMG profiles showed reduced activity in each of the subjects with respect to the control run (no motion restraint). This change was expected since tibial rotation in midstance, which is checked by eccentric soleus contraction, is decreased. Suitably processed EMG signal was found to be sensitive enough to reflect muscle activation changes due to orthotic alignment. This strongly suggests that EMG may be used to fine tune or optimize alignment of an orthosis. The EMG-alignment correlation is currently being established in a pathological (PPS) population. Initial results suggest a strong relationship between orthotic alignment and muscle activity changes. The effect of alignment on overall energy expenditure is also being monitored, as is patient feedback on brace settings. Finally, a mathematical model of an orthosis is being developed which will be used to predict changes in muscle activity and the resultant gait biomechanics caused by differing alignments. It is intended to display the physical nature of the changes in muscle demands as alignment is varied, as well as to verify results from the experimental phase of this work.

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