

Dynamic EMG During Walking as an Objective Measurement of Lower Limb Orthotic Alignment

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Abstract

An orthosis may be used to supplement reduced motor function in the lower limbs and associated ambulatory deficiencies. Optimal orthotic alignment (denotes range of ankle motion allowed by the orthosis) uses motor function efficiently yet avoids undue stress to joints and unsafe walking. Mechanically restricted joint motion alters gait muscle activation patterns. Soleus normally checks early stance phase tibial rotation by an eccentric contraction; a brace that restricts dorsiflexion may reduce soleus activity.

In this experimental work, 4 normal subjects and 4 polio survivors were fitted with ankle-foot orthoses. Surface EMG was recorded from five muscles on each subject in several orthotic alignment conditions. Stride EMGs for each muscle, subject and test condition were rectified, filtered, and averaged to create an ensemble averaged EMG profile. ± 1 standard deviation bands of the EMG profiles from various conditions (alignments) were then easily compared.

Quantitative EMG comparison was found to be sensitive enough to indicate changes in muscle activity. Changes were consistent across the normal subjects, and indicated compensation may occur in a standard way. In the polio survivor population an optimization rule that prioritized quadriceps activity reduction was used. Muscle responses to this condition were different after the accommodation period than those measured immediately after setting an alignment. Subject opinions of this condition were generally opposite to that expected, and the condition was rated unfavorably by the subject even though it spared the often overworked quadriceps muscles.

Key words: gait analysis, orthotic alignment, electromyography, optimization, polio.

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A common manifestation of gait dysfunction is unstable or inefficient walking. This may be a primary effect due to the muscular weakness associated with reduced neuromuscular function, or it may be secondary to compensatory techniques used by an individual. Lower limb orthoses are used to relieve pain, immobilize musculoskeletal segments, prevent or correct deformity and improve function during gait in individuals with walking problems. During walking the orthosis serves to provide support, restrict joint motion, affect the load and moment distribution in a joint, and alter the patterns of residual muscle activation. Ideally, for the people with neuromuscular deficiency, this should result in a reduction of muscular demand. These mechanisms of orthotic function are influenced by the orthotic alignment. Spatial design and construction parameters of the orthosis should be selected before fabrication, and influence final orthotic alignment. They are selected to complement a patient's individual capabilities and limitations. Position of the anatomical ankle joint with respect to the orthotic ankle

joint, or the height and depth of a calf band are examples of geometric characteristics; orthotic material is a fabrication characteristic. In the clinical environment, determination of these characteristics is subjective.

Currently, a clinical team relies on their ability to apply principles of biomechanics and to integrate quantitative and qualitative data from non-instrumented gait analysis as well as patient opinion to determine orthotic alignment. Even if the biomechanics were well understood and readily applied by the clinical community, alignment would be a subjective procedure since the fundamental scientific knowledge in this area is incomplete. Furthermore, the various modes of compensation employed by individuals are unclear, and deserve further study. The result of our work in this area would be the development of a simple, easily interpretable method that allows a clinician to select the best alignment from readily obtainable and quantifiable information.

Objectives

An overall goal of this study was to evaluate the use of dynamic EMG during walking as an objective technique in the determination of optimal orthotic alignment. However, a preliminary objective was to determine if EMG processed in the simple method chosen was sensitive enough to show minor differences in muscle use due to minimally different physical conditions. This was a central part of the early work on normal subjects. In the management of polio survivors, a protocol which emphasizes minimization of muscle activity is clinically attractive. Often muscles are overworked without an orthosis but an improperly aligned orthosis may impose unacceptable muscle demands or provide insufficient support in individuals with weakened neuromuscular systems. Demands placed on lower extremity muscles, while walking, change in response to how joint motion is limited by a orthosis. Based on this idea it appears that muscle activity recorded as EMG may provide a useful indicator to measure quality of alignment of the orthosis.

Prior clinical observations in our laboratory have indicated that dynamic EMG may be sensitive enough to discriminate between subtle changes in muscle activity due to different orthotic alignments, although this had not been studied in a more objective method. However, current clinical practice makes indirect use, if at all, of a raw or minimally processed EMG profile to assess orthotic alignment. The body of knowledge on the relationship between orthotic alignment and lower limb muscle demands is incomplete. Further scientific knowledge in this area would facilitate the incorporation of EMG information into orthotic prescription and allow EMG to serve as a simple tool to optimize and standardize orthotic alignment practice.

Surface EMG used to reveal changes in muscle use under controlled conditions was an extension of understanding the physical factors that affect walking, and it has been documented [1]. The effect of varying alignments (using fixed positions, not a range of motion) of an ankle foot orthosis on gait parameters of normal subjects has been studied [2], but in a limited capacity. Some of the more conventional heuristics used in orthotic alignment have been studied. The effect of ankle posture on knee moments has been researched in a limited scope [3, 4]. It is implicit, and confirmed in clinical practice, that plantarflexion of the ankle during stance phase tends to stabilize the knee by preventing knee flexion, and thus may reduce muscle activity of knee extensors. However, changes in muscle activity (as shown in EMG signals) in response to alignment restrictions or other postural constraints have not been successfully quantified, nor have they been shown to occur consistently in normal subjects or other populations.

In the current study, ensemble averaged and integrated EMG profiles from various orthotic alignments obtained during walking from normal volunteers and individuals with residual effects of poliomyelitis were used for a quantitative assessment of changes in muscle activity. This was used to determine an optimal alignment based on an objective selection criteria that emphasized minimization of muscle use and prioritized the quadriceps group.

Methods

Normal subjects

Four normal subjects ranging in age from 24 to 39 years were recruited to walk with a double upright, articulated ankle foot orthosis in various test conditions while lower extremity EMG was recorded. Each subject was unilaterally fitted with an articulated, adjustable ankle foot orthosis. EMG signals from rectus femoris, vastus lateralis, vastus medialis, soleus, and tibialis anterior muscles were recorded using preamplified bipolar surface IOMED EMG electrodes. The electrodes were affixed to the skin with double-sided adhesive tape after cleaning and abrading the contact area. Ipsilateral heel and toe switch data were recorded for event identification during self-selected walking. Walking speed was recorded with a tachometer; data with speed significantly different from the condition average were not used. Subjects were instructed to walk for a short time before data collection to become accustomed to walking with an orthosis. The subject was instructed to walk comfortably along the 10 meter long laboratory walkway, so that approximately 10 steady state strides could be collected. In the baseline trial (unrestricted condition), the orthosis provided no motion restriction. In the test condition, ankle motion was limited to 10° of plantarflexion and -5° of dorsiflexion. Note that -5° dorsiflexion = +5° plantarflexion - in other words, the orthosis moved from 100° to 95° for a total of 5° of motion. Orthosis angles were determined before each trial using a modified manual goniometer. For each new alignment test condition the subject was allowed to walk for 5-10 minutes to become accustomed to the new motion restriction settings before data were collected.

Polio survivors

Polio survivors (PS) were chosen as a test population because they exhibit focal motor weakness. Individuals who used a unilateral articulated adjustable ankle foot orthoses, did not have any recent fractures, surgery, or any type of cardio-respiratory ailments, and did not require upper extremity assistance for ambulation were recruited. Four subjects between the ages of 31 and 60 years were tested. Data were collected as done for the normal subjects described before; in addition, surface EMG of the long heads of biceps femoris was recorded. An overview of testing is presented in Figure 1.

Test Session #1

Subjects walked in their orthosis as the alignment was varied for several different conditions. The originally prescribed alignment was designated condition A (baseline). The test conditions were as follows:

Test Condition	Physical change to range of motion of orthosis*
B	Increased by +4° on the plantarflexion side
C	Increased by +4° on the dorsiflexion side
D	Increased by +6° on the plantarflexion side
E	Increased by +6° on the dorsiflexion side

* with respect to the baseline range which was Condition A

For example if a subject's initial alignment was 10°/5° (plantarflexion/dorsiflexion) which would correspond to a

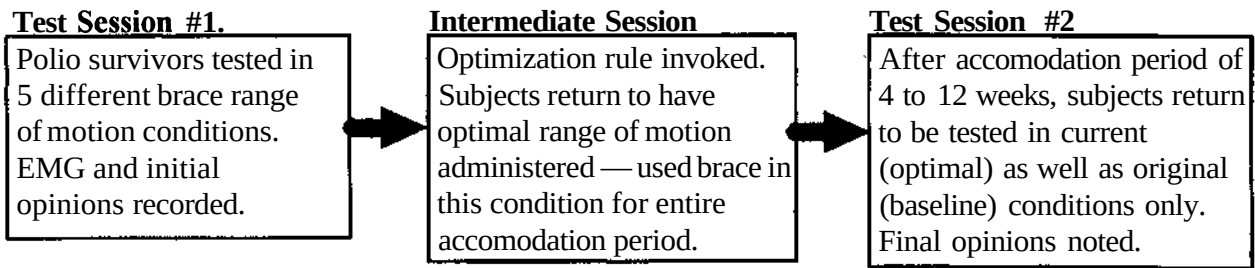


Figure 1. Polio survivors testing flowchart.

range of motion of 100° to 85° , then the test conditions would be $104^{\circ}/85^{\circ}$, $100^{\circ}/81^{\circ}$, $106^{\circ}/85^{\circ}$, and $100^{\circ}/79^{\circ}$. Thus, there were a total of five (four test variations plus one baseline) conditions. Again, subjects were given an adjustment period during which they became accustomed to the new alignment before data were collected. The order of the conditions (including the baseline condition) was randomized to eliminate effects of fatigue and other biases from affecting the outcomes. Subjects were blinded to which condition they were receiving.

Surface EMG data from rectus femoris, vastus medialis and lateralis, tibialis anterior, soleus, and biceps femoris (long head) were collected using preamplified electrodes. Bilateral heel and toe switch data were also recorded. Average walking speed was measured with a tachometer. Data were sampled at 500 Hz. EKG signals were monitored by a Transkinetic EKG telemetry system.

Subjects were given sufficient time to restore resting heart rates before walking in each new test condition. During each resting period, a questionnaire was given to gauge subject opinions about the orthosis in the most recently completed test condition.

Intermediate Processing

Data were processed to determine the optimal alignment. The optimal alignment was determined by a scheme that placed priority on reduction of the quadriceps activity. In short, that alignment which most reduced the activity of

the three surface quadriceps muscles (vastus lateralis, vastus medialis, and rectus femoris) was the best. For example, reduction in activity of two quadriceps muscles was better than a condition that reduced only one quadriceps muscle. However, a condition that increased any quadriceps muscle activity was given last priority, even if it also reduced other quadriceps or other muscle activities. A more complete selection algorithm is shown in the decision chart in Figure 2. Reduction was quantified by the percentage change in measured "activity" values. Differences in muscle activities between most conditions were considerable and so it was never necessary to use data beyond (see Figure 2) the soleus EMG to select the optimal alignment. At this intermediate visit, subjects were given this optimal alignment, and instructed to continue to use the orthosis as they normally would for all daily activities.

Test Session #2

After several weeks (minimum seven), subjects returned for final testing. The same protocol as followed for the initial visit (Test Session #1) was used, except data were only collected for two conditions - the original alignment (baseline of the first visit) and the alignment selected as optimal after the first visit. Each subject was advised not to be excessively more or less active than usual the day before being tested. In this test session, subjects wore the same shoes that they wore during the initial testing. At the

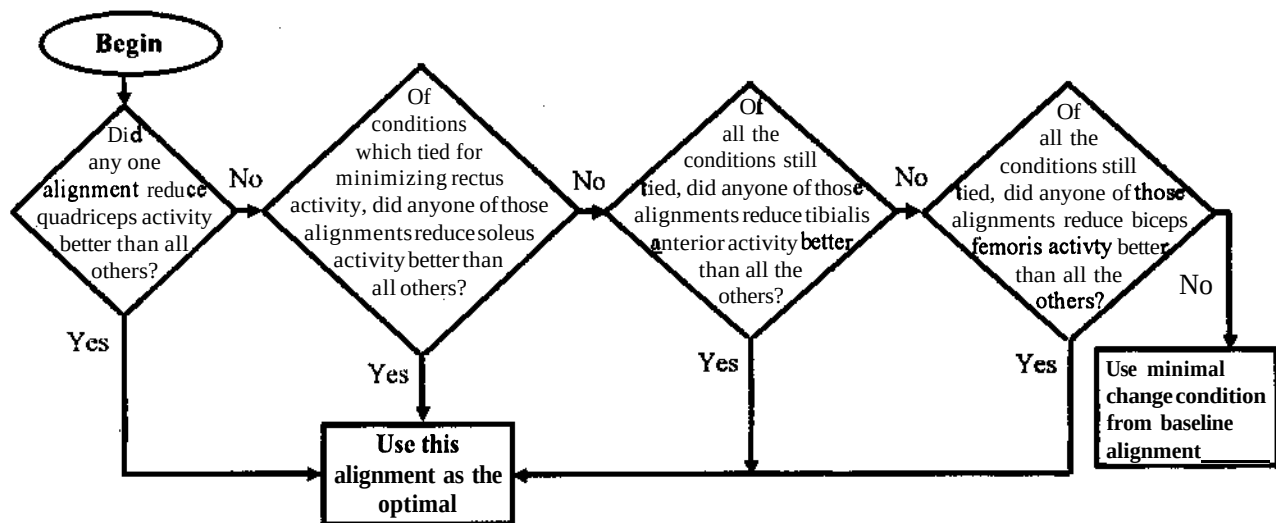


Figure 2. Decision algorithm for selecting optimal alignment.

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Table 1. Outcome possibilities of quantitative EMG signal analysis.

Duration	Amplitude		
	(A) Increase	(B) No Change	(C) Decrease
(1) Increase	No Good	No Good	Unclear*
(2) No Change	No Good	No Change	Good
(3) Decrease	Unclear*	Good	Good

* Indicates that this outcome may or may not produce good results; the data would need to be analyzed in a more individual and perhaps more subjective way to ascertain.

conclusion of the test, the questionnaire was again employed to gauge changes in subject opinions.

Data processing

EMG signals were amplified, band pass filtered using analog filters to remove signal artifact and to prevent aliasing errors, and then A/D converted and sampled by computer at 500 Hz. EMG signals were full-wave rectified and then low pass filtered (Butterworth 2nd order digital filter with cutoff frequency of 5 Hz) producing a linear envelope. Ensemble averaging routines that were written based on the general method noted by Shiavi and Winter [6, 7] were employed. Heel switch data were used to select stride EMG signals from all the EMG data recorded. EMG data from strides with significantly different stride velocity than the average ($\pm 10\%$) were not used. Each stride was time-base normalized to convert the time axis from milliseconds to percentage of stride. A linear interpolation routine was used to generate a time-base normalized linear envelope profile that now consisted of exactly 101 (0 to 100%) values. Each linear envelope was mathematically integrated using Simpson's Rule for integration and the data were output for statistical analysis. Initial and final values for Simpson's method were simply chosen as the 0% and 100% values, respectively, of the normalized linear envelope trace. The average and standard deviation of each 1% of all the normalized linear enveloped waveforms of a given muscle and

Table 2. *p* values for *t*-test comparisons of muscle activity between baseline and test condition in data obtained from normal volunteers.

Subject Code	Muscle Name				
	Tibialis Anterior	Soleus	Vastus Medialis	Vastus Lateralis	Rectus Femoris
AE	0.1000	0.0030	0.0610	0.3840	0.4410
TW	0.0094	0.0000	0.4370	0.1640	0.0130
MB	0.0890	0.0002	0.1470	0.4130	0.0050
SD	0.0000	0.0540	0.3070	0.0640	0.0000

test condition were calculated. These ensemble averaged EMG profiles were logged for future visual comparison. Ensemble averaged profiles were generated for each muscle, each test condition, and each subject.

Results and Discussion

Normal volunteers

Just as Saunders et al. [5] established six determinants of gait that minimize energy expenditure, a similar type of correlation between orthotic alignment and EMG data is a goal of the current study. EMG is an ideal measurement technique as it is already routinely used in gait analysis. Based on clinical observation, it was hypothesized that muscle activation, duration times and amplitudes may change in response to changes in orthotic alignment. Muscle activity is a natural choice for the objective function to be minimized in this study since the target population has decreased neuromuscular abilities, and overuse and compensation symptoms are common concerns. Shorter periods and lower amplitude bursts of activity are desired to reduce neuromuscular demands. The principle methods by which increasing muscular force demands can be met are increases in recruitment and firing rates of motor units. A higher amplitude EMG signal would imply either more motor units or a faster cycling of motor unit activity - both of which may be taxing to polio survivors as well as others with compromised neuromuscular systems. Generally, a combination of reduction in surface EMG signal duration and/or amplitude may be beneficial. This gives rise to several possible combinations of changes in EMG activity, as shown in Table 1. Evaluating the area under the linear enveloped EMG waveform provided a measure that combined duration and magnitude into a single quantity - activity. Cases B3, C2, and C3 of Table 2 were the desired outcomes.

It was theoretically possible for low-amplitude, long duration muscle activity to be found statistically less than large amplitude, short burst-type muscle activity (i.e. outcome case A3 of Table 1). The former condition may be more taxing to the neuromuscular system, and thus would be less desirable. To exclude this rather extreme scenario and any more realistic variations of it from being designated as statistically less activity, it was important to be able to assess the phase and amplitude characteristics of

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the EMG signals in addition to the overall activity or area under the curve. Ensemble averaged profiles were suitable for this task. An actual profile, shown in Figure 3, provides an example of a more typical and more ideal outcome (outcome case C3 from Table 1). It shows a clear delay in activation as well as reduced amplitude for the test condition (Statistical comparison of total activity yielded a value of $p = 0.003$ that confirmed the clear visual assessment of significant reduction for test condition).

One important finding was that in some cases, statistical comparison indicated significant changes between conditions even when the ensemble average did not suggest this based on a visual examination. In a comparison of data from 42 pairs of ensemble averages, the following criteria for a "significant change" was used: (i) for visual analysis, any time the ± 1 standard deviation bands of the ensemble averaged EMG from the two conditions overlapped less than 50% and (ii) for statistical analysis, a p -value of less than or equal to 0.05. Sixty-nine percent of the time, the visual and statistical methods agreed. Twenty-six percent of the time, the statistical method showed a significant change (for at least some part of stride) when the visual analysis did not indicate such a difference. The remaining 5% of the time (2 out of the 42 comparisons), the visual method suggested a significant change whereas the statistical method did not. It is unclear whether the 26% are false positive statistical differences; it may be that not all the information in the EMG signal is easily understood through a cursory visual analysis. This is important as a portion of clinical analysis of EMG data is still done visually.

Overall, the analysis methods outlined provided a means of evaluating the amount of EMG activity during one stride and then comparing all such activity values from one condition to that of another condition. This method provided a consistent, physically significant, and reliable means to compare muscle activity. Area under the linear enveloped EMG waveform combined signal duration and magnitude into an overall measure of muscle activity. These values from the test condition were easily compared to that of the baseline by statistical methods (t -tests) to assess if there were differences in activities. Statistical

outcomes were also checked against a visual assessment of the ensemble averaged EMG data to prevent false or clinically meaningless significance.

Muscle activity changes occurred in a consistent manner for most muscles in the normal subject population. One hundred percent of the vastus lateralis and vastus medialis muscles showed consistent direction and magnitude changes. Seventy-five percent of the rectus femoris and soleus muscles showed consistent changes (statistically). For the tibialis anterior, 100% (all subjects) showed a decrease in activity in the test condition, but only two of the four subjects showed a statistically significant change. Muscle activities were expected to change in a consistent manner for all the test subjects since alignment test conditions were identical (within limits of manual goniometric measurements of ankle foot orthosis range of motion). EMG comparisons are summarized in Table 2.

An admittedly oversimplified approximation of the expected changes in muscle activities was made considering the biomechanics of the test condition. It provided a qualitative means to corroborate the quantitative EMG activity reductions. For example, normally the ankle dorsiflexes to -15° during midstance, and soleus controls this by an eccentric contraction. Since the orthosis allowed -5° dorsiflexion in this test condition, the decrease in soleus activity was justified by the physical limitations. A decrease in the soleus activity during approximately 35-55% stride (mid to late stance) in the test condition in which only -5° of dorsiflexion were allowed was found in all cases. Rectus femoris is used to check knee flexion, and a more plantarflexed (less dorsiflexed) orthosis has been shown to reduce knee flexion moments [3], so decreased rectus femoris activity may be expected. Vastus lateralis and medialis also function to control knee flexion, but neither showed a reduction in activity. It is unclear from this simple method of analysis why all the quadriceps muscles did not show similar changes. It is likely that mechanisms which individuals use to compensate for changing external conditions are involved. Tibialis anterior activity also showed a consistent reduction that may be explained by the reduction in motion permitted by the

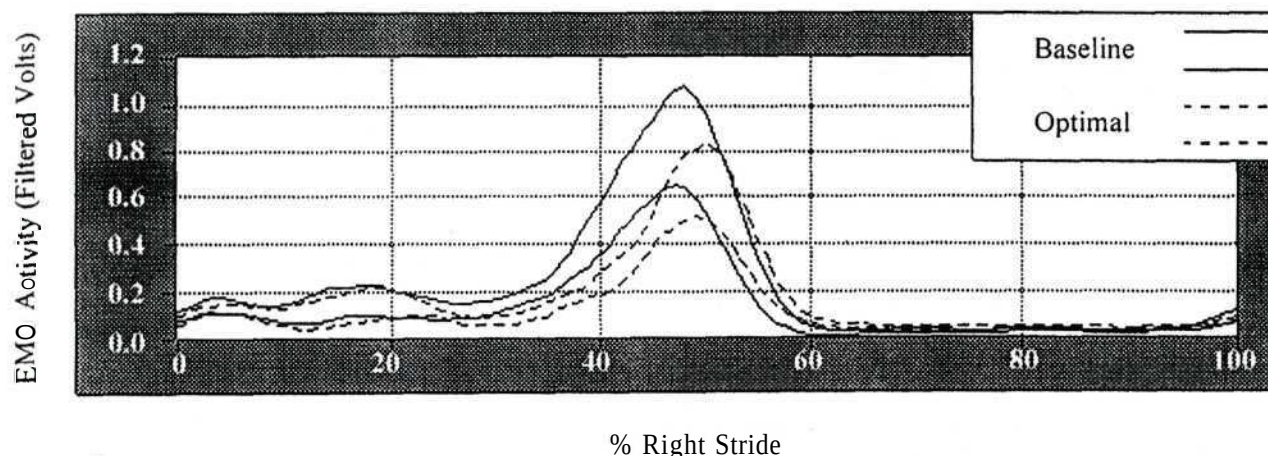


Figure 3. ± 1 Standard deviation bands of the ensemble averaged EMG signal from baseline and test condition.

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orthosis from heel strike to foot flat as tibialis anterior undergoes an eccentric contraction to control the plantar flexor moment.

To summarize, changes in muscle activities occurred in a consistent manner across all subjects in the normal population. This suggests that the compensatory mechanisms under these controlled conditions are relatively standard in normal individuals. In addition, the statistical processing and analysis methods selected have been validated. The next major step was to substantiate the methods and to study the compensation mechanisms in a population with neuromuscular deficiencies.

Polio survivors

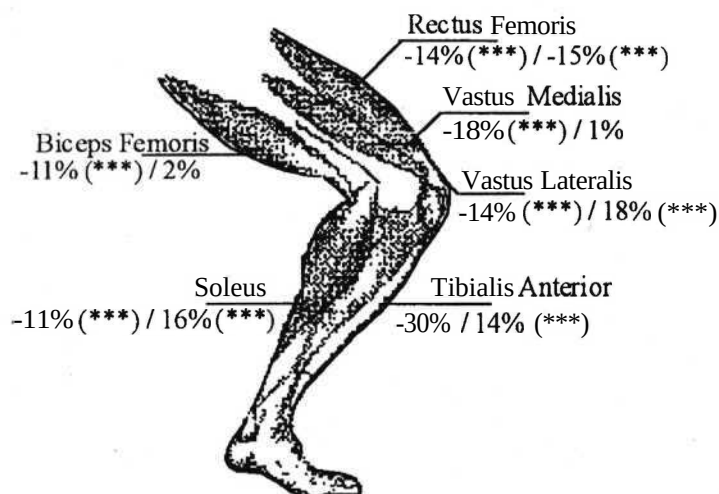
The core test methods were similar to those used with the normal subjects, but EMG data were interpreted slightly differently since the alignment conditions were different. Each subject's original orthotic alignment was used as baseline for the polio survivors. Each test condition provided a fixed angular change in range of motion with respect to those baselines. Although test condition C required adding 4° of plantarflexion to the baseline alignment for all subjects, the absolute range of motion for each subject in condition C was different since each began with a different range of motion. Thus no intersubject comparisons were made. Linear enveloped, integrated stride EMG signals from each of the 4 test conditions were statistically compared to stride EMG signals from the baseline condition (Condition A). One way Analysis of Variance established inequality of all means, and t-tests were used subsequently to compare activity values from each test condition to the baseline.

One of the most interesting findings was that individual responses to the test conditions were found to be time-dependent. The muscle activity response initially and after the accommodation period were not the same. A pictorial representation of changes in muscle activities is shown for one subject in Figure 4. The figure compares the changes

in muscle activities between the optimal and the baseline test conditions of one test subject. Below each muscle is a percentage that represents the change in muscle activity as gauged by integrated EMG signals. All the other test subjects showed similar responses in terms of magnitude and significance of the changes.

A range of motion setting with which an individual was unfamiliar resulted in an extreme response. Extreme indicates either a disproportional increase or decrease in muscle activity. After the accommodation period, the original (baseline) alignment could now be considered a "new" condition, since the subjects had been using the optimal condition for several weeks prior to this test session. In this case an extreme response was also evident. A possible interpretation of this trend is that subjects often show such extreme responses to any unfamiliar alignment condition. This is an important finding to be studied further since it affects the time frame for determination of suitable orthotic alignment.

Subject opinions of the optimal alignment condition immediately and after the accommodation period were recorded. A positive correlation between reduction in muscular effort required to walk in an orthotic alignment and the subject's perception of the alignment was expected, but this was not found. In fact, nearly the opposite was found. The most obvious trend seen in the responses shown in Figures 5 and 6 is that the subjects found unfavorable the alignments that were objectively found to be optimal. In comparison to the EMG data, as shown in figure 4, the subject opinions showed a tendency towards favorable when muscle use was increased with respect to the original (baseline) condition. The initial response to a new condition tended to be extreme. In other words, the initial response was either a move towards minimal or maximal muscle activity. In either case, the subjects may have compensated for changes in muscle activity by increasing overall energy expenditure. It is also possible these subjects had been used to co-contracting the knee stabilizing



Notes:

(1) Convention is: % Change before accommodation / % Change after accommodation

(2) A negative percent change indicates the optimal condition showed less activity than the baseline.

(3) Key to t-test significance:

* : $p < 0.05$ ** : $p < 0.01$

*** : $p < 0.005$

(4) No asterisk indicates the change was not significant ($p > 0.05$)

Figure 4. Percent change in EMG activity values and level of significance: immediately after administering new alignment / after accommodation period; subject: BG.

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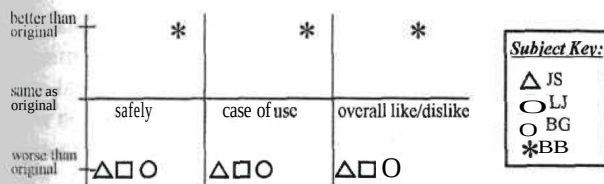


Figure 5. Polio survivor opinions of optimal alignment condition before accommodation period.

musculature to provide a feeling stance phase stability. These reasons may help to explain the largely unfavorable response obtained to the new optimal alignment when muscle demands were measured to be significantly reduced. This finding makes critical the need to measure energy expenditure and joint moments to verify the EMG findings as well as form a basis of clarification of the subject opinions.

EKG signals were measured to calculate heart rate and thus approximate energy expenditure. The heart rate plateau which must be reached to allow the correlation to energy expenditure was not reached due to the short duration of the test conditions. It was decided not to lengthen the test session in order to avoid fatigue from biasing the EMG recordings.

In the preliminary study with the normal population, a simple method was used to relate the change in amount of range of motion and change in muscle activity. Generally, it is likely that the more motion a body segment undergoes, the more muscle activity may be needed to control that motion. This correlation was established, in part, since the restrictions to the range of motion were severe. In the data obtained from the polio survivors, there appeared to be no definitive trends between amount of range of motion and muscle activity. Even the common assumption that increased plantarflexion reduces quadriceps muscle activity was not found to be true for this population. In addition, increased dorsiflexion did not result in more quadriceps activity. An increase in range of motion did not necessarily increase muscle activities; a decrease in range of motion did not necessarily decrease muscle activities, as shown in Figure 7. Since walking speed did not significantly change in these trials, muscle activity changes could not be due to that, however they may have been at the expense of overall energy expenditure or activation of other muscles used in walking.

Conclusions

Quantitative EMG comparison was found to be sensitive enough to indicate changes in muscle activity between the baseline (no motion restriction) and test condition as subjects walked. In a population free from any neuromuscular pathologies, the test condition was an ankle foot orthosis which provided a limitation to range of motion of the ankle. In this population, the changes in muscle activity were well defined and statistically significant. Muscle activity changes were also qualitatively correlated to the ankle range of motion restrictions due to the orthosis according

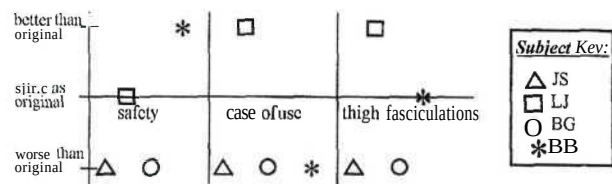


Figure 6. Polio survivor opinions of optimal alignment condition after accommodation period.

to a purposely over simplified rule: the less motion a joint is permitted to undergo, the less active the supporting musculature will need to be to stabilize that joint. The analysis of data from the normal population strengthened the validity of the EMG comparison method and served to establish a consistent pattern of muscular compensation. Determining methods and trends in neuromuscular compensation in a pathological population is an important goal.

Polio survivors who normally used an ankle foot orthosis to walk were chosen for the continuation of the study because they exhibit focal muscle weakness. The baseline condition was now each subject's original orthotic alignment, and thus was different for all subjects. A standard set (four test conditions) of minor changes to the range of motion were made to each subject's orthosis. It was important to determine (i) if the EMG comparison method was applicable for a population with neuromuscular weakness, as well as (ii) if the subjects' responses changed over time and (iii) if an objective function could be used to determine the goodness of the trial alignments. The objective function chosen was to minimize the activity of the quadriceps muscles, since they are often overworked in this population.

Comparisons of EMG data showed significant changes in muscle activity in response to the test conditions, and thus the EMG comparison method was useful for both populations. Quantitative changes in muscle activity were sometimes noted when visual analysis of ensemble averaged profiles seemed to indicate no differences. This was an important finding that deserves further study. Statistical

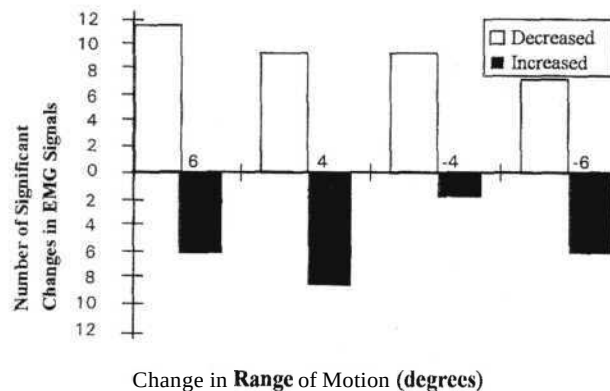


Figure 7. Breakdown of changes in muscle activities with respect to changes in range of motion. Data for this chart taken from polio subject testing (Test Session #1) data. A positive value for range of motion indicates plantarflexion, a negative value indicates dorsiflexion.

differences noted imply that subjective visual analysis of EMG may not allow easy perception of some relevant information such as minor differences in muscle activation patterns. It is difficult to gauge what statistically significant difference correlates to a clinically significant change. Further studies are underway to attempt to correlate measured EMG differences to joint moment and energy expenditure changes. An important aspect of clinical significance is how the subject perceives the change.

Subject assessment of the test alignment conditions was tabulated. The mathematically optimal condition was the one that followed the optimization algorithm and ideally reduced the quadriceps muscle activity the most. It was expected that a relief in quadriceps muscle activity would be a favorable condition. It was found that subjects tended to react in an extreme (disproportional increases or decreases in muscle activity) manner to an alignment that was unfamiliar, and so in many cases there was an alignment which substantially reduced quadriceps activity. However, subject opinions about these conditions were generally unfavorable. In fact, the conditions that increased muscle activity were often considered favorable. It is speculated that a feeling of stability due to cocontractions may contribute to the favorable rating assigned to these conditions. Another important finding was that the measured muscle activity response to the optimal alignment changed after a period of use. An altered response could have been due to actual physical or physiological adaptation, changes in the orthosis due to wear, or a combination of these effects. While it was found that the range of motion of some of the orthoses did change slightly after use, it is unlikely that the magnitude of muscle activity changes measured could be due entirely to this; especially given the nature of the muscle use trends noted. The subject opinion to the optimal alignment was generally consistent over time, and so it more closely matched the final muscular response. This implied that the subject may have been aware of the long-term or steady state response to an alignment condition immediately after trying it. Another theory to explain the inverse correlation between muscle activity and subject opinion is subjects may have increased overall energy expenditure while reducing muscle use. This would contribute to their generally unfavorable opinion of the optimal condition. EKG measurements were insufficient to assess this theory since the duration of walking in each test condition was shorter than that required for heart rate to plateau.

In the normal population, severe range of motion restrictions produced responses that could be estimated by simple biomechanical rules. In the polio survivor population, test conditions only differed by 2 to 4 degrees. While the measured muscle responses seemed disproportionately large for the size of the motion restriction that caused them, the muscle responses did not always occur in a manner consistent with simple biomechanical rules used earlier. An example of such a rule is that more ankle plantarflexion normally produces less quadriceps activation because it forces the knee into relatively more extension during

stance phase of gait. This suggests that the biomechanical mechanisms that are commonly used to estimate muscle demands may be more complex than conventional mechanical heuristics suggest, especially in a population with neuromuscular weakness. Fine-tuning the range of motion aspect of orthotic alignment, as studied in this work, may have a disproportionately large impact on efficiency of resultant gait.

Future work includes continued testing on polio survivors, as well as the measurement of joint moment and energy expenditure data to strengthen the assessment of change in muscle activity. In addition, a model that may allow prediction of changes in muscle responses to certain conditions like those used in this study and which incorporates compensation effects used in response to such conditions is being developed.

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