Crosstalk: Surface Versus Intramuscular Electrodes for the Peroneus Brevis and Peroneus Longus

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CROSSTALK: SURFACE VERSUS INTRAMUSCULAR ELECTRODES FOR THE PERONEUS BREVIS AND PERONEUS LONGUS

By

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THESIS

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CROSSTALK: SURFACE VERSUS INTRAMUSCULAR ELECTRODE S FOR ELECTROMYOGRAPHY OF THE PERONEUS BREVIS AND PERONEUS LONGUS

ABSTRACT

The purpose of this descriptive study was to measure crosstalk when using surface electrodes over the peroneus brevis and peroneus longus muscles. EMG was recorded from the right lower extremity of 14 normal subjects, taken from a sample of convenience. Surface electrodes were applied over the peroneus brevis and peroneus longus muscles; fine wire electrodes were inserted into the peroneus longus, peroneus brevis, anterior tibialis and extensor digitorum longus muscles of each subject. During manual muscle tests for the peroneals, anterior tibialis, and extensor digitorum longus, data were recorded from all electrodes simultaneously over 5 trials of each. Pearson product-moment correlation and stepwise multiple regression were used to analyze the processed, normalized EMG data. Crosstalk was found in data collected by surface electrodes over the peroneus longus and brevis. Crosstalk was more evident from neighboring muscles when the peroneals contracting at submaximal levels. Furthermore, crosstalk was more evident from synergistic muscles than from antagonists when the targeted muscle was working near its maximum. Cocontraction of antagonistic muscles was found during maximal voluntary isometric muscle contractions. The results of this study suggest that when specific information regarding timing, onset, and/or duration of peroneus longus or peroneus brevis activity is desired, surface electrodes are likely to include unacceptable amounts of inaccurate data.
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Key Terms

**AMPLITUDE:** That which expresses the level of muscle signal activity (Basmajian & Deluca, 1985).

**ATTENUATION:** Lowering of EMG frequency as it passes through surrounding tissues.

**CEREBRAL PALSY:** Neurodevelopmental impairment caused by a nonprogressive defect or lesion in single or multiple locations in the immature brain. The lesion can occur in utero or shortly after birth (Olney & Wright, 1995).

**COCONTRACTION:** Simultaneous activation of either synergistic or antagonistic muscles about a joint.

**CONTINUOUS:** A type of EMG timing error which is used to describe muscle activity that is uninterrupted for 90% or more of the gait cycle (Perry, 1992).

**CROSSTALK:** EMG signal picked up by an electrode that is generated from muscles other than the one of interest (Basmajian & DeLuca, 1985).

**DYNAMIC EMG:** Recording of muscle activation signals during functional activities.

**FORCE:** Any influence that causes a change in joint position or alters the direction or speed of motion.

**IMPEDANCE:** Opposition to flow of alternating electrical currents measured in ohm’s (LeVau & Anderson, 1992).

**KINEMATIC:** The magnitude and timing of individual joint action (Perry, 1992).

**KINETIC:** The force that causes movement (Winter, 1990).

**MAXIMUM VOLUNTARY ISOMETRIC CONTRACTION (MVIC):** Voluntary contraction of a muscle or muscle group in which the subject attempts to achieve a maximum force (Winter, 1990).

**OUT OF PHASE:** An EMG timing error that is used to describe muscle activity in swing or stance phase of gait which is reversed from the expected norm (Perry, 1992).
**RAW EMG SIGNAL:** Unprocessed EMG signal that is the basis of all methods of interpreting the myoelectric activity from muscles (LeVeau & Anderson, 1992).

**SHAPE:** Characteristics of an EMG signal which remain unaltered with linear scaling in either amplitude or time domains (Basmajian & Deluca, 1985).

**SPASTICITY:** Velocity dependent hypertonicity in a muscle or muscles accompanied by exaggerated tendon reflexes (Olney & Wright, 1995).

**VARIANCE RATIO:** A statistical tool derived from EMG ensemble average which measures the repeatability of a waveform.

**WAVEFORM:** Potential, voltage, or current associated with an EMG signal as a function of time. It incorporates shape, amplitude and time duration (Basmajian & Deluca, 1985).
CHAPTER 1
INTRODUCTION

Background To The Problem

Acquiring or improving the ability to walk is often one of the main goals of patients with neurologic disorders. In order to assist patients toward this goal, medical professionals study both normal and abnormal lower extremity movement patterns of the gait cycle. The gait cycle is defined as the interval from the point that one lower extremity makes ground contact to the moment when the same extremity contacts the ground again (Norkin & Levange, 1992). It can be divided into two major phases: stance and swing. Stance phase is the period when the reference or weightbearing limb is in contact with the ground. Swing phase is the period when the reference limb is not in contact with the ground.

Dynamic electromyography (EMG) has been used to identify the period and relative intensity of muscle activity during gait (Perry, 1992). EMG activity patterns have been established in normal individuals and have been compared to pathologic patterns during gait. Dynamic EMG has been useful in assisting with decision making regarding procedures to correct or improve abnormal gait patterns by providing objective information about muscle function (Brashear &
Raney, 1986; Perry & Hoffer, 1977; Thompson & Hoffer, 1991). In particular, dynamic EMG has been used (along with kinematic and kinetic data) in the planning and evaluation of surgical procedures designed to improve gait in children with cerebral palsy (CP). Because muscle activity of children with CP often does not follow normal patterns for onset, cessation and intensity during gait (Olney & Wright, 1995), dynamic EMG can be a key assessment tool for individualizing treatment.

CP is a nonprogressive neurological disorder resulting from damage to the immature brain either prenatally, during the birth process, or during the first three years of life (Olney & Wright, 1995). CP is often associated with muscle spasticity and loss of selectivity in muscle function. Spasticity is velocity dependent hypertonicity in a muscle or muscles accompanied by exaggerated tendon reflexes (Olney & Wright, 1995). Muscle selectivity is defined as “voluntary control that allows individual muscle activation for the appropriate duration and intensity that is functionally required” (Miller & VanderKooi, 1996 p. ix). Problems with spasticity and selectivity in the lower extremities result in muscle imbalances which contribute to musculoskeletal deformities and abnormal gait.

Surgical interventions have been used to correct lower extremity musculoskeletal deformities and improve gait patterns in patients with CP. Two common procedures include tendon lengthening and tendon transfer. A variety of methods have been employed to increase the length of a tendon, all of which
have the effect of reducing the force the musculotendinous unit can exert on the joint(s) it crosses. Tendon transfer involves surgically moving the tendinous insertion of a muscle to another site for the purpose of changing the biomechanical line of pull of the muscle in order to help correct abnormal forces on a bone or across a joint and improve function. In some cases the tendon is split and only one portion is transferred in order to balance its effect on a joint. The altered line of pull of a transferred muscle also changes the joint motion produced by contraction of that muscle. For example, in the case of a nonfunctional or paralyzed tibialis anterior muscle, which results in footdrop during the swing phase, the peroneus brevis tendon may be transferred to the dorsum of the foot (Brashear & Raney, 1986). Normally, the peroneus brevis acts as a plantarflexor and evertor of the ankle/foot complex because its tendon lies posterior to the axis of motion of the talocrural joint and its insertion is lateral to the axis of the subtalar joint, respectively. The surgery moves the tendon’s line of pull anterior to the ankle joint center, making the muscle a dorsiflexor, thus improving ankle/foot position during swing phase.

Research has shown that once transferred, a muscle rarely changes its phase of activity during the gait cycle (Gritzka, Staheli, & Duncan, 1972). This means that although a transferred muscle will produce a different motion after surgery, it will be active at the same time during the gait cycle as it was prior to surgery. Therefore, when considering tendon surgery, determination of the onset, cessation, intensity and phase of muscle activity during the gait cycle is
invaluable and can be used both to help determine the best surgical choice and predict the probable outcome (Perry & Hoffer, 1977; Thompson & Hoffer, 1991). However, in order to make optimal surgical decisions based on EMG analysis, it is imperative that activity of the targeted muscle be reflected accurately. The method of EMG data collection may affect that accuracy.

EMG data can be collected through the use of either surface or fine wire (intramuscular) electrodes. Surface electrodes are applied to the skin over the target muscle. Fine wire electrodes are inserted directly into the belly of the target muscle. There are advantages and disadvantages of both types of electrodes. Surface electrodes, as compared to intramuscular electrodes, are noninvasive, easier to apply, and are not painful. However, EMG recordings from surface electrodes will more likely (than intramuscular electrodes) include artifact signals from sources other than the target muscle (Basmajian & DeLuca, 1985; Perry, 1992). Possible sources of artifact include power lines, electrical equipment, and neighboring muscles. EMG signals originating from muscles other than the one being targeted is termed crosstalk (Morrenhof & Abbink, 1985). Crosstalk can skew the accuracy of the analysis of onset and cessation times, as well as the intensity of muscle activation. Thus, crosstalk results in a reduced ability of the electrode to selectively record EMG signals from the target muscle alone. Previous studies of the posterior calf muscles have demonstrated that surface electrodes over the gastrocnemius muscle can pick up crosstalk from the soleus and posterior tibialis muscle (Perry, Easterday, & Antonelli, 1981; Toft,
Sinkjaer, Anderson, & Larsen, 1991). Crosstalk in the peroneal muscles has not been thoroughly investigated. The peroneal muscles are located in the lateral lower leg and include the peroneus longus, peroneus brevis, and peroneus tertious. The peroneus tertious is a very small muscle whose function is quite different from that of the other two peroneal muscles and, therefore, will not be considered in this study. The peroneus brevis and longus function to evert and plantarflex the ankle/foot complex. The peroneus brevis also contributes to dynamic lateral stability of the ankle (Kendall, McCreary, & Provance, 1993) while the peroneus longus contributes to the dynamic preservation of the longitudinal arches of the foot (Norkin & Levangie, 1992). During the stance phase of gait, both are normally active only during mid and terminal stance (Perry, 1992). When normal neurologic control is impaired by CP, overactivation of one or both of these peroneal muscles and/or weak activation of the opposing invertor muscles can occur. Either of these conditions can result in unbalanced eversion forces across the subtalar joint, which may lead to a dynamic rearfoot valgus deformity during gait. Possible corrections for this problem include transfer or lengthening of one or both of the peroneal tendons (Perry & Hoffer, 1977; M. Forness, personal communication, March 11, 1997). The literature has not clearly defined the most appropriate choice of electrodes (surface versus intramuscular) when performing EMG analysis of peroneal muscle activity. It is generally desirable to choose surface over intramuscular electrodes due to their ease of application and decreased risks to the patient. However, it is important
that this choice does not significantly affect the accuracy of the EMG recordings. Therefore, the purpose of this study was to measure crosstalk when using surface electrodes over the peroneus brevis and peroneus longus muscles.
CHAPTER 2
LITERATURE REVIEW

Electromyography

Role of Electromyography in Normal Gait Analysis

Electromyography (EMG) is the method of recording electric signals generated by muscle activity. Dynamic EMG involves recording EMG during a functional activity such as walking. Sutherland (1978) stated that electromyography was one of the most important tools that a gait laboratory used. Dynamic EMG is used to define the timing and intensity of muscle activity relative to the gait cycle (Perry & Antonelli, 1993). The use of dynamic EMG in gait has been well documented, with normative data established in adults and children (Perry, 1992; Shiavi, 1987; Winter & Yack, 1987; Winter, 1990). Normative EMG patterns have been useful in interpreting pathologic gait and in elucidating the need for and effectiveness of treatment for patients with gait pathologies (Knutson & Soderberg, 1995).

EMG’s Role in Pathological Gait

As described by Perry (1992), an abnormal EMG record during walking can be due to the following: errors in neurological control, muscle weakness, compensatory strategies, and obligatory postures. Any of these conditions can result in alterations in the normal timing and intensity of muscle activity during
a particular phase or throughout the entire gait cycle. In patients with CP, the relative timing of muscle activity has functional significance. Muscle overactivity may represent an obstruction of a desired function or compensation for an abnormal joint posture. Compensatory strategies and postural adaptations due to inappropriate neuromuscular control could also account for decreased muscle activity, described as delayed, curtailed, or absent. According to Perry, muscle activity that is continuous throughout the gait cycle is always undesirable. Abnormal muscular intensity can also be described as excessive, inadequate, or absent.

To accurately identify whether muscle responses are appropriate for the period of time analyzed, EMG can be synchronized with kinematic and kinetic data taken during computerized gait analysis. Kinematic data provide an objective measurement of the dynamic joint range of motion. Kinetic analysis provides information on the moments and power at each joint.

Attributes of EMG

The electromyogram provides an indirect measure of muscle function. The central nervous system initiates a muscle contraction by depolarization of an alpha motor neuron which is directed along the motor neuron to the muscle fibers' motor end plate. There it triggers an exchange of ions across the muscle membrane (Basmajian & DeLuca, 1985; Perry, 1992). Once this exchange reaches threshold level, the muscle cell depolarizes. This depolarization results in an action potential which is propagated in both directions along the sarcolemma of
the muscle fiber (Lamb & Hobart, 1992; Perry, 1992). It is the action potential which is recorded as the electromyogram.

Muscle is composed of groups of muscle fibers or cells called motor units (MU)s. A MU, the smallest muscular unit, is composed of a single alpha motor neuron, its neuromuscular junction, and all the muscle fibers it innervates. A motor unit originates in the spinal cord as an alpha motor neuron. The alpha motor neuron’s axon exits the spinal cord within a peripheral nerve. The axon enters a muscle and branches to innervate a few to several hundred muscle fibers. Motor unit size is relative to the muscle’s function and size. MUs in muscles used for fine or postural control, innervate a small number of muscle fibers. For example, lumbricals in the hand contain 95 MUs with 100 fibers in each MU. Conversely, MUs in muscles used for gross and phasic movements, usually innervate a large numbers of muscle fibers. For example, the lateral head of gastrocnemius contains 579 MUs with up to 1784 fibers in each MU (Perry, 1992).

Characteristics of Muscle and Influence on EMG

Electrical activity of a muscle is measured by electrodes placed on the surface of the skin or within the muscle belly. An EMG waveform’s shape and amplitude represent the MUs’ size, action potentials’ propagating velocity, electrode contact area or interwire spacing, and electrode distance from the MUs (Winter, 1991). The amplitude of EMG recordings are influenced by several factors: concentration of collagen between muscle fibers, amount of adipose
tissue, density of fast and slow twitch muscle fibers, force and type (isometric versus dynamic, concentric versus eccentric) of muscular contraction, muscle length, and joint angle (Basmajian & DeLuca, 1985; Perry et al., 1981; Perry, 1992).

The anisotropic design of muscle and related connective tissue and electrode placement can affect EMG amplitude. Muscles contain varying densities of collagenous sheaths between their fibers. A decreased level of EMG amplitude can be found in areas containing high concentrations of collagenous sheaths and fewer muscle fibers (Perry et al., 1981). Since muscle is covered by fascia, adipose tissue and skin, it is difficult to know if electrodes are placed over or within an area of high or low collagen concentration. Amount of adipose tissue can influence EMG amplitude. When using surface electrodes adipose tissue can act as a high-impedance barrier lowering the amplitude of the EMG signal read on the surface (Loeb & Gans, 1986).

EMG amplitude is also influenced by location and concentration of fast and slow twitch muscle fibers and their distance from the detection electrode. In general, fast twitch fibers are larger in diameter, closer to the surface and demonstrate greater amplitude action potentials than slow twitch. Conversely, slow twitch fibers are generally deeper and have been found to generate lower amplitude signals (Hillstrom & Trilo, 1995). Therefore, an EMG electrode on the surface may give a very different amplitude reading than one placed within the muscle belly.
EMG amplitude is an indirect measure of muscle force (Basmajian & DeLuca, 1985; LeVeau & Andersson, 1992; Perry, 1992). Under isometric conditions EMG amplitude has been shown to be linearly proportional to muscle force or torque (Basmajian & DeLuca, 1985; Perry, 1992). With dynamic contractions the relationship between EMG amplitude and force was shown to be “nonlinear and nonmonotonic” (Basmajian & DeLuca, 1985 p. 199).

The length-tension properties of muscle seem to have variable effects on EMG amplitude. Inman, Ralston, Saunders, Feinstein and Wright (1952) showed that length and tension are inversely proportional to EMG amplitude during isometric contractions. Muscles in a lengthened position displayed increased tension and low EMG amplitude, while muscles in a shortened position displayed decreased tension and high EMG amplitude. With dynamic contractions EMG amplitude seems to remain constant despite decreasing tension during concentric contractions and increasing tension during eccentric contractions. Therefore, during concentric and eccentric contractions, EMG represents the contractile state of the muscle regardless of the tension within the tendon (Winter, 1990).

Joint angle and the corresponding length of the muscle may also have an impact on EMG amplitude (Giroux & Lamontagne, 1990; Mirka, 1991; Zakaria, Kramer, & Harburn, 1996). Mirka and Zakaria et al. used one isometric maximum voluntary contraction (MVIC) and its corresponding joint angle as the reference, and compared its EMG to the EMG of MVIC’s at other joint angles.
They found variability in the EMG amplitudes among the various joint positions. Mirka and Zakaria et al. concluded that it was inaccurate to relate the EMG of a MVIC at one joint angle to the EMG of MVIC's at other joint angles, because with each joint angle different portions of the muscle were within the readable range of the electrodes. Moreover, changes in the amount of tension a muscle was able to generate at different joint angles may also have influenced the EMG amplitude (Mirka, 1991; Zakaria et al., 1996).

Electrode Configurations

Application of electrodes or electrode configurations have been described by Basmajian and DeLuca (1985). Electrodes can be arranged in a monopolar or bipolar configuration. In a monopolar setup the active electrode is placed at the detection surface and a reference electrode is placed in an area unrelated to the one of interest. The myoelectric signal is taken as the difference between the electrical potential at the detection surface and the reference electrode. Monopolar arrangements have the tendency to pick up signals from sources other than the muscle being investigated.

In a bipolar arrangement two detection electrodes measure electrical potentials. Signals common to both electrodes representing noise or artifact are eliminated. Those signals which are different (potential difference or voltage) at each electrode are taken to be the true EMG reading. The difference between the two signals is measured, amplified and then recorded. A ground electrode is
also needed in this arrangement. The bipolar arrangement is preferred because it decreases the amount of noise.

**Signal Artifact**

Noise or artifacts are any unwanted signals that are detected along with electrical recording of muscle activity (Basmajian & DeLuca, 1985). There are several sources of artifact generation. Mechanically induced artifacts are due to mishandling and or movement of cables during dynamic activities. Other artifacts are created by movement between the skin and electrode (Basmajian & DeLuca, 1985). Additional sources of noise include power lines, electrical devices, radio signals, and television signals. Artifact can be reduced by minimizing electrode impedance and using high quality instrumentation which can filter out unwanted signals (Soderberg, 1992).

**Surface Electrodes**

Surface electrodes have long been used as a reliable measure of muscle activity (Perry, 1992; Basmajian & DeLuca, 1985). They are inexpensive, non-invasive and easily used in the clinical environment. Electrodes generally are four to eight millimeters in diameter with a silver-silver chloride surface (Jacobson, Gabel, & Brand, 1995; Koh & Grabiner, 1993; Winter & Yack, 1987). In the bipolar technique, electrodes are placed parallel to the muscle fibers over the midline of the muscle belly or over a motor point, spaced 12-20 mm center-to-center (Perry, 1992; Soderberg, 1992). Parallel arrangement of surface electrodes is chosen because the anisotropic property of muscle tissue where impedance is
in a perpendicular direction is up to ten times greater than that in a longitudinal
direction (Basmajian & Deluca, 1985).

There are potential problems when using surface electrodes. For example,
larger electrodes with increased center-to-center distance produce elevated EMG
readings; however, increased intra-electrode distance has been shown to result in
greater contamination from adjacent muscles, i.e. crosstalk (Basmajian & DeLuca,
1985; Perry et al., 1981; Perry, 1992). Another potential problem with the use of
surface electrodes is impedance. Muscle activity is measured through the layers
of connective tissue, adipose, and skin when surface electrodes are used. These
tissues act as an internal impedance to the propagation of electrical currents by
attenuating higher frequency waves and progressively lowering the frequency of
the EMG reading (Basmajian & DeLuca, 1985; Perry, 1992). Impedance can be
reduced by shaving, abrading, and swabbing the skin with alcohol prior to
electrode application (Basmajian & DeLuca, 1985; Winter, 1990; Perry, 1992).

The specific muscle group targeted when using surface electrodes may
also influence the quality of EMG signal. Surface electrodes are most effective
when they are applied over large, superficial muscle groups (Basmajian &
DeLuca, 1985; DeLuca & Merletti, 1988; Soderberg, 1992). Surface electrodes are
more likely to pick up information from surrounding muscles (crosstalk), when
used over deep and/or small muscles (Basmajian & DeLuca, 1985; DeLuca &
While surface electrodes demonstrate decreased specificity, they have been shown to be a more reliable measure of muscle activity compared to fine wire electrodes (Winter, 1990). Kadaba, Wootten, Gainey and Cochran (1985) found that surface electrodes demonstrated a lower variance ratio than fine wire for both day-to-day and intersubject reliability.

While surface electrodes have been shown to display a more reliable representation of an individual's muscle timing pattern, it has been assumed that normal movement patterns are not altered by the surface electrodes worn by the subjects. Young, Rose, Biden, Wyatt and Sutherland (1989) found that surface EMG electrodes attached to children with cerebral palsy altered their gait. Results of their study showed that surface electrodes placed on lower extremity muscles decreased the cadence and walking velocity of their subjects. Young et al. concluded that this decrease was due to the EMG cable attached to the child. However, Jacobson et al. (1995) and Winchester, Carollo and Wrobbel (1996) found that surface electrodes did not significantly alter gait patterns in normal adults.

Fine Wire Electrodes

Fine wire electrodes have been widely accepted for use in EMG analysis. These electrodes measure activity directly from the muscle without intervening adipose and skin. Fine wire electrodes are most appropriate when a specific action of a muscle is desired, the muscle under investigation is deep, or the action of adjacent muscles is to be compared (Perry, 1992). Fine wire electrodes
are usually made of non-oxidizing metals such as platinum alloys, silver and nickel-chromium, and are insulated with nylon, polyurethane, or Teflon (Basmajian & DeLuca, 1985). Diameters of wire electrodes can vary between 25 and 50 microns. Like surface electrodes, fine wire electrodes are usually arranged in a bipolar configuration when used to record muscle activity. Insertion of the wires into a target muscle is accomplished with the use of a 25-27 gauge hypodermic needle. Once the wires are inserted into the muscle belly, the hypodermic needle is removed. With the aid of an oscilloscope, placement accuracy can be confirmed by manual muscle testing and/or electrical stimulation (Perry, 1992). If wire insertion is consecutively followed by passive range of motion and a few vigorous contractions, displacement of the electrodes are seldom a problem. Basmajian and DeLuca (1985) recommended six muscle contractions to assure electrode setting within the muscle. Two potential problems associated with intramuscular electrodes include shearing of the wire and shorting out (due to the two wires coming in contact) (Soderberg, 1992).

Because the fine wire technique is invasive, fine wire electrodes are not as commonly used as surface electrodes. Discomfort has been reported during wire insertion. Although subjects describe minimal pain after insertion, muscle cramping may occur (Perry, 1992; Soderberg, 1992). Other reasons for limiting the use of fine wire electrodes include the risk of infection, the possibility of the wire breaking off inside the muscle, and bruising. These associated risks with fine wire electrodes are minimal and have not been shown to influence testing.
procedures in normal adults. For example, even though fine wire insertion discomfort has been identified, studies have shown that fine wire electrodes do not result in an antalgic gait pattern in normal adults (Jacobson et al., 1995; Winchester et al., 1996).

When fine wire electrodes were used on children with CP, temporary changes in the gait pattern have been reported. Young et al. (1989) used high speed cinematography to measure gait parameters in children with CP. They compared three situations: no electrodes, surface electrodes and fine wire electrodes. Changes in gait were apparent when both fine wire and surface electrodes were used. Young et al. speculated that the gait changes associated with fine wire electrodes could have been due to pain, anxiety, discomfort and fatigue since the fine wire electrodes were always used last. However, Jacobson et al. (1995) speculated that the results could have been due to Young’s method of measuring gait parameters.

Crosstalk

Crosstalk is defined as EMG signal recorded by an electrode that is generated from muscles other than the one of interest (Basmajian & DeLuca, 1985). Crosstalk occurs because of the anisotropic nature, or nonhomogeneity, of muscles. Action potentials create currents which can travel in many directions beyond the boundaries of the muscle being activated. These signals may be detected at electrodes placed some distance from the source (DeLuca & Merletti,
Crosstalk could, therefore, pollute the EMG signal and lead to misinterpretation.

Research has shown a greater incidence of crosstalk with surface, than with fine wire electrodes (Basmajian & DeLuca, 1985; Perry, 1992). In one investigation, only surface electrodes were placed over the peroneus brevis, soleus, and tibialis anterior muscles. The tibialis anterior was electrically stimulated generating a supramaximal contraction. When the tibialis anterior contracted, activity was recorded by the electrodes over peroneus brevis and soleus (DeLuca & Merletti, 1988). Koh and Grabiner (1993) undertook a similar study. They placed surface electrodes on the medial and lateral hamstrings and electrically stimulated the quadriceps femoris via the femoral nerve. Results showed that average crosstalk measurements were 17% of maximal voluntary effort (MVE) for the lateral hamstring, and 11% MVE for the medial hamstring when compared to quadriceps MVE. Solomonow (1994) cited the limitations of both studies stating that the authors failed to confirm their crosstalk measurements with fine wire electrodes. Furthermore, Solomonow speculated that the relationship between cutaneous receptors and mixed sensory-motor nerves resulted in overflow stimulation of muscles not directly stimulated, thus mimicking crosstalk.

Perry et al. (1981) used surface and fine wire electrodes to study crosstalk from the gastrocnemius, soleus, and tibialis posterior muscles. EMG was recorded while each muscle was isolated by a manual muscle test. Normalized
EMG values for the two types of electrodes for each muscle did not correlate, so cross talk was suspected. Perry used a mathematical model to estimate the contribution of each muscle to the EMG reading of the surface electrode. EMG readings from the surface electrode over the soleus were determined to be the sum of 36% soleus, 31% gastrocnemius, 22% tibialis posterior and 11% from other muscles, as measured by fine wire electrodes. Perry concluded that "no surface electrode can relate to a single muscle if others are in close proximity" (p. 14). Toft et al. (1991) performed a similar study using surface and intramuscular electrodes in the gastrocnemius and soleus muscles. By comparing the surface and needle electrode data and using a mathematical model, they also found significant crosstalk between the two muscles.

Winter (1990) has argued that fine wire electrodes are more susceptible to picking up crosstalk because of their relative proximity to deeper muscles. Winter stated that signals from a radius of approximately two centimeters could be detected by a fine wire electrode in the gastrocneumius. Therefore, Winter speculated that if the wire was placed deep into the gastronemius, it was more likely to pick up activity from the soleus and tibialis posterior, which lies beneath or adjacent to the gastrocnemius.

Human and animal studies have shown that the amount of adipose tissue between the muscle and skin can contribute to crosstalk in surface EMG (Solomonow, Baratta, Zhou & D' Ambrosia, 1988; Solomonow et al., 1994). Solomonow et al. (1994) measured the EMG activity of calf musculature in cats.
with surface and fine wire electrodes. Crosstalk was found to be a negligible 5% in surface and 2.67% in wire of the mean absolute value MAV (mean value of the fullwave rectified M wave in a 25 ms window). However, when excess adipose tissue was present, crosstalk percentages in both fine wire and surface had changed. Crosstalk detected by fine wire electrodes had either stayed the same or had risen to a maximum of 3.77% of MAV, while crosstalk at the surface electrodes rose to 20% of MAV. The authors speculated that this increase was strictly due to adipose tissue. This study supported the hypothesis that fine wire electrodes were less exposed to signals from distant sources and more sensitive to electrical activity close to the electrode.

Methods have been devised to reduce crosstalk when using surface electrodes. The double differential (DD) and branched electrode (BE) are two such methods. The reduction in crosstalk is measured by comparing the DD or BE signal to the standard bipolar signal or single differential (SD) signal. In the DD method three electrodes are placed linearly on the skin over a muscle. After passing through two differential amplifiers, the three electrodes create two bipolar signals called single differential (SD) signals. These signals travel through a third differential amplifier which result in the DD signal. The DD method rejects signals with long wavelengths through a high pass filter. It is believed that long wavelength electrical signals are a major contributor to crosstalk (Koh & Grabnier, 1993). The BE method involves the following: connecting three electrodes in series, connecting the middle electrode to one
terminal of a differential amplifier; and shorting the collateral electrodes at the other terminal. Branched electrode, like DD, removes signals with long wavelengths (Koh & Grabnir, 1993). When compared to the SD signal, DD and BE signals were found to reduce crosstalk to the same extent (Koh & Grabner, 1993). DeLuca and Merletti (1993) also compared DD to SD and found that crosstalk was significantly less with the DD technique. DeLuca and Merletti stated that the DD technique picked up a smaller detection volume and therefore was less susceptible to crosstalk than the SD method.

In summary, EMG crosstalk has been shown to exist and skew data, suggesting that these data should be interpreted with caution. Most investigators agree that fine wire electrodes are less susceptible to crosstalk than surface electrodes. Therefore, fine wire electrodes should be the method of choice when small or deep muscles are being studied.

**Recording EMG**

Whether using surface or indwelling electrodes, a biological amplifier with specific parameters is needed to record EMG signal with minimal distortion. Undistorted signal refers to linear signal amplification over the range of the amplifier recording system. The amplifier has a dynamic range; this range should include the highest amplitude EMG signal recorded to prevent high amplitude EMG signals from being clipped off. Amplifier gain is the degree to which the signal is amplified and represents the ratio of the input voltage to the output voltage. Amplifier gain should not exceed the range of input signals, or
else peaks of EMG may be cut off. For example, a 2-mV input with a gain of 1000 will have an output of 2 V (Winter, 1990).

There are three additional considerations to be made when specifying the EMG: input impedance, frequency bandwidth and common-mode rejection ratio (CMRR) (Winter, 1990). Input impedance of a biological amplifier should be 100 times greater than the source impedance (electrode-skin) so that the signal is not overly attenuated. The surface electrode-skin interface has a finite impedance which depends on several factors including thickness of the subcutaneous and skin layers, cleanliness of the skin and area of the electrode surface. Fine wire electrodes have a higher impedance due to the small surface area of bare wire in contact with the muscle tissue (Winter, 1990).

The frequency bandwidth is the range of frequencies that the amplifier accepts. It should be set so all frequencies present in the EMG are amplified without attenuation. Cutoff frequencies allow for some of the noise to be removed from the signal. The low cutoff frequency is usually around 15 Hz, which is aimed at removing noise from electrode-skin movements. The high cutoff frequency is usually between 1,000 and 2,000 Hz when using fine wire electrodes and 500 to 800 Hz when using surface electrodes (Winter, 1990).

Finally, the common-mode rejection ratio (CMRR) removes signals with equal amplitudes that appear at both detection electrodes, thus reducing distortion of the EMG signal. The body is a good conductor and acts as an antenna by picking up identical electromagnetic signals at both electrodes. These
common signals called "hum", come from power lines, fluorescent lights, power cords, and electronic machinery (Winter, 1990).

**Signal Processing**

After amplification, the EMG signal is considered raw data. Some researchers report their data in the form of raw EMG (Perry, 1986; Winter, 1994). Winter (1991) states that it is difficult to interpret the amplitude and shape of these raw signals. Interpretation of the raw signal is usually accomplished by visual inspection. With training and experience the examiner is able to determine onset and cessation of muscle activity. However, there are problems with this method of analyzing EMG data. Different threshold levels for onset and cessation result in varying and potentially misleading onset and cessation patterns (LeVeau & Andersson, 1992). Inman et al. (1952) found that the complexity of the raw data led to inaccurate comparisons of peak-to-peak amplitude.

Unlike raw data, properly processed EMG data can be compared and correlated with other physiological or biomechanical signals (Winter, 1990). There are several methods and combinations of methods for processing the raw EMG signal. These methods include full wave rectification, linear envelope, integration, and root mean square processing. Researchers do not agree on the exclusiveness of any one method; therefore, they all deserve ample discussion.

Full-wave rectification is the first step in processing the raw EMG data (Winter, 1990). Rectification converts the signal to a single polarity, by inverting
all the negative signals to positive. A rectified signal gives a good indication of
the changing contraction level of the muscle, and thus can be used to determine
onset and cessation of muscle activity. It also serves as an input to other
processing schemes.

Filtering a full-wave rectified signal through a low-pass (3-6 Hz) critically
damped filter, creates a linear envelope (LE). LE has also been described as a
moving average because it displays a pattern that reflects changes in the state of
muscle contraction over a given time period (LeVeau & Andersson, 1992).

Another method of processing the full wave rectified signal is integration.
Integration involves summing the rectified EMG over a certain time interval. It
is used to determine the total amount of muscle activity at any given time
interval. For gait studies, Perry (1992) recommended summing the signal every
ten milliseconds. Integrating over a specified time is similar to a moving
average.

Root Mean Square (RMS) is a method of processing using raw data that is
also similar to linear envelope because it gives a moving average over time. The
RMS measures the electrical power, and thus provides an indication of the
magnitude of muscle force from the signal (LeVeau & Andersson, 1992). Since
raw EMG time varying signals are squared with RMS processing, full-wave
rectification is not required (LeVeau & Andersson, 1992).

In conjunction with rectification and LE, ensemble averaging (EA) is a
method that can be used to summarize multiple strides for one person, or
summarize group data. Using gait as an example, the raw data are fullwave rectified and a linear envelope is created for each stride. The linear envelopes (for a given number of strides) are passed through a curve fitting program which sets each stride to 100%. Next, the linear envelopes are summed and averaged to create EA, which has the effect of neutralizing variability between subjects (Winter, 1991).

When comparing two or more subjects, muscles, or activities, the raw or processed EMG data must be normalized to accommodate for individual variation in: muscle bulk, overlying fat, intersubject postural differences, variable neural control strategies, minor differences in electrode placement and the number and mixture of MU's sampled (LeVeau & Andersson, 1992; Perry, 1992; Winter, 1991). Two methods of normalization that are commonly used include percent maximal voluntary contraction (MVIC) and percent ensemble average (EA).

Isometric voluntary contractions at 100% or 50% can be used for normalization. These methods involve performing a reference maximal voluntary isometric contraction. The myoelectric values obtained from testing procedures (of the gait cycle for example) are then expressed as a percentage of reference MVIC or as a percentage of the value which is 50% of the MVIC.

Percent of the ensemble average (EA) is another method of normalization. Once an EA is created, the maximum or mean EMG is recorded. This (mean or maximum) value is then compared to the mean EMG determined for each 1% of
the gait cycle. A ratio is created between the maximum (or mean) EA and the EA
at each 1% of the gait cycle. Myoelectric values are thus expressed as a
percentage of the peak or mean EA EMG obtained during the gait cycle (Winter,

Yang and Winter (1984) found that the 50% MVIC method was shown to
have greater intersubject variability, as indicated by the coefficient of variation
(CV), than the peak or mean ensemble average methods. Thus, Yang and Winter
(1984) suggested that peak or mean EA in gait analysis should be the
normalization method of choice because it increased the sensitivity of surface
EMG. However, Knutson et. al (1994) stated that CV was inversely related to
reproducibility (reliability), and suggested that the method of normalization
should be based on reproducibility because "sample data which is more
reproducible will more accurately reflect the population" (p. 48). To that end,
Knutson et al. compared MVIC, peak dynamic EMG, and mean dynamic EMG
normalization methods to determine which produced the most reproducible data
set. They used intraclass correlation coefficient (ICC), and variance ratio (VR) as
the statistical tools to measure reproducibility. Results showed that VR and ICC
values were higher for the MVIC method, which indicated increased
reproducibility compared to the other two normalization methods studied.
Additionally, Perry (personal communication, February 22, 1997) stated that
normalizing to percentage of MVIC was more useful. This process allows
examiners to compare the submaximal activity during gait to relative maximum
potential of that same muscle during a voluntary contraction. Perry concedes that in cases where individuals demonstrate decreased selectivity (such as in cerebral palsy), EA is the best choice because these individuals may be unable to voluntarily elicit maximal contractions of selected muscles.

Perry et al. (1981) also used another form of normalization when measuring crosstalk from surface electrodes over gastrocnemius and soleus, while simultaneously recording from the same muscles with fine wire electrodes. Data from each electrode were normalized to the sum of all data collected from that electrode over all trials for each subject. This accommodated for the differences in motor unit sampling between subjects and between electrode type, as well as other nuances between the two types of electrodes. According to Perry, once normalization is performed in this fashion, data from fine wire and surface electrodes can be compared.

Reproducibility and Validity

When interpreting gait EMG data, signal reproducibility and validity must be considered. Validity of EMG measurements is difficult to assess and often assumed, without considering the issues of crosstalk, selectivity of the electrodes, artifacts, and appropriate processing. Reproducibility measures can be improved by controlling for several factors: performing the EMG testing once on a given day; selecting the best type of electrode for the task; proper preparation of the recording site; optimal inter-electrode spacing; and location of electrodes in reference to anatomical landmarks (which may be more difficult
with fine wire due to the small surface area). Anatomical variation between muscles and motivation are difficult factors to control (Soderberg & Knutson, 1995).

Yang and Winter (1984) reported that EMG values recorded during an MVIC taken at one joint angle and applied to a dynamic activity such as gait, resulted in increased intersubject variability. They advised against using this method. Knutson, Soderberg, Ballantyne and Clarke (1994) also found variability to be increased when using MVIC versus mean or peak dynamic EMG as the reference for dynamic EMG analysis. However, in the same study Knutson et al. found that measures of reproducibility were higher when using MVIC. Therefore, Knutson et al. recommended using MVIC as the reference value when recording dynamic EMG.

When comparing subjects or different muscles of a single subject, data should be normalized. Normalization takes out the variations that would normally prevent direct comparisons. Yack and Winter (1986) found that normalization reduced variability by 50% when compared to unnormalized data. These researchers also compared variability between different muscle groups. They found that the least variable EMG patterns were in the most distal, single joint muscles, while the most variable were in more proximal and biarticulate muscles.

Limited research has been done on the reproducibility (reliability) of surface and intramuscular electrodes. However, Jacobson et al. (1995) found that
fine wire electrodes within vastus medialis and biceps femoris had approximately equal variance ratios when compared to surface electrodes.

Kadaba et al. (1985) found that reproducibility, as determined by variance ratio, (VR), was greater for surface versus intramuscular electrodes. They suggested that lower repeatability for fine wire was likely due to the inability to sample the same volume of muscle on different days even with accurate re-application of electrodes. However, Kadaba et al. did not report using a normalization method in their data analysis. Knutson et al. (1994) noted that there has been confusion in the literature between variability and reproducibility when interpreting normalized EMG data. Therefore, the authors used four statistical measures, inter and intrasubject coefficients of variation (CV), VR, and ICC, to evaluate which of three normalization values rendered the most reproducible data set. According to the authors, inter-subject CV is typically used to measure group variability, and intra-subject CV is used to measure precision; nonetheless, these measures were included in the study since other literature have used these terms interchangeably. The authors concluded that when measuring reproducibility of a normalized EMG data set, VR and ICC should be used.

**Surgical Interventions**

Surgery is a common method of treating gait disturbances in patients with cerebral palsy. According to Fulford, "general indications for surgical intervention are increasing deformity and lack of functional improvement in
spite of adequate therapy" (p. 56, 1990). Almost every child with spastic CP has orthopedic surgery at least once in his or her lifetime (Binder, 1989).

Many authors agree that clinical examination and observational gait analysis have not been consistently reliable and valid methods for assessing gait disturbances or in predicting surgical outcomes (DeLuca, 1991; Sutherland 1978; Etnyre, Chambers, Scarborough & Cain, 1993; Gage, Fabian, Hicks & Tashman, 1984). According to Perry and Antonelli (1993), difficulty in making predictable operative results (prior to gait analysis) with tendon transfers in children with CP was due to lack of understanding of the behavior of the muscle transferred. Prior to the utilization of dynamic EMG with gait analysis, observation (Gage, 1993), palpation (Close & Todd, 1959) and manual muscle testing were the primary methods available to assess pre- and postoperative muscle activity. Manual muscle tests, range of motion, and gait observations have given the practitioner general information to develop treatment regimens; however, these assessment tools failed to provide the necessary details to determine the patients’ overall coordination for locomotion. (Etnyre et al., 1993). According to DeLuca (1991), static examination is not sophisticated enough to differentiate the complex patterns of neuropathology seen in gait patterns of children with CP. Primary deficits, which are a direct result of central nervous system (CNS) damage, must be distinguished from secondary deficits, which are compensations for that damage (Gage, 1991). Differentiating between primary and secondary deficits can be difficult, if not impossible, using clinical
examination alone. Two children may show similar joint range of motion, muscle strength and flexibility, and muscle tone in the clinical examination, but different pathomechanics, as revealed by computerized gait analysis with dynamic EMG (DeLuca, 1991).

Literature has shown that gait analysis with dynamic EMG provides the most accurate method of preoperative and postoperative assessment (Etnyre et al., 1993; Gage et al., 1984; Sutherland, 1978; Thompson & Hoffer, 1991). Etnyre et al. (1993) used pre- and postoperative measurements of gait, including EMG and joint range of motion, to study the effects of surgical correction of equinus gait in twenty-four patients with spastic cerebral palsy. Muscle activity in the triceps surae muscles during gait was measured by EMG before and after surgery. Results showed that the mean postoperative duration (53.7% of the gait cycle) of triceps surae EMG activity for all patients was significantly less than the average preoperative (61.8% of the gait cycle) duration. The authors speculated that surgical intervention effected the level of spasticity or the motor program for walking, resulting in a more normal pattern of EMG during each gait cycle.

EMG can differentiate muscles which exhibit prolonged activity and/or are contracting out of phase. Therefore, dynamic EMG can help determine the most appropriate treatment (i.e. tenotomies, tendon or muscular lengthening, tendon transfer). If abnormal patterns of muscle activity are assumed or not accurately identified, adverse surgical outcomes could result. For example, White (as cited in Perry & Antonelli, 1993) used fine wire EMG to assess the
activity of the posterior and anterior tibialis' in children with CP and varus foot deformity. Varus deformity may be due to prolonged activity of the tibialis posterior and anterior, both invertors of the foot (Kendall et al., 1993). Since a spastic posterior tibialis is commonly found in children with varus foot deformity, it was reasoned that surgical lengthening and/or tendon transfer of this muscle would help correct varus foot deformity. However, EMG analysis revealed that prolonged activity of the tibialis posterior was not always the cause of the varus foot deformities (Perry & Antonelli, 1993). Prolonged activity of the posterior tibialis was found in only 45% of the cases. In 55% of the cases prolonged activity of the anterior tibialis was found. The posterior tibialis was acting out of phase (swing rather than stance) in 11% of the patients. In 75% of the patients, dysfunction of both anterior and posterior tibial muscles was found. These EMG findings demonstrate the value of EMG in gait analysis when making decisions about surgical intervention for gait abnormalities in children with CP.

**Spastic Valgus Foot Deformity**

Normally, the foot is everted by the peroneus brevis and peroneus longus in stance and by the extensor digitorum longus in swing phase. The peroneus brevis acts on the hindfoot while the longus controls the forefoot (Fulford, 1990). Hindfoot valgus, a deformity common in individuals with diplegic CP (Bleck, 1987), is usually due to a muscle imbalance (Fulford, 1990). One explanation for
this is that one of the hindfoot evertors, peroneus brevis, may be stronger than
the hindfoot invertors (Fulford, 1990). Skinner and Lester (1985) analyzed
dynamic EMG of thirteen children with CP who had a valgus hindfoot
deformity. The authors described three different patterns of muscle activity to
explain the valgus position: hyperactive peroneals with a strong posterior tibial
muscle, hyperactive peroneals with a weak posterior tibialis, and hyperactive
extensor digitorum longus (Bleck, 1987).

Spastic peroneals have been recognized by orthopedic surgeons as the
major force leading to a valgus deformity (Bleck, 1987). The literature reports
two surgical procedures that have used the peroneals to correct valgus deformity
in children with CP. One procedure is a peroneus brevis tendon transfer.
Bennet, Rang and Jones (1982) performed peroneus brevis transfers on five
children with CP and valgus foot deformities based on the preliminary work of
Drennan and Sharrad (1971) and Duckworth and Smith (1976). From the 1971
and 1976 reports, peroneus brevis transfers were performed based on weak or
inactive posterior tibialis muscles and strong evertors and dorsiflexors. Bennet et
al., through dynamic fine wire EMG, also revealed inactivity of the posterior
tibialis muscle in each child. Based on these findings, and the support from
previous studies (Drennan & Sharrad, 1971; Duckworth & Smith, 1976), Bennet
and co-workers transferred the peroneus brevis to the tendon of the tibialis
posterior in all five children. Follow-up results of the surgeries were not
reported.
Other surgeons have lengthened the peroneus brevis to correct for hindfoot valgus. Nather et al. (1984) lengthened the peroneus brevis in 14 feet with equinovalgus deformity. Results revealed that the equinus deformity was decreased in 50%, and was corrected completely in 25% of the feet. In conjunction with clinical findings, Perry and Hoffer (1977) examined fine wire EMG profiles of both the peroneus brevis and peroneus longus during gait to determine the best surgical procedure to correct valgus hindfoot deformity. When either the peroneus brevis or longus was active only during stance phase, it was transferred posteromedial to the insertion of posterior tibialis. This converted the transferred muscle to an invertor instead of an evertor. When the peroneus longus or peroneus brevis was active throughout the gait cycle, it was released. These surgical procedures were modified according to individual EMG profiles. For example, in a patient who showed “well defined” peroneus brevis activity during stance, the peroneus longus was transferred to the navicular and peroneus brevis was transferred to the stump of peroneus longus. This was done to allow peroneus longus to function as a dorsiflexor and peroneus brevis to provide dynamic support to the longitudinal arch and continue to act as an evertor. In another patient, whose EMG showed continuous activity in both peroneals, the peroneus longus was lengthened, and the brevis was transferred distally to the stump of longus. In those patients whose peronei were transferred, postoperative EMG recordings showed that the muscles did not
change their gait phase patterns. By the six month follow-up, all patients had become brace-free and had achieved the desired ankle and foot function.

**Summary and Implications for Study**

Dynamic EMG and computerized kinematic and kinetic analysis are often used to evaluate abnormal gait and assist with treatment decisions. In particular, dynamic EMG is useful when considering surgery to correct or improve abnormal gait patterns in children with CP. There is continued debate on the choice of electrodes to use with dynamic EMG. It is appropriate to use surface electrodes when information regarding large superficial muscles is needed, since surface electrodes are easy to apply and present a low risk to the patient. Basmajian (1985) recommended using surface electrodes on children to avoid the discomfort of inserting wire electrodes. Also, surface electrodes have been shown to be more reliable than wire electrodes (Giroux & Lamontagne, 1990; Kadaba et al., 1985). However, crosstalk can result in inaccurate EMG data when surface electrodes are used. It is generally accepted that fine wire electrodes should be used when studying either deep or small muscles. It has also been suggested that fine wire electrodes are most appropriate when studying specific activity of individual muscles, especially when tendon transfer or lengthening is being considered (Perry et al., 1981).

Crosstalk has been demonstrated in the gastrocnemius and soleus muscles of the calf, which are neither deep nor small (Perry et al., 1981). The peroneal muscles, also neither deep nor small, bear some anatomical similarities
to the gastrocnemius/soleus complex. Like the soleus, the peroneus brevis is partially covered by another muscle, the peroneus longus. Therefore, it may be difficult to differentiate where peroneal muscle EMG signals originate from when surface electrodes are used.

Surface electrodes placed over the peroneus brevis muscle have registered EMG signal when the tibialis anterior muscle was artificially stimulated (DeLuca & Merletti, 1988). There is debate as to whether this was crosstalk or overflow electrical stimulation to the peroneus brevis (Solomonow et al., 1994). However, the tibialis anterior is adjacent to the peroneals and has a fairly large cross-sectional area which can create a large volume of EMG signal (Morrenhoff & Abbink, 1985). Additionally, the peroneus longus, tibialis anterior, and extensor digitorum muscles all share a common fascial connection at their origin (Kendall et al., 1993). For these reasons, the possibility of crosstalk from tibialis anterior to a surface electrode over peroneus longus is a reasonable concern. Crosstalk from the tibialis anterior muscle to the peroneals would likely appear as prolonged activity due to tibialis anterior’s function during swing phase of the gait cycle. Because of its common fascial attachment with peroneus longus and close proximity to both peroneus brevis and longus, crosstalk from the extensor digitorum longus to a surface electrode over either peroneal muscle is also a possibility. To date, no study has been published which specifically examines crosstalk in the peroneal muscles during a dynamic activity with natural muscle activation.
The peroneus longus and peroneus brevis muscles are sometimes considered for surgical lengthening or tendon transfer when a child with CP presents with an equinovalgus or valgus foot deformity. When considering specific surgical procedures, it is imperative that the EMG analysis is accurate. There is a need to know the specific timing and relative intensity of each of the peroneal muscles in such a case. It is also important to avoid the risk of infection and the discomfort involved with intramuscular electrodes whenever possible without compromising the accuracy of the EMG analysis. Therefore, the question of which type of electrode to use when assessing peroneal muscle activity is valid. The intention of this study was to assess whether crosstalk from tibialis anterior, extensor digitorum longus, or the other peroneal muscle (longus or brevis) contributes to the EMG signal from surface electrodes over the peroneus brevis and longus muscles. If crosstalk is minimal, surface electrodes can be used with confidence over the peroneus longus and brevis muscles. However, if there is a high level of crosstalk in the EMG signals from surface electrodes over the peronei, fine wire electrodes should be used prior to surgical decision-making.
CHAPTER 3

METHODS

Subjects and Study Site

This study was conducted at the Center For Human Kinetics Studies (CHKS). The CHKS is operated cooperatively by Mary Free Bed Hospital & Rehabilitation Center and Grand Valley State University. Fourteen normal volunteer subjects over the age eighteen were recruited from a sample of convenience. Subjects were screened through a medical history form (Appendix A) and a brief clinical examination (Appendix B). Exclusion criteria are presented in Appendix C. The examination was performed by a student physical therapist and supervised by a licensed physical therapist. Each subject was required to read and sign a consent form approved by the Human Subjects Review Boards of both Grand Valley State University and Mary Free Bed Hospital (Appendix D). Subjects participated in a one-visit test. The test included the following: clinical examination, insertion of fine wire electrodes into the four lower leg muscles being studied (extensor digitorum, anterior tibialis, peroneus longus, peroneus brevis), the application of surface electrodes over the two peroneal muscles of the right lower extremity only, and five trials each of manual muscle tests for the extensor digitorum longus, anterior tibialis, and peroneals. During pilot testing, the test procedure also included a gait test.
Instrumentation

Foot Switches

For the pilot test, BTS\(^1\) footswitch devices were used in order to synchronize EMG activity to the contact phases of the gait cycle. Footswitches were used to mark the point of first initial contact, toe off, and second initial contact on the EMG recordings during gait. The footswitches were embedded in rubberized shoe inserts that were secured with tape to the bottom of the subjects’ feet. Subjects were tested with a light sock over each footswitch and without shoes. A wire lead from the footswitches was connected to a patient unit that was strapped around the subject’s waist. The footswitch information was transmitted to the acquisition computer (basic unit) via a fiber optic cable.

Forceplates

For the pilot test, two AMTI\(^2\) force plates mounted flush with the floor, were used to measure first initial contact, toe off, and second initial contact. Force plate data was used to verify accuracy of the footswitch data. Spacing of the force plates was adjusted to accommodate for individual stride lengths. The force plates were covered by carpet so subjects were not conscious of stepping on them. The force plates recorded bodyweight forces and moments in three dimensions with foil strain gauges. Force data were sampled at 1000 Hz. The

\(^1\) Bioengineering Technology Systems, Milan, Italy

\(^2\) AMTI, Advanced Medical Technologies Inc., Newton, MA
force plates were set to collect data when 15 N (3.37 lbs.) was exerted on the plate to decrease the risk of false triggers.

Electromyography

A TELEMG³ Multichannel Electromyograph was used to measure myoelectric activity. This electromyograph has a 10 bit resolution and a CMMR of greater than 10dB. The surface electrodes were bipolar silver/silver-chloride discs, 8 mm in diameter, and had a differential impedance of one megaohm. The fine wire electrodes were made of Nylon Karma Alloy wire and were introduced into the target muscle with a disposable 26 ½ in. gauge needle. Fine wire electrodes for this study were prepared at the Center For Human Kinetic Studies (Figure 1) using the following protocol:

1. Materials included non-reusable 26 ½ in. gauge needles, fine wire, butane lighter, scissors, and sterilization pouches.
2. The needle was uncapped; an 18-20 inch length of fine wire was inserted into the beveled end of the needle and pulled through creating a loop.
3. Approximately 4 mm of wire insulation was burnt off at the middle of the loop.
4. The loop was cut, and approximately 1-2 mm of exposed wire was left at each end.

³ TELEMG, Bioengineering Technology Systems, Milan, Italy
5. The ends of the wires were staggered so that their exposed ends would not make contact with each other.

6. The ends of the wires were pulled back to meet the end of the needle and were gently bent over the edge of the needle creating a barb. The needle was re-capped.

7. The needle-fine wire apparatus was placed in a sterilization pouch, and the wires were kept away from the edges of the bag. The pouches were sent to St. Mary's Health Service (SMHS) for gas sterilization.

During testing, the surface and intramuscular EMG signals were collected at 2000 Hz, amplified with a gain of 100, and sent to a BTS\(^4\) 905 Transmitter Unit (patient unit). In the patient unit myoelectric signals were filtered with a bandpass of 1-800 Hz to minimize artifact. EMG signals underwent analog to digital conversion and serial formatting and were transmitted through a 100-micron optical fiber to the basic unit. At the basic unit, signals were amplified, converted from digital to analog, and saved as a raw file for further filtering.

Electric Stimulation Unit

A TD 20 EMG/EP\(^5\) electric stimulator was used to verify correct placement of the fine wire electrodes within the target muscle. Parameters which included duration, repetition rate, intensity, and mode, were set to each subject's

\(^4\) Bioengineering Technology Systems, Milan Italy

\(^5\) TECA Corporation, Three Campus Drive, Pleasantville, New York
Figure 1. Steps in Making a Bipolar Wire Electrode From Muscles Alive (5th ed.)

tolerance. When the target muscle was electrically stimulated through the fine wire electrode, palpation and/or observation of a contraction of that muscle verified correct placement.

**Procedures**

Following the signing of the consent form and completion of the clinical examination, subjects were prepared for electrode application and insertion. Preparation for electrode application involved shaving excess hair and using alcohol wipes to remove dirt, oil and dead skin overlying targeted muscles on the right lower extremity. These steps were necessary to reduce skin impedance and signal artifact. During fine wire insertion and surface electrode application, each subject was positioned long-sitting on a plinth. A towel roll was placed under the subject's right knee for comfort.

A licensed physical therapist, trained and practiced in fine wire insertion, inserted all fine wire electrodes. The same student physical therapist assisted with fine wire insertion and applied the surface electrodes for all subjects to minimize interrater error. All wire electrodes were inserted and removed following the fine wire application procedure approved by Mary Free Bed Hospital and Rehabilitation Center. The following protocol was used. First, the therapist and assistant washed their hands and donned clean gloves. Next, an assistant opened the needle pack. Holding the needle base the assistant carefully slid the needle out of the plastic hypodermic case. She then confirmed that the wires within the bevel of the needle slid independently of each other, and that
the ends of the wire had been bared (the insulation was burnt off) and were not touching. The assistant then handed the electrode to the therapist. An EMG guide/atlas was used to locate the insertion point and direction of application of the needle/wire (Perotto, 1994). An EMG recording sheet was used during the testing procedure (Appendix E). The order of the muscles targeted with fine wire and/or surface electrodes and their corresponding numbered connector was recorded on the sheet. The fine wire electrodes were inserted in the following order: peroneus longus (channel one), extensor digitorum (channel two), anterior tibialis (channel three), and peroneus brevis (channel four). The wire leads from the surface electrodes of peroneus longus and brevis were connected to the patient unit and labeled as channel five and six, respectively.

To locate correct placement for fine wire insertion into the peroneus longus, the therapist measured three fingerbreadths inferior to the lateral aspect of the subject's fibular head (Figure 2). The subject plantarflexed and everted the foot which facilitated palpation of the contracted peroneus longus muscle belly. To locate correct placement for the extensor digitorum, the therapist palpated four fingerbreadths down from the tibial plateau and four fingerbreadths lateral to the tibial crest (Figure 3). The subject extended his or her toes so that a contraction of the extensor digitorum could be palpated. This method was a modification of Perotto's recommended procedure (1994) and was used because greater accuracy was found in targeting the extensor digitorum, as verified by electrical stimulation. To find correct placement for the anterior tibialis
electrode, the therapist palpated four fingerbreadths from the tibial plateau and one fingerbreadth lateral to the tibial crest (Figure 4). The subject dorsiflexed and inverted the foot so that a contraction of the anterior tibialis muscle belly could be palpated. To locate the peroneus brevis muscle, the therapist placed his fingers one handbreadth proximal to the lateral malleolus and anterior to the peroneus longus tendon (Figure 5). The subject planterflexed and everted the foot so that a contraction of the peroneus brevis could be palpated.

Once correct finger placement of each target muscle was confirmed by palpation and/or muscle contraction, the therapist inserted the needle. He was careful to touch only the hub and not the needle of the apparatus. The therapist then discarded the used needle into a Sharps container. The small amount of bleeding which resulted from needle penetration was cleaned with peroxide-soaked gauze pads.

Following each fine wire insertion, the therapist then performed 2-3 passive movements of the subject's right ankle. The subject then performed 3-5 active contractions of the targeted muscle to ensure that the barbs of the wires were securely set. With an active movement specific to the muscle that received the wire electrode, raw EMG signal was then checked for noise with an oscilloscope. A connector and preamplifier were secured adjacent to the wire insertion site with tape. Next the free ends of the electrode wire were looped and attached to the spring connectors. Then, the insulation on the wire within the connector was removed by rubbing/twisting the spring and excess wire was
clipped away from the spring connectors. Attention was directed to the signal and the noise made by the oscilloscope, when each connector site was tapped and when the subject contracted the muscle. Contraction of the desired muscle produced a noise that sounded smooth and consistent. A noisy, abrupt sound inconsistent with any pattern possibly indicated that the wires were touching and shorting out. In this case the therapist adjusted the wires by pulling one of them out slightly. If there were still problems with the signal, the electrodes were disconnected and then re-attached to the spring connectors or the wires were removed and the procedure was repeated.

Correct placement of the wires in the target muscle was also assessed with electrical stimulation. The cable from the electrical stimulator was attached to the spring connectors using EZ clips; a ground electrode was taped to the right patella. The therapist stimulated the electrode by increasing the voltage until a muscle contraction was observed or palpated. When electrode placement was confirmed, "CBS" (confirmed by stimulation) was documented on the EMG recording sheet. Two electrodes were not confirmed by stimulation (anterior tibialis electrode in subject 11 and peroneus brevis electrode in subject 15). However, in both cases, correct electrode placement was verified by the oscilloscope, and by observing the raw EMG data for expected muscle activity during the testing procedure. If electrical stimulation revealed that the placement was not accurate, the wire was removed and the procedure was repeated using a fresh needle preparation.
Figure 2. Location and fine wire placement of peroneus longus wire electrode.


Copyright 1994 by Charles C. Thomas Publisher.
Figure 3. Location and fine wire placement of extensor digitorum longus electrode. From Anatomical Guide for the Electromyographer: The Limbs and Trunk (3rd ed.) (pp. 142, 143) by A. Perotto, 1994, Charles C. Thomas Publisher: Springfield, IL. Copyright 1994 by Charles C. Thomas Publisher.
Figure 4. Location and fine wire placement of anterior tibialis wire electrode.


Copyright 1994 by Charles C. Thomas Publisher.
Figure 5. Location and fine wire placement of peroneus brevis wire electrode.


Copyright 1994 by Charles C. Thomas Publisher.
Surface electrodes were applied over the peroneus longus and peroneus brevis muscles immediately following the fine wire insertion into each peroneal muscle respectively. The bipolar surface electrodes were applied with an orientation that was parallel to the direction of the underlying muscle fibers. For each of the peroneal muscles, one disc of the bipolar electrodes was placed proximal to and one was placed distal to the fine wire electrode insertion site (Soderberg, 1992). The electrodes were spaced approximately 10-15 mm center-to-center.

Following verification of fine wire electrode placement and surface electrode application, leads from all of the electrodes were secured into the patient unit which the subject wore around his or her waist. The lead wires from all six channels were taped securely to the leg to reduce artifact from movement of the wires and to reduce the likelihood of tension on, and loosening of the electrodes.

**Pilot Study**

A maximal voluntary contraction was elicited from each of the targeted muscles using Kendall et al.’s (1993) isometric manual muscle tests (MMT). To reduce discrepancies in isometric testing between testers, the same student physical therapist performed all MVIC tests.

For each test, the subject laid supine. To insure that the proper testing position was used, a second student physical therapist observed and verified proper testing positions and procedures. For the anterior tibialis test position,
the examiner supported the leg just above the ankle. The ankle was dorsiflexed to neutral and inverted. The examiner then applied pressure against the medial dorsum of the foot in the direction of plantarflexion of the ankle and eversion of the foot. For the extensor digitorum MMT, the examiner stabilized the ankle in neutral dorsiflexion and applied pressure against the dorsum of the proximal phalanges in the direction of flexion. The peroneal muscles were tested with subject supine as well, but the lower extremity was positioned in medial rotation. The examiner supported the leg above the ankle, which was plantarflexed and everted. Pressure was applied against the ventral and lateral border of the foot in the direction of inversion of the foot and dorsiflexion of the ankle.

Simultaneous EMG recordings were made from the surface electrodes over the peroneus longus and brevis muscles and from wire electrodes in all four muscles investigated. The recordings were stored in the computer for later processing. The muscle test recording interval was 10 seconds. During each MVIC, the tester repeatedly encouraged each subject to "hold," to help insure that a maximal contraction was maintained. The order of muscle testing was randomized for each subject. A trial consisted of the three separate manual muscle tests. A total of three trials were taken on all subjects.

Footswitches, which also acted as a ground for the fine wire electrodes, were applied after muscle testing. Footswitches matching the size of the patient's feet were taped to the bottom of each foot. The footswitch was cabled to the patient unit, which was in turn connected to the basic unit via a fiber optic
cable. Next, to insure that the subject did not exhibit an abnormal gait pattern during testing, the subject walked to become accustomed to the equipment, and to alleviate discomfort and/or muscle cramping from the fine wire electrodes. The subject then was instructed to walk across the test space at a self-determined, natural pace. Starting position was adjusted to accommodate for the subject's stride length and to insure that first initial contact of the right leg was on the first force plate.

The test involved three trials, as recommended by Arsenault (1992). For each trial, the subject stepped consecutively on both force plates. The first and second initial contacts of the right foot were marked while the footswitch simultaneously recorded the same information. EMG recordings were taken from all electrodes during the three trials.

Following the three trials, raw signals were visually checked for artifact. Once successful EMG data collection had been confirmed, all equipment was removed from the subject. With the examiner wearing clean gloves, wire electrodes were removed while applying gentle skin pressure near the insertion site. The ends of the wire were checked to see that no portion of the wire remained embedded in the muscle (the barbs were easily spotted). The used wire was discarded in a red plastic bag labeled with a biohazard symbol. A peroxide-soaked gauze pad was used to remove any blood that remained from the fine wire needle insertions.
Actual Test

The results of the pilot study showed that fine wire raw EMG data recorded during gait were unacceptable. Upon visual inspection, muscular activity did not seem to follow the normal pattern during gait, and there appeared to be mechanically induced artifact in the raw EMG data from the fine wire electrodes. This artifact could not be removed with filtering. The following efforts were made to minimize noise collected by the fine wire electrodes during gait: the amplifiers and the leads of the fine wires were further stabilized and fastened with tape; the free ends of the fine wires were attached to the coils of the spring connectors as close to the base of the amplifier as possible; smaller loops were made when connecting the free ends of the wire to the spring connectors; an alternative method that replaced the spring connectors was used to secure the free ends of the fine wire; any blood that had dried and scabbed around the fine wire insertion sites was removed with a peroxide-soaked gauze pad; a subject was instructed to walk with a slow cadence; while walking, a subject held hands with another person who acted as a ground; an additional ground was used along with the footswitches. Combinations of all these attempts to reduce the noise in the data from the fine wire electrodes were inadequate. Therefore, the gait test was not used in this study as was originally proposed.

The study was then modified and EMG data were recorded only during the maximal voluntary isometric contractions of extensor digitorum, anterior tibialis, and peroneus longus and brevis. The following procedure was used.
Subject preparation, fine wire insertion and surface electrode application procedures were followed as reported earlier. A ground which consisted of a wet Velcro strap, was wrapped around the proximal aspect of the left tibia, replacing the footswitches. After all leads were taped down and secured into the patient unit, the subject was asked to walk around until they did not feel any discomfort from fine wire insertion. This was done to insure that pain would not interfere with eliciting a maximal muscle contraction. Resting EMG data were recorded at the beginning of each testing session; however, they were not used in data analysis. One trial consisted of the three separate manual muscle tests and the number of total trials per subject was increased to five. EMG data were collected for a period of five seconds during each MVIC test. The same student physical therapist collected the EMG data, and visually inspected them after each test was performed. If there was evidence of excessive noise during a specific test, then the data were discarded and the same test was repeated. If data were still suspicious, verification of proper electrode placement with electrical stimulation was performed again. After all trials were completed, the electrodes were removed, as earlier described.

Data Analysis

Data Processing

Following data collection, raw EMG data were time cut to include only one second of data, high and low pass filtered, and full wave rectified. Time
cutting was done by visual inspection. A custom Matlab\textsuperscript{6} software program was used to filter both surface and intramuscular EMG. Surface signals were high pass filtered at 20 Hz and low pass filtered at 400 Hz. Raw intramuscular signals were also high pass filtered at 20 Hz and low pass filtered at 800 Hz. Raw signal waveforms were full-wave rectified and then saved as an EMG file for further processing using Excel \textsuperscript{7}.

In Excel, data from each electrode, for each MMT test, were averaged over all trials for each subject and then normalized to the sum of all data collected from that electrode. This process of normalization accommodates for the differences in motor unit sampling between the two types of electrodes and allows for their comparison (Perry et al., 1981).

Statistical Analysis

The normalized EMG data (NU) from the surface electrodes were examined relative to the signal from the wire electrodes from all the muscles tested in order to determine the relative selectivity of the surface electrodes. Normalized EMG units from the wire electrodes for the peroneus longus, peroneus brevis, tibialis anterior, and extensor digitorum were considered the independent variables. A Pearson product-moment correlation was used to determine the relationship among the independent variables, and between the

\textsuperscript{6} Matlab version 5.0, Mathworks Inc., Natick, MA, 1996

\textsuperscript{7} Microsoft® Excel 97, Microsoft Corporation
dependent variables (surface EMG data from peroneus longus and peroneus brevis) and each of the independent variables.

Ideally, if no crosstalk existed, all of the variance in surface EMG of the peroneus longus (SPL) and peroneus brevis (SPB) could be explained by the respective fine wire data. To investigate the variance in SPL and SPB data, stepwise multiple regression was used. Multiple regression is appropriate for use when analyzing the relationship between one dependent variable and several independent predictor variables. Stepwise multiple regression maximizes the prediction accuracy with the least number of predictors (Portney & Watkins, 1993). Using SAS / STAT® computer program a stepwise multiple regression was run on all data sets with a qualifying F statistic in order to determine significant contributors to the data from the surface electrodes for the peroneus longus during each test condition. The same test was then applied to the data from the peroneus brevis. Entrance criteria for the regression analysis was set at $p \leq 0.10$.

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8 SAS / STAT version 6, SAS Institute Inc., 1989
CHAPTER 4

RESULTS

The purpose of this study was to assess whether myoelectric signals as measured by fine wire electrodes inserted into anterior tibialis (WAT), extensor digitorum longus (WED), peroneus longus (WPL), or peroneus brevis (WPB) contributed to EMG signal (i.e. crosstalk) recorded from the surface electrodes over the peroneus brevis (SPB) and longus muscles (SPL).

The subjects of this study included 8 males and 7 females for a total of 15 subjects (Table 1). One female subject (13) was excluded due to evidence that the WED had been displaced, bringing the total number of subjects to 14. On visual inspection of raw data from this subject after testing was complete, it appeared that this electrode had migrated or had been displaced into peroneus longus. Electrical stimulation confirmed that the electrode was in peroneus longus. In addition, subjects 5 and 7 were considered outliers based on review of their normalized data (Appendix F). In subject 5, WED demonstrated more activity (0.145118 NU) during the anterior tibialis MMT than during its own MMT (0.04858 NU). Similarly, in subject 7 the WED recorded more activity during the peroneal MMT (0.14271 NU) than during its own MMT (0.053948 NU). Therefore, data analysis was performed first with all 14 subjects, and then was
repeated with the two outliers removed. The two analyses were then compared.

A few significant differences were noted between the two groups, which will be discussed later.

Table 1

Demographic Data

<table>
<thead>
<tr>
<th>Subjects</th>
<th>Age (years)</th>
<th>Weight (lbs.)</th>
</tr>
</thead>
<tbody>
<tr>
<td>8 males</td>
<td>28 ± 5.44</td>
<td>177 ± 24.19</td>
</tr>
<tr>
<td>6 females</td>
<td>26 ± 3.56</td>
<td>155 ± 29.25</td>
</tr>
</tbody>
</table>

The raw data from each electrode for each MMT were time cut to one second. This one-second was rectified, then high and low pass filtered. The filtered data were then normalized to all data collected from their own electrode. These normalized data for all subjects are displayed in Appendix F. Averages of the normalized data for all 14 subjects were then calculated (Table 2) along with average relative (percentage of) EMG per electrode average MVIC recording (Table 3).

Table 2

Average of Normalized Units for 14 Subjects

<table>
<thead>
<tr>
<th></th>
<th>Anterior Tibialis MMT</th>
<th>Peroneus MMT</th>
<th>Extensor Digitorum MMT</th>
</tr>
</thead>
<tbody>
<tr>
<td>WPL</td>
<td>0.016525</td>
<td>0.149874</td>
<td>0.033602</td>
</tr>
<tr>
<td>WED</td>
<td>0.056878</td>
<td>0.0534</td>
<td>0.089723</td>
</tr>
<tr>
<td>WAT</td>
<td>0.11632</td>
<td>0.014963</td>
<td>0.068016</td>
</tr>
<tr>
<td>WPB</td>
<td>0.029415</td>
<td>0.134357</td>
<td>0.036228</td>
</tr>
<tr>
<td>SPL</td>
<td>0.046653</td>
<td>0.106395</td>
<td>0.046952</td>
</tr>
<tr>
<td>SPB</td>
<td>0.03536</td>
<td>0.115923</td>
<td>0.048717</td>
</tr>
</tbody>
</table>
Table 3

<table>
<thead>
<tr>
<th></th>
<th>Anterior Tibialis MMT</th>
<th>Peroneus MMT</th>
<th>Extensor Digitorum MMT</th>
</tr>
</thead>
<tbody>
<tr>
<td>WPL</td>
<td>0.138552</td>
<td>1.00</td>
<td>0.394392</td>
</tr>
<tr>
<td>WED</td>
<td>0.74022</td>
<td>0.769527</td>
<td>1.00</td>
</tr>
<tr>
<td>WAT</td>
<td>1.00</td>
<td>0.139792</td>
<td>0.590188</td>
</tr>
<tr>
<td>WPB</td>
<td>0.386266</td>
<td>1.00</td>
<td>0.644097</td>
</tr>
<tr>
<td>SPL</td>
<td>0.381954</td>
<td>0.728543</td>
<td>0.422448</td>
</tr>
<tr>
<td>SPB</td>
<td>0.362518</td>
<td>0.913841</td>
<td>0.754003</td>
</tr>
</tbody>
</table>

Pearson product-moment correlation (r) coefficients were computed, using the normalized data (Appendix F) to determine the relationships between data from the four wire electrodes and between data from surface and wire electrodes (Tables 4 through 7 [14 subjects] & Tables 7 through 9 [12 subjects]). The alpha level was set at p ≤ 0.10 and a critical value for r was determined to be r= 0.458 for a sample size of n=14 (Portney & Watkins, 1993).

During the anterior tibialis MMT, none of the wire electrode data displayed a significant relationship with each other (Table 4). SPL correlated significantly with WPB (r=0.58374, p=0.0284), although it did not correlate significantly with WPL or with WAT or WED. Additionally, SPB data did not correlate with data from any of the fine wire electrodes.
Table 4

Cross Correlations for Anterior Tibialis MMT

<table>
<thead>
<tr>
<th></th>
<th>SPL</th>
<th>WPL</th>
<th>WED</th>
<th>WAT</th>
<th>WPB</th>
<th>SPB</th>
</tr>
</thead>
<tbody>
<tr>
<td>SPL</td>
<td>1.00</td>
<td>0.31302</td>
<td>0.24200</td>
<td>-0.2059</td>
<td>*0.58374</td>
<td>*0.75125</td>
</tr>
<tr>
<td>WPL</td>
<td>*1.00</td>
<td>-0.0819</td>
<td>-0.3059</td>
<td>0.23903</td>
<td>0.02944</td>
<td></td>
</tr>
<tr>
<td>WED</td>
<td>*1.00</td>
<td>0.42731</td>
<td>-0.09250</td>
<td>-0.15956</td>
<td></td>
<td></td>
</tr>
<tr>
<td>WAT</td>
<td>*1.00</td>
<td>-0.39888</td>
<td>-0.6755</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>WPB</td>
<td>*1.00</td>
<td>0.40013</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>SPB</td>
<td>*1.00</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

*p ≤ 0.10

Correlations associated with the peroneal MMT (Table 5) were significant between WPL and WPB (r=0.47633, p=0.0851), demonstrating an interrelationship between these two independent variables. The corresponding peroneus longus electrodes (SPL & WPL) correlated significantly (r=0.67263, p=0.0084), although SPL correlated to a greater extent with WPB (r=0.74478, p=0.0022). SPB correlated significantly with WPB (r=0.69752, p=0.0055) yet to a greater extent with WPL (r=0.76686, p=0.0014). In addition, SPL, SPB, WPL and WPB displayed a significant negative correlation with WAT (r= -0.2059, -0.6755, -0.3059, -0.39888 respectively) (Table 5). This negative correlation was expected because the peroneals are antagonistic to anterior tibialis.
During the extensor digitorum MMT, there was a significant relationship between WPL and WPB ($r=0.59015$, $p=0.0263$). SPL correlated significantly with WPL ($r=0.56751$, $p=0.0343$), WAT ($r=0.50711$, $p=0.0642$), and had the highest correlation with WPB ($r=0.68830$, $p=0.0065$). SPB correlated significantly with WPL ($r=0.60295$, $p=0.0225$) and to a greater extent with WPB ($r=0.65003$, $p=0.0118$) (Table 6).
By inspection, the cross correlation results with outliers excluded (Tables 7 through 9) demonstrated a few differences as summarized below. WPL did not correlate significantly with WPB during the peroneal MMT (Table 8), indicating that there was not a significant interrelationship between the data from these two electrodes. In addition, WAT did not show significant correlation with SPL during the extensor digitorum MMT (Table 9). The final difference noted was that SPB correlated significantly with WED ($r=0.53059$, $p=0.0759$) during the extensor digitorum MMT (Table 9).

### Table 7

**Cross Correlations for Anterior Tibialis MMT**

<table>
<thead>
<tr>
<th></th>
<th>SPL</th>
<th>WPL</th>
<th>WED</th>
<th>WAT</th>
<th>WPB</th>
<th>SPB</th>
</tr>
</thead>
<tbody>
<tr>
<td>SPL</td>
<td>*1.00</td>
<td>0.19818</td>
<td>0.24744</td>
<td>-0.36142</td>
<td>*0.58779</td>
<td>*0.83475</td>
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<tr>
<td>WPL</td>
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<td>0.13788</td>
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<td>0.00812</td>
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<tr>
<td>WED</td>
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<td>0.32705</td>
<td>0.12478</td>
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<tr>
<td>WAT</td>
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<td>-0.43248</td>
<td>-0.4684</td>
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<tr>
<td>WPB</td>
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<td>SPB</td>
<td>*1.00</td>
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*p ≤ 0.10
Table 8
Cross Correlations for Peroneal MMT

<table>
<thead>
<tr>
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<th>WPB</th>
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<td>*1.00</td>
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<td></td>
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<td>WAT</td>
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<td></td>
<td>*1.00</td>
<td>*-0.64746</td>
<td>*-0.69467</td>
</tr>
<tr>
<td>WPB</td>
<td></td>
<td></td>
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<td></td>
<td>*1.00</td>
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<td>SPB</td>
<td></td>
<td></td>
<td></td>
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<td>*1.00</td>
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</table>

*p ≤ 0.10

Table 9
Cross Correlations for Extensor Digitorum MMT

<table>
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<tr>
<th></th>
<th>SPL</th>
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<th>WED</th>
<th>WAT</th>
<th>WPB</th>
<th>SPB</th>
</tr>
</thead>
<tbody>
<tr>
<td>SPL</td>
<td>*1.00</td>
<td>*0.51866</td>
<td>0.25142</td>
<td>0.43722</td>
<td>*0.63759</td>
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</tr>
<tr>
<td>WPL</td>
<td></td>
<td>*1.00</td>
<td>-0.00877</td>
<td>-0.24957</td>
<td>*0.57221</td>
<td>*0.51962</td>
</tr>
<tr>
<td>WED</td>
<td></td>
<td>*1.00</td>
<td></td>
<td>0.32250</td>
<td>0.03531</td>
<td>*0.53059</td>
</tr>
<tr>
<td>WAT</td>
<td></td>
<td></td>
<td></td>
<td>*1.00</td>
<td>-0.05901</td>
<td>0.18177</td>
</tr>
<tr>
<td>WPB</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>*1.00</td>
<td>*0.63306</td>
</tr>
<tr>
<td>SPB</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>*1.00</td>
</tr>
</tbody>
</table>

*p ≤ 0.10

Stepwise multiple regression was used to predict the contribution of muscle activity as measured from the four fine wire electrodes to the muscle activity measured under each of the two surface electrodes (Tables 10 through 12). During the anterior tibialis MMT, WPB made a significant contribution to the EMG data from SPL, accounting for 34% of the variance with a 97% probability that this was not due to chance (Table 10). None of the other muscles
investigated in this study contributed significantly to the variance in SPL.

Activity in SPB was not predicted by activity from any of the muscles targeted with wire electrodes during the anterior tibialis MMT (Table 10).

Table 10

Results of Stepwise Regression for Anterior Tibialis MMT

<table>
<thead>
<tr>
<th>Surface Electrode</th>
<th>Significant muscle</th>
<th>r²</th>
<th>F</th>
<th>Prob &gt; F</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peroneus Longus</td>
<td>Peroneus Brevis</td>
<td>0.3408</td>
<td>6.2026</td>
<td>0.0284</td>
</tr>
<tr>
<td>Peroneus Brevis</td>
<td>Nothing significant</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

During the peroneus MMT, data from only WPL and WPB made significant contributions to the EMG signal detected by SPL and SPB (Table 11). These contributions accounted for 68% and 73% of the total variance respectively.

Table 11

Results of Stepwise Regression for Peroneus MMT

<table>
<thead>
<tr>
<th>Surface Electrode</th>
<th>Significant muscle</th>
<th>r²</th>
<th>F</th>
<th>Prob &gt; F</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peroneus Longus</td>
<td>Peroneus Brevis</td>
<td>0.5547</td>
<td>14.9482</td>
<td>0.0022</td>
</tr>
<tr>
<td></td>
<td>Peroneus Longus</td>
<td>0.1307</td>
<td>4.5694</td>
<td>0.0558</td>
</tr>
<tr>
<td>Peroneus Brevis</td>
<td>Peroneus Longus</td>
<td>0.5881</td>
<td>17.1317</td>
<td>0.0014</td>
</tr>
<tr>
<td></td>
<td>Peroneus Brevis</td>
<td>0.1428</td>
<td>5.8353</td>
<td>0.0343</td>
</tr>
</tbody>
</table>

During the extensor digitorum MMT, data from WPB made the most notable contribution to SPL (see Table 12). WPB explained 47% of the variance with a greater than 99% probability that the variance was not due to chance. When looking at the regression results for SPB, WPB contributed significantly to
the variance seen in data from SPB. In fact, WPB explained 42% of the total SPB variance, with a 99% probability that this was not due to chance (Table 12).

Table 12
Results of Stepwise Regression for Extensor Digitorum MMT

<table>
<thead>
<tr>
<th>Surface Electrode</th>
<th>Significant muscle contributions</th>
<th>r²</th>
<th>F</th>
<th>Prob &gt; F</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peroneus Longus</td>
<td>Peroneus Brevis</td>
<td>0.4738</td>
<td>10.8034</td>
<td>0.0065</td>
</tr>
<tr>
<td></td>
<td>Anterior Tibialis</td>
<td>0.2199</td>
<td>7.8975</td>
<td>0.0170</td>
</tr>
<tr>
<td>Peroneus Brevis</td>
<td>Peroneus Brevis</td>
<td>0.4225</td>
<td>8.7809</td>
<td>0.0118</td>
</tr>
</tbody>
</table>

Stepwise multiple regression was also conducted without the outliers (subjects 2 and 5), which resulted in one significant difference in regression results (Tables 13 through 15) as compared to analysis with all 14 subjects. This difference was found during the extensor digitorum MMT. During this test, WED accounted for 26% of the variance in SPB (Table 15), whereas it did not show a significant contribution in the 14 subject analysis. The probability greater than F=0.0281 indicates a 97% probability that this variance was not due to chance.

Table 13
Results of Stepwise Regression for Anterior Tibialis MMT

<table>
<thead>
<tr>
<th>Surface Electrode</th>
<th>Significant muscle contributions</th>
<th>r²</th>
<th>F</th>
<th>Prob &gt; F</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peroneus Longus</td>
<td>Peroneus Brevis</td>
<td>0.3455</td>
<td>502787</td>
<td>0.0444</td>
</tr>
<tr>
<td>Peroneus Brevis</td>
<td>Nothing significant</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>
### Table 14

**Results of Stepwise Regression for Peroneus MMT**

<table>
<thead>
<tr>
<th>Surface Electrode</th>
<th>Significant muscle contributions</th>
<th>r²</th>
<th>F</th>
<th>Prob &gt; F</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peroneus Longus</td>
<td>Peroneus Brevis</td>
<td>0.5214</td>
<td>10.8949</td>
<td>0.0080</td>
</tr>
<tr>
<td></td>
<td>Peroneus Longus</td>
<td>0.1847</td>
<td>5.6573</td>
<td>0.413</td>
</tr>
<tr>
<td>Peroneus Brevis</td>
<td>Peroneus Longus</td>
<td>0.5319</td>
<td>11.3651</td>
<td>0.0071</td>
</tr>
<tr>
<td></td>
<td>Peroneus Brevis</td>
<td>0.1995</td>
<td>6.6871</td>
<td>0.0294</td>
</tr>
</tbody>
</table>

### Table 15

**Results of Stepwise Regression for Extensor Digitorum MMT**

<table>
<thead>
<tr>
<th>Surface Electrode</th>
<th>Significant muscle contributions</th>
<th>r²</th>
<th>F</th>
<th>Prob &gt; F</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peroneus Longus</td>
<td>Peroneus Brevis</td>
<td>0.4065</td>
<td>6.8497</td>
<td>0.0257</td>
</tr>
<tr>
<td></td>
<td>Anterior Tibialis</td>
<td>0.2267</td>
<td>5.5644</td>
<td>0.0427</td>
</tr>
<tr>
<td>Peroneus Brevis</td>
<td>Peroneus Brevis</td>
<td>0.4008</td>
<td>606880</td>
<td>0.271</td>
</tr>
<tr>
<td></td>
<td>Extensor Digitorum</td>
<td>0.2586</td>
<td>6.8339</td>
<td>0.0281</td>
</tr>
</tbody>
</table>
CHAPTER 5
DISCUSSION

General trends observed from the normalized data, stepwise multiple regression and cross correlation analyses indicate that there was crosstalk in surface electrode data during the anterior tibialis, extensor digitorum and peroneal MMTs. Of the four muscles investigated, the peroneals appeared to be the major source of crosstalk. The peroneus brevis, as measured by WPB was the most consistent contributor to the variance seen in SPL data over the three different MMTs. The peroneus longus, as measured by WPL, accounted for a large amount (58.8%) of the variance seen in SPB data during the peroneal MMT. However, other muscles not included in this study are likely to have contributed to crosstalk in data from SPL and SPB as well. The study design used in the current study may have facilitated muscle activity of the gastrocnemius and soleus. Pressure from the plinth on the posterior calf during testing may have stimulated cocontraction from this muscle group even during MMTs that were antagonistic to the actions of the gastrocnemius and soleus. There are both central and peripheral control centers for both reciprocal inhibition and cocontraction (Basmajian & DeLuca 1985). These authors report increased cocontraction during maximal isometric contraction as well as when large forces are expected across a joint. Thus, during MVIC testing where the subject
practiced the procedure prior to manual muscle testing, as was the case in the current study, cocontraction rather than reciprocal inhibition may have been more dominant.

Also, posterior tibial muscles not tested in this study such as the gastrocnemius and soleus, may have contributed more to crosstalk when the peroneals were working at submaximal levels, i.e. anterior tibialis and extensor digitorum MMTs. Total variance in surface electrode data from both peroneus longus and peroneus brevis explained through stepwise multiple regression was much lower during the anterior tibialis MMT (0-34%) and the extensor digitorum MMT (42-69%), than during the peroneal MMT (69-73%). Thus, a greater portion of the variance was left unexplained during the extensor digitorum and anterior tibialis MMTs than during the peroneal MMT. The unexplained variance is most likely due to crosstalk from muscles not investigated in this study. In a similar study using regression analysis to investigate crosstalk in data from surface electrodes over muscles of the lower leg, Perry et al. (1981) noted a similar pattern. A greater portion of the variance was unaccounted for during MMTs for muscles other than the one targeted with the surface electrode. Perry et al. proposed that the unaccounted for variance in surface electrode data was likely the result of crosstalk from muscles not included in their study. In summary, the general trends noted in the results of this study indicate that surface electrodes over the peroneals are better at detecting muscle group
activity and are more accurate when that muscle is working relatively hard compared to the surrounding muscles.

While comparison of the data from surface versus fine wire electrodes demonstrated evidence of crosstalk, comparisons between data from the wire electrodes show evidence of cocontraction among the muscles investigated (Table 3). Although each muscle was most active during its own MMT, EMG activity was recorded by each fine wire electrode during all MMTs. While there is a possibility that this EMG activity could have been the result of crosstalk in the data from the fine wire electrodes, this is unlikely since there were no significant cross correlations between the data from the fine wire electrodes except between peroneus longus and peroneus brevis which may be explained by the fact that these two muscles generally perform as a group. Therefore, when considering the EMG data recorded by fine wire electrodes in muscles other than the one targeted by the MMT, cocontraction is suggested. During a MVIC, strong cocontraction from agonistic muscles, such as anterior tibialis during the extensor digitorum longus MMT was expected. Cocontraction was also suggested to a smaller degree in muscles considered to be antagonists, such as peroneals during anterior tibialis MMT (Table 3).

The normalized, averaged data from all subjects (Table 2) showed that the EMG activity (NU) from both WPB and WPL was higher than the activity of the corresponding surface electrodes during the peroneal MMT. However, during the anterior tibialis and the extensor digitorum MMTs, the EMG activity of each
surface peroneal electrode was higher than the activity of the corresponding wire electrodes. The increased surface versus fine wire EMG activity is likely the result of crosstalk. This finding is similar to what Perry et al. (1981) found when they investigated the selectivity of surface and fine wire electrodes in the gastrocnemius and soleus muscles. In their study, EMG activity recorded from the surface electrodes of the gastrocnemius and soleus was found to be higher than the corresponding wire electrodes during MMTs for other muscles. Also, EMG activity recorded from the fine wire electrodes of the gastrocnemius and the soleus was found to be higher than the corresponding surface electrodes during the gastroc-soleus MMT. Our findings support Perry et al.'s conclusion that crosstalk is a more significant issue when neighboring muscles are relatively more active than the target muscle.

The results of the cross correlation and the stepwise regression analyses for all 14 subjects suggest several implications. During the anterior tibialis MMT, SPL correlated significantly with WPB and did not significantly correlate with WPL. During this test, low levels (compared to MVIC) of activity were expected from the peroneals since they are antagonistic to the actions of anterior tibialis. At low or moderate levels of muscle activity, fewer motor units are activated simultaneously than during contractions at or near the muscle's maximum (Basmajian & DeLuca, 1985). Therefore, the amount of EMG activity read by each wire electrode could be highly variable depending on which motor units were activated. In looking at the average (NU) activity for all subjects (Table 2),
WPB (0.029415 NU) recorded almost twice the activity of WPL (0.016525 NU). The WBP value still represents minimal activity from peroneus brevis, relative to the maximum recorded during the peroneal MMT (Table 3). Also during the anterior tibialis MMT, SPL recorded over three times as much activity (0.046653 NU) as WPL. Thus, it is likely that SPL recorded activity from both peroneals and perhaps activity from other muscles as well. Since a general trend showed that both surface peroneal electrodes correlated significantly with the wire peroneal electrode with the highest NUs, SPL would be expected to correlate most with WPB. This may explain why in the regression analysis of SPL during the anterior tibialis MMT, the only variable that contributed significantly to the variance was WPB.

The asynchronous firing of fewer motor units seen in a muscle contracting at low levels, and the increased selectivity of fine wire electrodes may explain why WPL and WPB did not account significantly for any of the variance in SPB during the anterior tibialis MMT. Results of the regression analysis of SPB support the findings of the cross correlations in that there were no significant relationships found between SPB and any of the fine wire electrodes during the anterior tibialis MMT. The lack of relationships between SPB and any of the wire electrodes suggests that SPB may have recorded from many or all of the muscles included in this study as well as muscles of the lower leg. It is possible that due to a large variety of contributors to the data recorded, none of them showed a significant contribution in the cross correlation or regression analyses.
The stepwise regression analysis of SPL during anterior tibialis MMT showed that EMG activity from WPB accounted for 34% of the variance found in SPL (prob > F = 0.0284). This is evidence of crosstalk from peroneus brevis to SPL. Based on their regression equation, Perry et al. (1981) made similar conclusions regarding crosstalk from soleus to surface EMG over the gastrocnemius. However, findings from the present study are inconsistent with the results of Giroux and Lamontagne (1990). These authors reported that both finewire and surface electrodes recorded statistically similar signals in shoulder muscles of normal male adults under isometric and dynamic conditions. However, these researchers used a different normalization process (100% MVC) and a different statistical analysis (ANOVA). Giroux and Lamontagne questioned the appropriateness of their normalization procedure when investigating muscles of the shoulder and neck.

Sixty-five percent of the EMG activity of SPL was unaccounted for during the anterior tibialis MMT. This is a surprisingly high percentage. It was expected that WAT would have contributed significantly to SPL data because anterior tibialis is likely to be highly active during this test and peroneus longus is expected to be relatively inactive (Kendall et al., 1993). DeLuca and Merletti (1988) found that surface electrodes over the peroneus brevis and soleus recorded activity when the anterior tibialis was electrically stimulated supramaximally. The findings from the present study are inconsistent with DeLuca and Merletti's results. However, there is question as to whether DeLuca
and Merletti actually recorded crosstalk or whether crossover stimulation was the reason for the peroneus brevis surface EMG activity.

Although activity from muscles other than peroneus brevis did not contribute significantly to the recorded EMG of SPL during the anterior tibialis MMT, crosstalk from posterior calf muscles not included in this study could have been responsible for a significant amount of the unaccounted variance. Crosstalk from the gastrocnemius and soleus could be considered as a potential source of EMG signal contribution to the data collected from SPL. Both Perry et al. (1981) and Toft et al. (1991) have demonstrated that crosstalk from the soleus can contribute to data from a surface electrode over the gastrocnemius. These two large and powerful muscles are in close proximity to the peroneus longus and could produce high volumes of EMG signal while co-contracting during a maximal voluntary contraction of anterior tibialis. Cocontraction among antagonistic muscle groups has been documented (Basmajian & DeLuca, 1985). Thus, cocontraction of gastrocnemius and soleus during a maximal anterior tibialis MMT might be expected. These facts support the hypothesis that crosstalk from gastrocnemius and soleus may have contributed to the SPL EMG signal during the anterior tibialis MMT.

During the peroneal MMT, the wire electrodes in the peroneals each recorded more total activity (WPL=0.149874 NU, WPB=0.13457 NU) than did their corresponding surface electrodes (SPL=0.106395, and SPB=0.115923). SPL showed a stronger relationship with WPB than with WPL, and SPB showed a
more significant correlation with WPL than with WPB. Total activity (averaged
NU from all subjects) of WPL was greater than WPB while total activity in SPB
was greater than SPL. Therefore, total activity seemed to influence cross
correlations. Additionally, stepwise regression analysis for the peroneal MMT
showed that WPB accounted for more of the variance in SPL, while WPL
accounted for more of the variance in SPB. WPB explained 55% of the variance
in the SPL data, WPL explained only 13% of the variance. A similar pattern was
seen with the SPB and WPL: WPL accounted for 58% of the variance in data
from the SPB electrode, and WPB accounted for only 14% of the variance.

An explanation for the above mentioned findings could be due to the
similar and synergistic actions of the peroneus longus and peroneus brevis.
Since these muscles have identical actions, one might expect similar output
during a peroneal MMT. In fact, WPL and WPB correlated significantly (r=
0.47633). The high level of intercorrelation between WPL and WPB may have
been reflected in the results of the regression analysis. The stepwise regression
utilized in this study takes into consideration intercorrelation between
independent variables. Thus, it prevents an erroneous additive effect from the
two intercorrelated independent variables (Portney & Watkins 1993). While
stepwise regression insures that no more than the total combined variance of two
interrelated independent variables is accounted for, it is not possible to
accurately determine the exact contribution of either related independent
variable. Thus, while it is likely that SPB recorded muscle activity (crosstalk)
from the peroneus longus and that SPL recorded crosstalk from peroneus brevis during the peroneal MMT, the methodology used in this study did not allow for differentiation of the exact amount of crosstalk that each muscle contributed. In addition, since only 68% of total SPL variance and 72% of total SPB variance is accounted for, other muscles of the lower leg, which were not investigated, could have contributed crosstalk to data in SPB and SPL.

The stepwise regression analysis of SPB during the extensor digitorum MMT for all 14 subjects shows that WPB accounted for 42% of the total variance in the data. WPB was the only electrode that significantly contributed to the variance in SPB. Cross correlations support the regression analysis in that WPB correlated highest with SPB (r=0.65003). WPL also correlated highly with SPB (r=0.60295). Again, the strong intercorrelation between WPB and WPL may account for WPL data not making a contribution to the regression equation of SPB data. Likewise, during the extensor digitorum MMT, the EMG signal from WPB accounted for 47% of the total variance in SPL. This coincides with a high cross correlation between SPL and WPB (r=0.68830). The high intercorrelation between WPL and WPB should again be noted (r=0.59015).

WAT accounted for 22% of the variance in SPL during the extensor digitorum MMT. Crosstalk from anterior tibialis is likely because of the following: 1) during the extensor digitorum MMT, the peroneals may cocontract and anterior tibialis is activated to assist with dorsiflexion; 2) anterior tibialis and peroneus longus share a common fascial bond at their origin and are in close
proximity to each other. Perry et al. (1981) stated that "when antagonistic muscles are adjacent (such as the tibialis anterior and the peroneus longus), surface electrodes can produce erroneous information" (p. 14). However, the cross correlation results showed weaker correlation between SPL and WAT ($r=0.50711$) than between SPL and WPL ($r=0.56751$) during the extensor digitorum MMT. Once again, it appears that intercorrelation between independent variables makes it difficult to determine the exact contribution of WPL and WPB data to the variance seen in SPL data. Because there was a high correlation between the independent variables of WPB and WPL ($r=0.59015$), stepwise regression determined that WAT accounted for more of the variance than WPL during the extensor digitorum MMT.

In the regression analysis excluding the two outliers, cross correlation analysis showed a stronger correlation between SPL and WPL than between SPL and WAT during the extensor digitorum MMT. In fact, the correlation between SPL and WAT ($r=0.43722$) did not meet the $p \leq 0.10$ significance level. Therefore, results of the 12 subject regression analysis that showed WAT accounting for 23% of the variance in SPL data is suspect and may be the result of statistical artifact.

Regression analysis excluding the two outliers, also showed that WPB accounted for 40% of the variance and WED accounted for 26% of the variance in SPB data during the extensor digitorum MMT. Cross correlation results supported the regression analysis in that both WPB and WED significantly correlated with SPB. Because WED appeared to have migrated in the two
subjects who were considered outliers, the data analysis excluding these outliers is considered more accurate than the 14 subject analysis. This may be most significant when comparing results of the extensor digitorum MMT for the two groups. WED accounted for 26% of the variance in the 12-subject regression, but did not appear in the results of the 14-subject analysis. Crosstalk from extensor digitorum is the most likely explanation for the 26% variance in SPB accounted for by WED, in the 12-subject analysis for the extensor digitorum MMT. Excluding the two outliers, it was also found that WPL correlated significantly with SPB ($r=0.51962$), but WPL’s contribution to the regression analysis of SPB was not significant. Again, this may have been due to the high intercorrelation between the independent variables of WPL and WPB.

Limitations

This descriptive study was designed to determine the best type of electrode to use in EMG analysis of peroneus longus and peroneus brevis muscles. In general, descriptive studies are limited in that they do not allow for the establishment of cause and effect relationships. The general purpose of a descriptive study is to describe phenomena as it exists in order to allow for better decision making, to generate hypotheses and to define the need for further and more rigorous research (Portney & Watkins, 1993). The results of this study may be used to predict whether, when performing EMG analysis on a population similar to the one in this study, the use of surface electrodes over the peroneal muscles is likely to include data from other muscles.
Several variables were not controlled for in this study, including gender, age or general strength. To date, no research has been published demonstrating that any of these factors have had an effect on EMG electrode specificity. In addition, the amount of adipose tissue over the target muscle also was not controlled for. This has been shown to have a deleterious effect on the specificity of surface electrodes (Solomonow et al., 1994). It is common knowledge that women in general have a thicker layer of subcutaneous fat than men, and that the ratio of fat to muscle increases with aging. Thus, age and gender may have had an indirect effect on the specificity of surface electrodes.

The small sample size in this study is another limitation. As sample size decreases, it becomes increasingly difficult to show statistical significance in the data while maintaining an acceptable alpha level. It also limits generalizing any conclusions drawn from this study beyond the exact sample studied.

The validity of using fine wire data as the index of actual EMG activity is still debated in the literature (Perry, 1981; Basmajian & DeLuca, 1985; Kadaba, 1985; DeLuca & Merletti, 1988; Winter, 1990; Perry, 1992). Variations in the characteristics of data collected from fine wire versus surface electrodes as well as the possibility that crosstalk may also occur when using fine wire electrodes are issues of debate. However, the use of normalization may account for these characteristic differences (Perry, 1992; Perry et al, 1981; Knutson et al, 1993). Additionally, many respected researchers in the field doubt the existence of, or at least the significance of, crosstalk from wire electrodes (De Luca & Merletti, 1988;
Perry, 1992; Toft, et al. 1991; Turker 1993; Solomonow 1994). Solomonow reported small amounts (up to 3.77% MAV) of "crosstalk" from intramuscular electrodes in cats. Generalization to humans may have limited value. Additionally, the study by Solomonow used high levels of electrical stimulation to induce the measured muscle activity which raises the question of whether the phenomenon described was actually overflow stimulation instead of crosstalk. Furthermore, when referring to intramuscular electromyography, Turker stated that "motor unit potentials are usually free from volume conduction and therefore reflect only the activity of the muscle in which the electrode is placed" (pg. 705, 1993).

Both Perry et al. (1981) and Toft et al. (1991) conducted studies that measured crosstalk from surface electrodes over muscles of the lower leg, in which EMG data were collected simultaneously from intramuscular electrodes in the same muscles. The data from the intramuscular electrodes were used as the index of actual muscle activity in both studies. Measures were taken to verify that the intramuscular electrodes were detecting only activity from the target muscle in both studies. Thus, it is the belief of the current researchers that the use of data from fine wire electrodes as the index of actual EMG activity is the best method available when studying crosstalk in human subjects.

The best method for normalization continues to be a topic of debate in the EMG literature. The optimal method most likely varies according to the study design and purpose (Turker, 1993). The process of normalizing data from one
electrode to the sum of all data collected from that electrode, has been used in at
least one other study (Perry et al, 1981). It was chosen as the normalization
procedure for this study due to the similarities between this study design and
Perry et al’s. However, the reliability and validity of this form of normalization
has not been thoroughly evaluated.

Sources of Error

Systematic sources of error inherent in the Elite hardware, Matlab or SAS
/STAT software, and fine wire and surface electrodes can not be fully controlled.
Unacceptable levels of apparent artifact appeared in data from the fine wire
electrodes during gait tests, but not during the manual muscle tests. Therefore,
this study was modified to eliminate the gait tests. Strict adherence to fine wire
electrode assembly protocol was followed to minimize variations in quality of
these electrodes. Crosstalk in surface electrodes has been shown to increase with
increasing electrode size (Basmajian & DeLuca, 1985). Thus, the smallest size
surface electrodes available (8 mm) were used.

Random error was introduced by variations in wire and surface electrode
placement between subjects. Crosstalk has been shown to increase with
increasing center-to-center distance between the two discs of the bipolar surface
electrode (Basmajian & DeLuca, 1985). Thus the smallest discs available were
placed as close together as possible (10-15 mm center-to-center spacing) to
minimize error. The experience and skill level of the clinician may effect proper
electrode placement. Determination of bony landmarks, which is subjective
depends on reliable palpation skills. Use of a single researcher to place all
surface electrodes and another to insert all fine wire electrodes minimized this
error. Additionally, electrode application protocols were developed by
recommendations from published sources, and followed closely with one
exception. The electrode placement protocols were chosen according to the
recommendations of Perotto et al. (1994) and Soderberg (1992). The fine wire
electrode placement for extensor digitorum longus was modified slightly due to
inaccurate electrode placement when following Perotto's protocol during the
pilot study. Further attempts to control for error in fine wire electrode
placement included verification of placement with an oscilloscope, electrical
stimulation, and visual inspection of the raw EMG data. All trials were taken
during a single test session to eliminate inter-trial error. However, despite
attempts to control for consistent placement of fine wire electrodes, variation in
the terminal location of the wires between subjects could not be eliminated.

Another source of random error could have been inaccurate data due to
migration of the fine wire electrode after insertion and verification by
oscilloscope and electrical stimulation. This problem was confirmed in one case
when upon visual inspection of the raw data, migration of the fine wire electrode
in the extensor digitorum to the peroneus longus was suspected. In this case,
electrical stimulation was reapplied to the electrode in question and the
suspicion was confirmed. As a result, this subject's data were excluded.

Additionally, electrodes from extensor digitorum in the two outliers in this study
appeared to have migrated or been misplaced. However, in these two subjects the possible error was not caught until data processing was complete. Therefore, they were not eliminated from the study, but statistical analysis was performed both with and without the outliers. The two outliers in this study raise the question of migration of fine wire electrodes during ambulation and testing. In light of the possibility of electrode migration during testing, it may be wise to confirm that the electrodes are still in the target muscle at the conclusion of testing by re-applying electrical stimulation.

Variations in manual muscle test techniques and limb positioning due to the examiner's skill level may also have been a source of error. The current researchers felt that small variations in maximal voluntary contractions that may have resulted from these errors would not have significantly effected the measurement of crosstalk. However, if maximal isometric contractions were not actually achieved due to technical error on the part of the examiner, it would limit the generalizability of this study. General statements regarding the likelihood or significance of crosstalk during a maximal versus submaximal contraction in the target muscle and/or neighboring muscles would be less valid. Therefore, efforts were made to minimize such error by following Kendall et al.'s (1993) protocols for performing the manual muscle tests, as well as having an observer verify proper limb positioning during testing.
Implications for Future Research

There is a need for further research in the area of normalization procedures when comparing data from finewire and surface electrodes. Validity and reliability of the normalization method used in this study needs to be further established in human subjects. Results of this study suggest the possibility of crosstalk from gastrocnemius and/or soleus in the data collected by surface electrodes over peroneus brevis and/or peroneus longus. This needs to be further investigated. In addition, the two outliers noted in this study illustrate the possibility of fine wire electrode migration during dynamic activity. While Basmajian and DeLuca (1985) state that this is seldom a problem once the barbs are firmly set in the muscle, the outliers in the current study suggest further investigation may be warranted. Likewise, results of our pilot study demonstrate a need for the development of a standardized protocol when using fine wire electrodes in gait analysis. Furthermore, there is a need to repeat this study as it was originally intended, with a gait test included, and with a larger sample size. Finally, in order for the intention of this study to be fully clinically applicable, it needs to be replicated with the gait test included, on a population of children with cerebral palsy.

Clinical Implications

The results of this study are beneficial to the CHKS, and for gait laboratories and clinicians that use EMG for diagnostic purposes.
Although this study demonstrated strong evidence of crosstalk when using surface electrodes over the peroneus longus and peroneus brevis, the study has limitations which suggest that generalization of these results must be made with caution. Further research is necessary before it can be said with certainty that crosstalk is a clinically significant problem when using surface electrodes over peroneal muscles. Furthermore, the fact that muscles do not work maximally during gait suggests caution in generalizing of the results of this study. The possibility of crosstalk from antagonistic muscle cocontractions during a maximal voluntary contraction may not be a significant factor when using EMG analysis during gait. However, Falconer and Winter (as cited in Redfern, 1992) developed an isometric model which estimated the cocontraction between soleus and anterior tibialis during gait. Additionally, the results of this study suggest that crosstalk may be more significant when the muscle under the surface electrode is working at less than its maximum, as is generally the case during gait. During gait, the reciprocal activation of antagonistic muscle groups could result in crosstalk from antagonistic muscles skewing the onset and cessation evaluations of muscles targeted by surface electrodes. Although further research is needed, this study does add to the body of evidence that indicates that surface electrodes may not be the best electrode choice when seeking specific EMG information related to onset, cessation, duration, or activation levels of muscles of the lower leg.
Conclusion

The current study has demonstrated crosstalk in data collected by surface electrodes over the peroneus longus and peroneus brevis. It also demonstrated cocontraction of antagonistic muscles during maximal voluntary contractions. During MMTs where the peroneals were working submaximally, there was more evidence of crosstalk (in data from surface electrodes) from neighboring muscles, than when the peroneals were working at or near their maximum. Furthermore, the results of this study indicate a high probability of crosstalk from agonistic muscles, but muscles other than agonists may not be as significant a factor when the muscle under the surface electrode is working near its maximum. The results of this study also suggest that other muscles of the calf which were not included in the current study, may contribute to crosstalk read by surface electrodes over the peroneals. Our findings support Perry et al.’s (1981) conclusions that surface electrodes are best used to represent muscle group activity. While further research is needed, the results of this study suggest that when specific information regarding timing, onset, and/or duration of peroneus longus or peroneus brevis activity is desired, surface electrodes are likely to include unacceptable amounts of inaccurate data.
REFERENCES


APPENDIX A
CENTER FOR HUMAN KINETIC STUDIES
MEDICAL HISTORY

DATE: __________
PATIENT NAME: ____________________________ DOB ______
AGE _____ SEX _____ WEIGHT ______

ANSWER THE FOLLOWING QUESTIONS AND EXPLAIN ANY YES ANSWERS:

Is there any family history of:
Bleeding disorders .......................................................... Yes ___ No ___

Have you ever experienced any of the following? - If yes, explain below

1. Sprains/strains to lower extremities ...................................... Yes ___ No ___
2. Fractures/ broken bones of lower extremities ...................... Yes ___ No ___
3. X-rays, sonograms, computed tomography (CT) scans, bone scans
   magnetic resonance imaging (MRI) done in the past year? ..........Yes ___ No ___
4. Any injuries within the past six months which required medical attention/ caused difficulty walking for over 24 hours? ........Yes ___ No ___
5. Any pain at the present time? ............................................. Yes ___ No ___
6. Are you taking any prescriptions or over-the-counter medications?
   .................................................................................................Yes ___ No ___
7. Do you have a history of neurologic disorders? ......................Yes ___ No ___

List all surgical operations (involving the lower extremities) and explain yes answers:

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---------------------------------------------------------------------------------------------------------------------------------------
---------------------------------------------------------------------------------------------------------------------------------------
APPENDIX B
Clinical Examination

Subject’s Initials _____________ Date ___________

Lower Extremity Alignment: (note any moderate to severe abnormalities)

Observational Gait Analysis:
1. Equal stride length ____ Normal ____ Abnormal
2. Trendelenburg ____ Normal ____ Abnormal
3. Foot slap ____ Normal ____ Abnormal
4. Other gait deviations_____________________________________

Lower Extremity Screen:
Functional tests:
Squat ____ Normal ____ Abnormal
Toe raises (S1 & S2) ____ Normal ____ Abnormal
Heal walking (L4) ____ Normal ____ Abnormal

Strength: (score according to Kendall’s MMT)
- Hip flexors (L1 and L2) ____ Right ____ Left
- Knee extensors (L3) ____ Right ____ Left
- Great toe extensors (Ls) ____ Right ____ Left
- Foot evertors (Ls & Si) ____ Right ____ Left

ROM: (mark WNL or (U) for not within normal limits)
- Hip ____ Right ____ Left
- Knee ____ Right ____ Left
- Ankle ____ Right ____ Left

Sensation:
(L1 - Si) ____ Right ____ Left

Examiners Signature____________________________________ Date___________
APPENDIX C
Clinical Examination Parameters and Exclusion Criteria

**Bleeding Disorders:** Volunteers must have no history of bleeding disorders. This will be screened for with the medical history form.

**Neurologic Disorders:** Volunteers must have no history of neurologic disorders. This screened for with the medical history form and physical exam.

**Medications:** Volunteers will be excluded from the study if they report on the history form that they are currently or within the past three months have been on an anticoagulant medication.

**Orthopedic Injury:** Volunteers will be excluded from the study if they have a past or present orthopedic injury which has resulted in current use of braces or assistive gait device, or if gait abnormalities are discovered during observational gait analysis.

**Lower Extremity Alignment:** The examiner is looking for moderate to severe deviations from normal alignment. Volunteers will be excluded from the study if such alignment results in abnormal gait pattern as noted in the observational gait analysis.

**Observational Gait Analysis:** Volunteers will walk barefoot at a self determined pace. The examiner looks for Trendelenburg gait, footslap, equal stride length and weightbearing time for each extremity. Volunteers will be excluded from the study if any gait deviations are noted.

**Functional Tests:** Volunteers must be able to perform the following tests in order to be eligible to participate in this study.

- **Squat:** Will be performed with patient using examiner’s arm for stabilization only. Heels must remain on the floor throughout the squat.
- **Toe Raises:** Volunteer will be able to rise 10 consecutive times on his or her toes one foot at a time, using the examiner’s arms for stabilization only.
- **Heel Walking:** The volunteer must be able to walk 10 consecutive steps on his or her heels.

**Manual Muscle Tests:** Volunteers must score 5/5 on all manual muscle tests listed on the clinical examination form as specified by Kendall.
Range of Motion: Volunteers must demonstrate normal range of motion of all joints of both lower extremities according to specifications of Norkin and White (1995). The observational method of measurement will be used and any suspected joint limitations will be measured with a goniometer.

Sensation: All volunteers must demonstrate normal sensation in all dermatomes of both lower extremities. This will be tested with light touch with the volunteer’s eyes closed. A normal test will involve the volunteer successfully identifying the correct area being touched 100 percent of the tests.
CROSSTALK: SURFACE VERSUS INTRAMUSCULAR ELECTRODES FOR THE PERONEUS BREVIS AND PERONEUS LONGUS

I _____________________________, voluntarily agree to participate in the research project under the direction of Tanya Cardillo, SPT, Sue Dresden, SPT, and Jacque Solem, SPT, to be conducted at the Center For Human Kinetic Studies. I understand the following to be true:

1. The study is being conducted in order to determine the best type of electrode to use with electromyography when considering surgical procedures involving muscles of the calf.

2. I have been selected for this study because I do not have a current ankle, knee, hip or low back injury, neurological disorder, or bleeding disorder.

3. I understand that my decision to participate is voluntary and that I can withdraw at any time. I have been assured that withdrawal or nonparticipation would not result in any change in the care that I would receive at the Center For Human Kinetic Studies or Mary Free Bed Hospital.

4. The researchers will ask about my past medical history and perform a physical therapy evaluation on me. For these evaluation and testing procedures I will be wearing shorts in order to expose the skin for electrode placement. My estimated time commitment for this study will be 2 hours.

5. Muscle activity will be recorded with surface and fine wire electrodes. Four surface electrodes will be placed on the outside of my lower calf. The skin surface under the electrodes will be shaved and vigorously rubbed with alcohol to assure good contact. A student physical therapist will be applying the surface electrodes and will be under the supervision of a licensed physical therapist. Sterilized, disposable wire electrodes will be inserted directly into four different muscles of the calf. A licensed physical therapist trained in fine wire insertion will insert all fine wires.

6. The use of fine wire electrodes is an invasive procedure and I may experience some mild local bleeding, and pain from the insertion of the wire. Universal precautions will be followed to protect me. The associated risks with this fine
wire procedure are minimal and very rare. These risks include infection and wire breakage. In the unlikely event of a minor injury, first aid will be provided. If necessary, further medical care will continue under the direction of my physician in accordance with my own particular financial arrangement.

7. The Center for Human Kinetic Studies (CHKS) will have custody of conclusions drawn from this study and it will only be used for the purpose of analysis, education and/or reporting scientific results. I understand that my records will be kept strictly confidential, as explained to and understood by me.

PARTICIPANT STATEMENT:

The test has been explained to me and I consent to participate. I have had the opportunity to ask questions.

_________________________  ______________________
Signature of Participant      Date

I wish to receive project results:

_________________________  ______________________
Signature of Participant      Date

INVESTIGATORS STATEMENT:

I have offered an opportunity for further explanation of this test.

_________________________  ______________________
Signature of Researcher   Date

_________________________  ______________________
Signature of Researcher   Date

_________________________  ______________________
Signature of Researcher   Date

For additional questions concerning Human Subjects Review Committee policies and procedures, please contact Professor Huizinga at (616) 895-2472.

Tanya Cardillo, SPT, Sue Dresden, SPT, Jacquelyn Solem, SPT (616) 459-4291 (H), (616) 954-2318 (W)
# APPENDIX E
Data Collection Sheet

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APPENDIX F

Normalized Units For 14 Subjects
(average of five trials)

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