5.1 EMG Based Prosthetic Control

With the progress of physiological science, computer and sensor technology, the research on Above Knee (AK) prosthesis has gradually become the focus of the rehabilitation engineering. The human desire to control the prosthetic device using physiological signal i.e. muscle activity is no longer a wishful thinking. Active research is being carried out to design adaptive product that can be controlled through physiological signals from body and brain. In an AK device, the rate of walking is decided by the muscle power generated in an amputated segment and transferred to the prosthetic device. Prosthetics must adapt itself to the signals sensed from the day to day body response to make the device adaptive with growing age and changing behavioural aspects of patients.

Innovative research in this area is reported regarding prosthetic devices controlled by using the Electromyography (EMG) signals. There are varieties of commercially available EMG controlled devices; they are popularly known as myo-electric prosthetic knee. EMG is spontaneous and voluntary signal that reflects the current state of body. There are various factors which have to be taken into account for using EMG as a tool for prosthetic control. This chapter discusses them in detail. Following issues have been focused:-

- Various parameters of EMG that can be used as control parameters.
- Method of EMG acquisition, setup, hardware and analysis
- Development of a prosthetic control strategy in combination with other sensors

There are several characteristic of EMG signal for controlling the prosthetics. EMG for control is required to be acquired from different muscles. EMG signals vary with location of muscles. Identification of each muscle is done from their anatomical function. With surface electrodes, we address bulk of muscle signals and also there is signal deterioration due to muscle atrophy, cross talk, engineering and environment factors. As the EMG signal varies from the muscle...
function, it is important to identify the proper muscle which gives good signal characteristic. In case of lower limb prosthesis, the EMG signal is taken from the residual limb of the amputee. This EMG signal can be analysed into time domain and frequency domain. There are few parameters which are used in controlling the myo-electric devices. Different signals have different amplitudes and patterns which are directly related to muscle activity and decomposing these parameters helps to generate control parameters\cite{118, 199}. It has been reported that there exists a relationship between various gait parameters (knee and hip flexion angles) and the EMG signals from the associated muscles\cite{120, 121}. The gait angle can be predicted by using a surface EMG signal on a normal lower limb, and then, based on this angle and the corresponding EMG pattern, the gait posture can be predicted\cite{122, 123}.

After analyzing EMG in relation to gait parameters, a knowledgebase can be generated which can act as standard database. A control strategy based on knowledgebase generated using both SEMG and knee flexion angle (KFA) for controlling the developed knee joint has been proposed by researchers\cite{124}.

It is a difficult task to acquire a repeatable and reproducible signal from residual limb\cite{125, 126}. Muscle decay and atrophy take place with the passage of time faster on the amputated limb due to its limited activity. Non availability of full limb segment adds to challenges of proper placement of electrodes. For example, walking is best described by few muscles (gluteus and Biceps femoris) in the lower leg which do not exist for a transfemoral amputee. To circumvent these limitations, we have adopted a new strategy of placing EMG electrodes on the non-amputed limb of the unilateral AK amputee. Sensor to measure knee flexion angle (KFA) i.e. electrogoniometer is attached to the prosthetic knee joint.

This hybrid mapping of knee flexion angle with EMG acquired from normal limb is used to generate a knowledgebase on the basis of relationship between the activities of various muscles during the different phases of a Gait cycle. The EMG signal parameters like RMS amplitude and mean and median frequencies also share a relationship with the speed of walk\cite{128, 129}.
5.1.1 EMG and its Applications

As we know EMG is the analysis of the electrical activity of the skeletal muscles. The amplitude of EMG signal range between less than 50 µV and up to 20 to 30 mV with repetition rate of muscle unit firing at about 7–20 Hz, depending on the muscle under observation. EMG activity results from a voluntary or involuntary contraction. Voluntary muscle contraction is controlled by the central nervous system. It occurs due to the conscious effort originating in the brain, which sends signals in the form of action potentials, through the nervous system to the motor neuron that innervates several muscle fibers. Involuntary muscle contraction is a result of non-conscious brain activity or stimuli proceeding in the body to the muscle itself such as the pumping of heart.

The uses of EMG are categorized into: clinical or diagnostic EMG and kinesiological EMG or (kinesiology). EMG is used as non invasive diagnostic technique for diagnosis of different diseases. Kinesiology—the study of EMG patterns, establishes, the linkages between the muscular function and movement of the body segments and involves evaluation of timing of muscle activity with regard to the movements\(^{[110]}\). Increase in EMG amplitude is directly related to force produced in muscle due to isometric contractions. From this viewpoint, EMG has been related to various biomechanical variables. EMG amplitude does not exactly match the build-up of isometric tension and there is a time difference between these two. Precautions are required, when we are using the surface EMG for controlling the device based on amplitude which depends on the force produced by muscle. It becomes an unwieldy variable when many muscles are crossing the same joint or when muscles cross multiple joints. Kinesiological EMG is established as an evaluation tool for applied research, physiotherapy/rehabilitation, sports training and interactions of the human body to industrial products and work conditions which are depicted in Fig. 5.1.
Surface Electromyography is the technique for measuring levels of muscle activity using electrode placed on skin surface. It utilizes technology very similar to the ECG, but with significantly greater sensitivity. There are two types of SEMG: Static Scanning SEMG and Dynamic SEMG. With Dynamic SEMG, electrodes are attached to the skin, and muscle activity is measured and graphed as the patient moves through various stages of motion. EMG captured during gait is dynamic EMG and hence of importance to us. We are interested in measuring the electrical activity of muscles of lower limb for prosthetic control so we are more interested in Dynamic EMG as static is not feasible in our case\textsuperscript{[131]}.

**Electrode Types** used for EMG signal acquisition is surface electrodes or fine wire electrodes for both diagnostic and kinesiological EMG applications. Fine wire electrodes give the highly faithful reproduction of EMG signal but this process is invasive and thus not used much. Surface electrodes are the most widely used electrodes in EMG signal acquisition. They are categorized as active electrodes and passive electrodes. The basic difference between these two is the site of amplification of the acquired signal. In passive electrode, the EMG signal acquisition and amplification is done separately. Here skin resistance and tissue impedance and physiological cross talk are the major challenges, minimized by applying conducting gels and doing extensive
skin preparation. The passive electrodes are, although cost effective, but noise prone offering poor SNR.

Active electrodes have built-in amplifiers with high input impedance providing high signal to noise ratio at the electrode site (no gel is required). Surface electrodes being non invasive are painless. In addition they are more reproducible, easy to apply, and very good for recording dynamic EMG applications. Due to being larger in dimension, they have a large pick-up area and are more vulnerable to acquire cross talk potential from adjacent muscles.

5.1.2 EMG Features
To extract control parameter from the acquired EMG, Signal is analysed in time domain and frequency domain. The time feature root mean square values whereas frequency domain features include mean frequency and median frequency\cite{132,133}.

5.1.2.1 Time Domain Analysis
The amplitude of the EMG signal at any instant is proportional to the force exerted by the underlying muscle. This relationship can be easily appreciated by simply viewing the EMG signals in real-time while the intensity of the muscular contraction is increased. The root mean-square (RMS) is considered to be the most meaningful, since it gives a measure of the power of the signal.

**Root Mean Square (RMS)**
Mathematical derivations of the time and force dependent parameters indicate that the RMS value provides a rigorous measure of the information content of the signal because it measures the energy of the signal. Its use in electromyography, however, has been sparse in the past. The recent increase in its usage is due possibly to the availability of analog chips that perform the RMS operation directly. This parameter is recommended above the others for EMG analysis. It is useful in 1) assessing muscular force 2) estimating amplitude of signals and in comparing EMG activity. We have categorized RMS values for human gait derived into slow, normal and fast walking speed. Figure 5.2 shows a raw EMG signal and its RMS output.
The time-varying RMS value is obtained by performing the operations described by the term in reverse order given by equation 5.1

\[
\text{rms}(m(t)) = \left( \frac{1}{T} \int_{t_1}^{t_2} [f(t)]^2 dt \right)^{1/2}
\]

5.1.2.2 Frequency Domain Analysis

Analysis of the EMG signal in the frequency domain involves measurements and computation of parameters that describe specific aspects of the frequency spectrum of the signal. Fast Fourier transform of acquired signal is used for obtaining the power density spectrum of the signal. We have used median frequency and the mean frequency parameters of the power density spectrum for useful assessment. Median frequency is helpful in making fatigue assessments and in tracking spectral variation.

The median frequency is given by equation 5.2:

\[
\int_{f_0}^{f_{med}} S_m(f) df = \int_{f_{med}}^{f_0} S_m(f) df
\]
The mean frequency is defined as equation 5.3:

$$f_{\text{mean}} = \frac{\int_0^f f S_m(f) df}{\int_0^f S_m(f) df}$$  \hspace{1cm} (5.3)$$

where $S_m(f)$ is the power density spectrum of the EMG signal.

### 5.1.3 Muscle Sites and Activity during Gait

The location of the sensor on the muscle renders dramatically different SEMG signal characteristics. The muscle map is shown in Fig. 5.3 frontal view & Fig. 5.4 dorsal view gives a selection of muscles that typically have been investigated in kinesiological studies. The two yellow dots of the surface muscle indicate the orientation of the electrode pair in relation to the muscle fiber direction. The active electrodes were fixed to the motor points of the desired muscles of both the legs of the subject. The ground strap of the EMG setup was worn on the left leg.

![Anatomical positions of selected electrode sites - frontal view](image_url)

Fig. 5.3: Anatomical positions of selected electrode sites – frontal view
Out of many muscles which are involved in achieving walking function, we have identified few motor points based on their gait value.

**Tibialis Anterior (TA):**

In human anatomy, the tibialis anterior is a muscle that originates in the upper two-thirds of the lateral surface of the tibia and inserts into the medial cuneiform and first metatarsal bones of the foot. The tibialis anterior muscle is the most medial muscle of the anterior compartment of the leg. It functions to stabilize the ankle as the foot hits the ground during the contact phase of walking (eccentric contraction) and acts later to pull the foot clear of the ground during the swing phase (concentric contraction). It also functions to 'lock' the ankle, as in toe-kicking a ball, when held in an isometric contraction.

**Gastrocnemius (GA):**

The Gastrocnemius is located with the Soleus in the superficial posterior compartment of the leg. It is involved in standing, walking, running and jumping. Along with the Soleus muscle, it forms the calf muscle. It plantar flexes foot and flexes knee.
Soleus (SO):
The Soleus is located in the superficial posterior compartment of the leg. The action of the calf muscles, including the soleus, is plantar flexion of the foot (that is, they increase the angle between the foot and the leg). They are powerful muscles and are vital in walking, running and dancing. The Soleus specifically plays an important role in standing; if not for its constant pull, the body would fall forward.

Vastus Lateralis (VL):
The Vastus lateralis (Vastus externus) is the largest part of the Quadriceps femoris. It arises by a broad aponeurosis, which is attached to the upper part of the intertrochanteric line, to the anterior and inferior borders of the greater trochanter, to the lateral lip of the gluteal tuberosity, and to the upper half of the lateral lip of the linea aspera; this aponeurosis covers the upper three-fourths of the muscle, and from its deep surface many fibers take origin. Its main function is the extension of the knee.

Biceps Femoris (BF):
The Biceps Femoris is one of the hamstring muscles at the back of the thigh. It originates in two places: the ischium (lower, rear portion of the pelvis, or hipbone) and the back of the femur (thighbone). The fibers of these two origins join and are attached at the head of the fibula and tibia, the bones of the lower leg. The long head of the Biceps Femoris extends the hip at the beginning of walk; both short and long heads flex the knee and laterally (outwardly) rotate the lower leg when the knee is bent.

Gluteus Maximus (GM)
The gluteus maximus (or gluteus maximus) is the largest and most superficial of the three gluteal muscles. It makes up a large portion of the shape and appearance of the buttocks. It is a broad and thick fleshy mass of a quadrilateral shape, and forms the prominence of the nates. Its large size is one of the most characteristic features of the muscular system in humans, connected as it is with the power of maintaining the trunk in the erect posture. The muscle is remarkably coarse in structure, being made up of fasciculi lying parallel with one another and collected together into large bundles separated by fibrous septa.
5.2 SEMG Data Acquisition: Set up and Procedure

For generating the SEMG based control, EMG data was acquired using the available active surface electrode setup (M/s Biometrics Ltd). The data acquisition experiments were done to acquire the SEMG signals of the lower limb muscles and to analyze them for any patterns or trends followed during the normal course of walk.

**Alternatives of SEMG Acquisition Setup**

- Normal walk to ascertain the EMG parameters of different muscles is recorded. This will help us to find out which muscle is giving the optimum EMG output.

- Walking at three different self selected speeds of different individuals on level ground. This will help to grade the EMG according to walking speed. The EMG signal is analysed in all seven phases of gait cycle to know the amount of activity done by a particular muscle in each gait phase. As we have divided the developed prosthetic control into swing phase and stance phase control, therefore this analysis is very much required.

- The use of treadmill for analyzing SEMG is a matter of debate in literature. Some have termed treadmill acquisition more accurate than normal ground gait, therefore to verify this claim recording of EMG was done with subjects walking on treadmill at normal level and with 15 degree inclination. The EMG is analysed in time and frequency domain for all seven phases of gait.

- After analyzing EMG in all setup discussed above, SEMG is acquired from selective muscles in synchronization with electrogoniometer sensing the knee flexion angle. A knowledge base is generated which reproduces the range of EMG and knee flexion angle in each gait phase.

It has been observed that signals from different muscles measured with different alternatives, are similar and within acceptable limit which allows measurement from any alternative. The purpose of all these experiments is to analyze and establish patterns/trends for an individual or a group of individuals for repeated number of walks which can be generalized and can, in turn, be used as a control parameter for the designed prosthesis. The SEMG acquisition setup block diagram is
Strategy for SEMG Based Prosthetic Control

explained in Fig. 5.5a & 5.5b. The electrodes are attached to respective muscle sites with a medical grade tape and with a velcro strap to minimize movement artifacts which are generated due to shift in surface electrodes. The active electrode SX 230 sense the EMG signal and transfer it to computer through Biometrics acquisition unit which is an A/D converter with high gain amplifier. The EMG is stored into computer for further analysis. The stored EMG is analysed offline using Bioware (analysis software of Biometrics Ltd) and hardware independent EMG analysis software of Delsys Incorporation.

5.2.1 Hardware used for SEMG Signal Acquisition

The experiments were conducted using standard EMG system from Biometrics Ltd. The hardware details are as given below:

5.2.1.1 Sensor: Active SEMG Electrode

SX230 EMG sensor as shown in Fig. 5.6a was used, which has a pre amplifier attached to EMG electrode with an Input Impedance of > $10^7$ M Ohms. It is an integral dry and reusable type. The gain of this sensor is in the range of $10^5$ and bandwidth is in the range 10 Hz to 450 Hz. It requires a supply voltage ranging from 4.50 V dc to 5.0 V dc. The CMRR is greater than 96 dB (typically 110 dB) at 60 Hz. There is minimum requirement of skin preparation to have a good quality signal recorded, which is required for both static and dynamic EMG acquisitions.
For acquiring any physiological signal, it is required to have at least one neutral reference electrode per subject. Typically an electrically unaffected but nearby area is selected for positioning the reference electrode, such as joints, bony area, frontal head, tibia bone etc. We have used left leg as reference electrode site as shown in Fig. 5.6b. With active electrode, it is required to use only one reference electrode irrespective of the number of active electrodes used.

Fig. 5.6 (a): Active electrode SX230 (b) Reference Electrodes for EMG acquisition

5.2.1.2 EMG Preamplifier

The frequency range of a typical EMG signal extends from about 10 Hz to about 450 Hz. This includes the unwanted line-frequency of 50 Hz and the harmonics of this frequency. One of the most important design goals of the EMG surface pre-amplifier is to reject the considerable interference present on the skin resulting from mains pick-up.

The pre-amplifier used here has the following principal attributes:

- Differential electrodes to detect low level signals in a noisy environment.
- Very high input impedance (about $10^{15}$ $\Omega$) to minimize the differential pick-up due to mismatch in the skin contact resistance.
- A high-pass filter to remove DC offsets due to membrane potentials and to minimize low frequency interference caused by the pre-amplifier on the skin surface.
- A low-pass filter to remove the unwanted frequencies above 450 Hz. Fig. 5.7 shows the schematic of the surface EMG pre-amplifier.
5.2.2 SEMG Signal Acquisition

Correct acquisition of SEMG signals is an important and challenging task. Electrode placement on the motor joint and skin preparation are the preliminary tasks but their importance cannot be ignored. SEMG acquisition from a prepared skin reduces contact resistance and muscle cross talk. EMG signal, being random in nature, is difficult to analyze, with noise creeping in during acquisition adding to the analyzers agony. With correct site preparation and electrode placement, it is possible to have desired EMG acquired through surface electrodes. This involves removal of hair to improve adhesion of electrodes under humid condition or for sweaty skin types and/or under dynamic movement conditions.

5.2.2.1 Active Electrode Placement

The SEMG signal characteristic varies with the location of muscle and it is single most important factor for obtaining the best signal to noise ratio with the least amount of cross-talk. It is important to locate the fibers in the middle of the muscle which have a greater diameter than those at the edges of the muscle or near the origin of the tendons. The amplitude of action potential from the muscle fibers is proportional to the diameter of the fiber; the amplitude of the EMG signal will be greater in the middle of the muscle. Based on the muscle site and their function (Table 5.1), muscle sites are identified for our experiments. In the experimental setup, the EMG preamplifier fitted electrodes were fixed to the motor points of the desired muscles (Tibialis Anterior (TA), Gastrocnemius (GA), Soleus (SO), Vastus Lateralis (VL), Biceps...
Femoris (BF)) of both the legs of the subject using medical grade double sided tapes. The leads of all the four pre amplifiers were fed to the DataLINK base-unit DLK900 of the EMG acquisition system which acted as an interface to the PC.

Table 5.1: Selected muscles with their function

<table>
<thead>
<tr>
<th>Muscle Site</th>
<th>Function</th>
</tr>
</thead>
</table>
| Tibialis Anterior (TA) | ✓ Tibialis Anterior is responsible for dorsiflexion of the foot.  
✓ It also plays a crucial role in foot inversion. |
| Gastrocnemius (GA)   | ✓ Gastrocnemius plays a vital role in plantarflexion and knee flexion.                                                                   |
| Soleus (SO)          | ✓ Soleus finds use in plantarflexion.                                                                                                                                                               |
| Vastus Lateralis (VL)| ✓ Vastus Lateralis extends and stabilizes knee.                                                                                           |
| Biceps Femoris (BF)  | ✓ Biceps Femoris plays a crucial role in knee joint flexion and hip joint extension.  
✓ It laterally rotates knee joint when knee is flexed. |
| Gluteus Maximus (GM) | ✓ Gluteus Maximus is responsible for external rotation and extension of the hip joint.  
✓ It supports the extended knee through the iliotibial tract.  
✓ It is the chief antigravity muscle in sitting. |
| Quadriceps Femoris   | ✓ It plays a significant role in knee extension and hip flexion.                                                                          |

For having the acquaintance of experimental environment, subjects were instructed to walk a few rounds before actual data acquisition. As discussed, extensive experiments were carried out in setup and environment. The basic setup for all the experiments was kept the same. The results of different experiments have been discussed in the ‘results’ subhead.
5.3 Strategy for EMG Signal Analysis for Identification of Patterns and Control Parameters

For gait analysis, Dynamic electromyography (EMG) offers a means of directly tracking muscle activity. The intensity of muscle action recorded in terms of its electrical output is sufficient to indicate its mechanical effect. In order to have a complete gait analysis, it is required to analyze EMG is all its phases for duration and magnitude. The timing and intensity of the EMG during a phase or the entire gait cycle informs about neurological control and muscle integration. Amplitude of EMG signals during the dynamic gait is interpreted as a measure of force generated by muscle to achieve that walking speed. EMG is a random signal; for amplitude analysis, RMS envelop reflects the amount of EMG activity done. Due to complexity of the EMG record, its multi-spike, random amplitude quality defies simple interpretation. Little work is done to interpret the EMG signal in all phases of gait cycle during individuals walking at different speeds. There is no standard relationship available between EMG and muscular force. The following section describes the instrumentation, analysis and interpretation of EMG signal for different walking speeds of normal healthy individuals. The potential use of this analysis is in controlling the prosthetic device like artificial knee. The next section discusses the different alternative for SEMG acquisition setup.

5.3.1 Experiments for SEMG from Two Lower Limb Muscles of Both Legs

It is important to establish as to which muscle gives the best EMG output during the normal walk and to establish correlation between EMG of muscle of left leg and right leg. Experiments were conducted on subjects walking at self selected speeds of slow and fast with electrodes placed on selected muscles (Gastrocnemius and Soleus) as shown in Fig. 5.8. The collected data was stored and analyzed offline using Biometrics Management Software. The subject was instructed to walk a few rounds before actual data acquisition to get familiar with walking with the setup intact.
5.3.2 Experimental Setup for Integrated Sensor Data Acquisition of EMG and Electrogoniometer on Ground and Treadmill in All Gait Phases

After doing EMG experiments as explained in section 5.2.2, we selected four major muscle groups which are responsible for walking. Electrodes are applied to Vastus Lateralis (VL) (Fig. 5.9a), Gastrocnemius (GA), Tibialis Anterior (TA) (Fig. 5.9b) and Bicep Femoris (BF) (Fig. 5.9c) muscles of the subject. As we have to generate a control strategy for artificial knee, we have also attached Biometrics electrogoniometer to knee and hip location to measure knee flexion angle and hip flexion angle. To record the EMG in all gait phases, we have applied Biometrics force sensitive resistor (FSR) sensors to the heel and toe of the subject’s leg to determine events like heel strike and toe off and the phase of walk.
In this integrated sensor data acquisition scheme, synchronous data is obtained by EMG sensor from four muscle sites and by other sensors placed at respective locations to sense spatiotemporal gait parameter like KFA and Stance phase timings. Individuals were instructed to walk on level ground and then on treadmill. This helps to grade the EMG according to walking speed. These experiments are also able to make a comparison between level ground and level treadmill walking. While walking on treadmill, data is acquired with and without inclination of treadmill. The walking speed is categorized into slow, normal and fast. For ground walking, individuals were advised to walk with their self selected speed and for treadmill, walking speed is set as slow speed (2.4 m/s), normal speed (3.2 m/s) and fast speed (4 m/s). For experiments at treadmill with 15° inclination, data is acquired for slow and fast speed walks. For minimizing the effect of muscle fatigue and tiredness, the subjects were to walk for 20 seconds only which is sufficient as this period consists of multiple gait cycles in it.

**Software Tools for Analysis of Acquired EMG Data: Bioware**

At least three walk trials were recorded for each experiment at slow, normal and fast speed. All EMG data is stored in proprietary log file format of Biometrics. The stored data can be exported to other compatible formats for analysis by using different software-EMGworks analysis software (Delsys Inc.) also. The data is analysed in time (RMS) domain and frequency (mean and median) domain.

**Phased Analysis of Data**

Complete analysis of EMG during gait is done by first detecting the stance and swing phases. FSR sensor which is deployed as switch gives the heel strike and toe off information and thus, the phase of walk. As already discussed, the stance and swing phase is 60% and 40% of gait cycle respectively in terms of time interval. The two gait phases are subdivided into seven sub phases as tabulated in Table 5.2.
Table 5.2: Sub-phases of a gait cycle and their duration in terms of % gait cycle

<table>
<thead>
<tr>
<th>Sub phase</th>
<th>% Gait Cycle</th>
</tr>
</thead>
<tbody>
<tr>
<td>Loading Response</td>
<td>0-7.5</td>
</tr>
<tr>
<td>Mid Stance</td>
<td>7.5-32.5</td>
</tr>
<tr>
<td>Terminal Stance Swing</td>
<td>32.5-50</td>
</tr>
<tr>
<td>Pre Swing</td>
<td>50-57.5</td>
</tr>
<tr>
<td>Initial Swing</td>
<td>57.5-77</td>
</tr>
<tr>
<td>Mid Swing</td>
<td>77-85</td>
</tr>
<tr>
<td>Terminal Swing</td>
<td>85-100</td>
</tr>
</tbody>
</table>

5.4 Results

The results for the experiments conducted are discussed in the following two subheads:-

5.4.1 Results for SEMG from Two Lower Limb Muscles of Both Legs

The acquired SEMG data is analysed offline using Bioware. The signal is analysed for quantified amounts of RMS, median frequency and mean frequency. The pre-amplified SEMG signals for the Gastrocnemius muscle of left limb is shown in Fig. 5.10a and SEMG signal for Gastrocnemius muscle of right leg is shown in Fig. 5.10b. Similarly SEMG signal for Soleus left leg muscle is shown in Fig. 5.11a and Fig. 5.11b shows the SEMG signal for Soleus right muscle. RMS filter is applied on acquired SEMG signal. RMS filtered EMG signals from Gastrocnemius for left leg and right leg are shown in Fig. 5.12a & 5.12b respectively. Similarly Figs. 5.13a & 5.13b depict the RMS filtered signal from Soleus of both limbs. The frequency analysis is done and Median frequency component for both legs is calculated. Figure 5.14a & 5.14b show the median frequency filtered signal for Gastrocnemius for left leg and right leg respectively. Similarly 5.15a & 5.15b show the Median Frequency filtered EMG signals from Soleus for both limbs.
Fig. 5.10 (a) Acquired EMG signals from Gastrocnemius of left limb
(b) Acquired EMG signals from Gastrocnemius of right limb

Fig. 5.11 (a) Acquired EMG signals from Soleus of left limb
(b) Acquired EMG signals from Soleus of right limb
Strategy for SEMG Based Prosthetic Control

Fig. 5.12 (a) RMS Filtered EMG signals from Gastrocnemius of left limb 
(b) RMS Filtered EMG signals from Gastrocnemius of right limb

Fig. 5.13 (a) RMS Filtered EMG signals from Soleus of left limb 
(b) RMS Filtered EMG signals from Soleus of right limb
Strategy for SEMG Based Prosthetic Control

Fig. 5.14 (a) Median Frequency filtered EMG signals from Gastrocnemius of left limb
(b) Median Frequency filtered EMG signals from Gastrocnemius of right limb

Fig. 5.15 (a) Median Frequency filtered EMG signals from Soleus of left limb
(b) Median Frequency filtered EMG signals from Soleus of right limb
Quantified SEMG data for the RMS values of amplitude for Gastrocnemius and Soleus muscle of right and left leg is tabulated in Table 5.3. In the trace titles column, GRL and GLL refer to the signal obtained from Gastrocnemius Right Leg and Gastrocnemius Left Leg respectively. Similarly, trace titles SRL and SLL refer to the signal obtained from Soleus Right Leg and Soleus Left Leg respectively.

Table 5.3: Mean amplitude of EMG signal averaged over three trials for fast and slow walks

<table>
<thead>
<tr>
<th>Channel</th>
<th>Trace Title</th>
<th>Fast Walk (mV)</th>
<th>Slow Walk (mV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>GRL</td>
<td>0.044</td>
<td>0.03</td>
</tr>
<tr>
<td>2</td>
<td>SRL</td>
<td>0.055</td>
<td>0.03</td>
</tr>
<tr>
<td>3</td>
<td>GLL</td>
<td>0.066</td>
<td>0.05</td>
</tr>
<tr>
<td>4</td>
<td>SLL</td>
<td>0.058</td>
<td>0.04</td>
</tr>
</tbody>
</table>

It can be observed from the table that the RMS value of the signal amplitude tends to increase from slow to fast walk for both muscles of both the limbs. This can be attributed to the increase in motor activity with increase in speed of movement. Another notable observation is that the variation in amplitude of the signal for the same muscle of both limbs for fast and slow walks is similar. The numerical value for median frequency for Gastrocnemius and Soleus muscles of right and left leg is tabulated in Table 5.4. It manifest that median frequency component follows a pattern where the frequency decreases with increase in walk speed and variation in median frequency of the signal for the same muscle of both limbs for fast and slow walks is similar.

Table 5.4: Median Frequency of EMG signal averaged over three trials for fast and slow walks

<table>
<thead>
<tr>
<th>Channel</th>
<th>Trace Title</th>
<th>Fast Walk (Hz)</th>
<th>Slow Walk (Hz)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>GRL</td>
<td>69.00</td>
<td>84.77</td>
</tr>
<tr>
<td>2</td>
<td>SRL</td>
<td>72.45</td>
<td>73.60</td>
</tr>
<tr>
<td>3</td>
<td>GLL</td>
<td>63.45</td>
<td>66.01</td>
</tr>
<tr>
<td>4</td>
<td>SLL</td>
<td>65.20</td>
<td>69.60</td>
</tr>
</tbody>
</table>
5.4.2 Results for Integrated Sensor (EMG and Electrogoniometer) Data Analysis on Ground and Treadmill in All Gait Phases

EMG sensors were attached at four muscle positions and electrogoniometer at joint and the data was acquired synchronously and data was analysed offline.

5.4.2.1 Parameters while Walking on Level Ground

The data was analysed with respect to walking speed. Figure 5.16 represents the EMG signal for all four muscle [(1) VL, (2) BF, (3) GA, (4) TA] along with knee flexion angle [(8)] for one walk, walking with slow speed. Figure 5.17 and 5.18 shows the analysed data for normal and fast walking speed respectively. It is analysed with the walk cycle time being 1.5s, 1.35s and 1.2s for slow, normal and fast speeds. The activity of GA muscle was found to be dominant in the stance phase and that of TA in the swing phase. Hence their use for control in the respective phases is justified. For control purpose, an absolute value was needed for the control variable so the RMS and Mean frequency response of the signals were taken.

![Fig. 5.16: EMG signal for 4 muscles (1)VL, (2)BF, (3)GA, (4)TA and (8)Knee Flexion Angle acquired for one gait cycle for Slow speed](image-url)
1.1205 = 1 (mV)
-0.534 = 2 (mV)
0 = 3 (mV)
0 = 4 (mV)
8 (deg) 0

Fig. 5.17: EMG and KFA for normal walking speed

1.1205 = 1 (mV)
-0.534 = 2 (mV)
0 = 3 (mV)
0 = 4 (mV)
8 (deg) 0

Fig. 5.18: EMG and KFA for Fast walking speeds
It is observed that RMS analysis of EMG and knee flexion angle show detectable variation with respect to walking speed. The RMS values of the amplitude of the muscles VL (Vastus Lateralis), BF (Bicep Femoris), GA (Gastrocnemius), and TA (Tibialis Anterior) of the natural leg for the stance and swing phase and for all three paces of walk (slow, normal and fast) for a single individual are shown in Table 5.5. It was observed that the knee angle range increases with increase in speed of walk in the swing phase.

Table 5.5: RMS values of EMG signals and Knee flexion angle range for three walking speeds for stance and swing phase

<table>
<thead>
<tr>
<th>Trace Title</th>
<th>Stance</th>
<th>Swing</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Slow (mV)</td>
<td>Normal (mV)</td>
</tr>
<tr>
<td>VL</td>
<td>0.0502</td>
<td>0.0772</td>
</tr>
<tr>
<td>BF</td>
<td>0.096</td>
<td>0.114</td>
</tr>
<tr>
<td>GA</td>
<td>0.3023</td>
<td>0.3435</td>
</tr>
<tr>
<td>TA</td>
<td>0.1763</td>
<td>0.1868</td>
</tr>
<tr>
<td>Knee Angle Change in °</td>
<td>24.2</td>
<td>23.4</td>
</tr>
</tbody>
</table>

The frequency domain analysis of acquired data for mean frequency (MF) and median frequency (MDF) is done for both gait phases and reported in Table 5.6 and Table 5.7 respectively. No remarkably discriminative change in frequency is recorded. But the values approximately repeat for different walk cycles of a person and hence could be quantified as a control parameter for that person with a ±5mV change. It is found that the median frequency response for Gastrocnemius muscle followed the same pattern for both the experiments proving its repeatability and reproducibility. No such reproducible pattern is observed for mean frequency analysis.
Table 5.6: MF values of EMG signals and Knee flexion angle range for three walking speeds and both phases for a single walk cycle

<table>
<thead>
<tr>
<th>Trace Title</th>
<th>Mean Frequency</th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Stance</td>
<td>Slow (Hz)</td>
<td>Normal (Hz)</td>
<td>Fast (Hz)</td>
<td>Slow (Hz)</td>
<td>Normal (Hz)</td>
<td>Fast (Hz)</td>
</tr>
<tr>
<td>VL</td>
<td>38.10</td>
<td>60.47</td>
<td>37.66</td>
<td>36.34</td>
<td>42.10</td>
<td>33.16</td>
<td></td>
</tr>
<tr>
<td>BF</td>
<td>40.50</td>
<td>39.49</td>
<td>65.13</td>
<td>37.66</td>
<td>47.72</td>
<td>44.78</td>
<td></td>
</tr>
<tr>
<td>GA</td>
<td>104.75</td>
<td>68.81</td>
<td>84.78</td>
<td>44.25</td>
<td>54.50</td>
<td>39.06</td>
<td></td>
</tr>
<tr>
<td>TA</td>
<td>69.12</td>
<td>59.56</td>
<td>79.13</td>
<td>56.32</td>
<td>72.50</td>
<td>65.44</td>
<td></td>
</tr>
</tbody>
</table>

Table 5.7: MDF values of EMG signals and Knee flexion angle range for three walking speeds and both phases for a single walk cycle

<table>
<thead>
<tr>
<th>Trace Title</th>
<th>Median Frequency</th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Stance</td>
<td>Slow (Hz)</td>
<td>Normal (Hz)</td>
<td>Fast (Hz)</td>
<td>Slow (Hz)</td>
<td>Normal (Hz)</td>
<td>Fast (Hz)</td>
</tr>
<tr>
<td>VL</td>
<td>40.07</td>
<td>41.97</td>
<td>39.07</td>
<td>39.07</td>
<td>46.88</td>
<td>31.25</td>
<td></td>
</tr>
<tr>
<td>BF</td>
<td>46.88</td>
<td>31.25</td>
<td>78.13</td>
<td>31.25</td>
<td>39.07</td>
<td>24.41</td>
<td></td>
</tr>
<tr>
<td>GA</td>
<td>85.94</td>
<td>60.54</td>
<td>77.62</td>
<td>39.06</td>
<td>39.07</td>
<td>39.07</td>
<td></td>
</tr>
<tr>
<td>TA</td>
<td>76.66</td>
<td>78.13</td>
<td>85.94</td>
<td>61.07</td>
<td>93.75</td>
<td>78.12</td>
<td></td>
</tr>
</tbody>
</table>

Comparison is made between the analysed data for the two gait phase and it is found that there is little change in stance phase as compared to swing phase. The RMS values of EMG signals showed a similar pattern with the voltage values increasing with increase in speed of walk. The GA muscle was found to be dominant during the stance phase and TA during the swing phase.
with comparable change in RMS values with change in walking speed. Thus signals from GA and TA muscles were employed for control purpose.

5.4.2.2 Parameters while Walking on Level Ground (LG) and Treadmill (TM)

The use of treadmill is question debatable for EMG analysis, but it has advantages like speed control and better walking environment. While walking on level ground, one cannot judge the correct walking speed of the subject, whereas in treadmill, speed can be fixed and person has to walk with the fixed speed. Due to fixing of speed, gait may not remain natural. To reach a conclusion, we have acquired and analysed the data of EMG and KFA while subjects were walking on level ground (LG) and treadmill (TM). For better understanding, results are evaluated till the sub-phasic level of gait cycle.

5.4.2.2.1 RMS Analysis for Treadmill & Level Ground

As done previously, data is analyzed for RMS ranges and mean and median frequency. The RMS values for gait sub-phase, loading response, mid stance, terminal stance, pre-swing, initial swing, mid swing and terminal swing are compared for treadmill walk and level ground walk at slow and fast speed. Figure 5.19 depicts the RMS amplitude of four muscles (VL, BF, GA, and TA) in each gait sub-phase while the subject is walking on treadmill and level ground. It is observed that comparing the amount of EMG generated in four muscles is almost similar while walking on slow and fast speed on treadmill and level ground. Figure 5.20 shows the RMS amplitude between treadmill and level ground walking at fast speed. The analysis proves that the BF and GA muscles have prominent activity in each sub-phase and thus these muscles can be used for prosthetic control based on sub-phasic gait (EMG) activity.
5.2.2.2 RMS Analysis for Walking at Inclined & Level Treadmill

To ascertain the effort generated by human body while walking, analysis of data on an inclined plane is required to be done. Data is acquired for treadmill walking at level (no inclination) or inclined (with inclination of 15°). RMS value is analysed for each gait sub-phase for four muscles while the subject is walking at inclined and level treadmill with slow and fast walking speed.

Fig 5.19: RMS amplitude in gait sub-phases over treadmill & level ground at slow walking speed

Fig. 5.20: RMS amplitude in gait sub-phases over treadmill & level ground at fast walking speed
speed. Figure 5.21 reports the RMS values when the subject is walking with slow speed on inclined and level treadmill. Figure 5.22 shows the inclined and level treadmill RMS values, subject walking at fast speed.

Fig. 5.21: RMS amplitude in gait sub-phases over inclined & level treadmill at slow walking speed

Fig 5.22: RMS amplitude in gait sub-phases over inclined & level treadmill at fast walking speed
5.5 Strategy for EMG based Prosthetic Control

Development of an advanced and intelligent prosthetic for a trans-femoral amputee is a challenging task. Many researchers and prosthetic engineers developed a control strategy using knee flexion rate of the prosthesis and thus governing the walking speed of amputee in swing or stance phase. Some recent innovations also report a prosthetic device controlled by using the EMG signals. There are several limitations when SEMG or KFA are used for prosthetic control individually. For example in EMG, fixing of surface electrode to the residual limb, variation of signal due to muscle atrophy and other environmental constraints are some engineering factors. With the experimental result we have established a relationship between the knee flexion angle and SEMG signals from the four identified muscles which help in walking.

Keeping the view of above limitations, a control strategy is generated and implemented, based on knowledgebase developed using SEMG and KFA for controlling the knee joint. In our approach EMG electrodes are placed on the non-amputed limb of the unilateral above knee amputee to minimize the effects on quality of EMG signals due to amputation. Electro-goniometer is attached to the prosthetic knee joint to measures the flexion angle which are mapped to the EMG signals obtained from the lower limb muscles. A knowledgebase based on the mapping of relationship between the activities of various muscles during the different phases of a Gait cycle is developed. The flexion angle profile gives an understanding about the phase of the walk. The EMG signal parameters like RMS amplitude and mean and median frequencies also share a relationship with the speed of walk.

Fig 5.23 shows the system block diagram. The SEMG data is acquired and analyzed using signal processing techniques to determine the RMS amplitude (time domain) and their mean and median frequency responses (frequency domain). A knowledge base is created on the basis of KFA, SEMG RMS values and mean frequency (MF) values being mapped to the phase and walking speed of individual. The mapped knowledge base is stored in the memory of the controller which is used to control the flexion rate of the prosthetic by controlling the electro-pneumatic valve of the pneumatic cylinder used in prosthetic.
5.5.1 Knowledge Base Generation

Experiments were performed to generate knowledge-base for phase of walk, angle and EMG patterns to categorize walking speed as slow, normal and fast based on the self selected pace of different individuals. Angle and EMG signal data were collected with the subjects walking at three self selected pace of slow, normal and fast and stored for further processing. The FSR sensor gives the heel strike and toe off time thus indicating the exact start and end of a Gait cycle. This also helps in differentiating the two phases of a walk. The phase of walk can also be easily identified from the flexion angle curve and thus a walk cycle along with its constituent events like heel-strike, toe-off etc can be defined.

Table 5.8 shows generated knowledge base of the controller based on analyzed results for the experiments conducted on normal healthy individuals showed certain patterns which were crucial in deciding the control strategy. The most important revelation was that the angle ranges, EMG signal amplitude and frequency response were similar, followed a pattern and could be quantified for an individual. But the values for a different individual for the same phase of walk were quite different, though they often followed the similar incrementing or decrementing pattern with speed. This development was quite expected as the EMG signal pattern and the
flexion angle pattern depended on the individual’s physical and physiological traits and gait. A prosthetic device is user specific and hence the controller needs to be programmed according to the user’s physiological traits and requirements so the above developments do not prove to be a hindrance in the design process.

| Table 5.8: Knowledge base for the controller to decide the pace of walk |
|-----------------------------|---------------------|---------------------|---------------------|
| **Phase of Walk** | **Angle Range** | **EMG** | **EMG** |
| | | **Muscle** | **RMS (V)** | **MF (Hz)** | **Speed** |
| **Stance** | 24.2 | GA | 0.30±0.05 | 104.75±0.05 | Slow |
| | 23.4 | | 0.34±0.05 | 68.81±0.05 | Normal |
| | 24.5 | | 0.41±0.05 | 84.78±0.05 | Fast |
| **Swing** | [Angle] | TA | 0.15±0.05 | 56.32±0.05 | Slow |
| | [(Angle+3)±0.5] | | 0.22±0.05 | 72.50±0.05 | Normal |
| | [(Angle+6)±0.5] | | 0.27±0.05 | 65.44±0.05 | Fast |

5.5.2 Control Algorithm

Initially the controller starts its operation from slow speed mode. For the safety of patient and better control stability, first two walk cycle data is not utilized by the controller and during that period the controller runs in the slow speed mode. The third walk cycle data is utilized by the controller for analysis and the corresponding control action is implemented. The FSR sensors fixed on to both the prosthetic feet and the healthy feet of the subject gives information about phase of walk and start/end of gait cycle. The controller differentiates the phase of the walk as stance or swing according to the signal received i.e. heel-strike for the start of stance and toe-off for swing (Fig. 5.24).
Once the phase is decided the controller checks both the angles detected by the goniometer sensor fixed to the prosthetic knee and the EMG from the healthy limb muscles. For swing phase the controller checks both the KFA and SEMG signals from the TA muscle. In stance phase the controller neglects the KFA and just considers the SEMG signal from GA muscle for control analysis. The controller compares the sensor data with the knowledge base stored in the controller’s memory. In case of a match with the stored values the control for the corresponding speed is generated. In case of a mismatch i.e. when the values reported by the sensor is far different from the normally observed values stored in the knowledgebase the control for the previous phase speed is continued (Fig. 5.25).

Fig. 5.24: Flow diagram for gate cycle and phase determination during walk
In the swing phase, controller generates control according to the KFA and RMS and MF values of EMG if they fall in the expected range. It then compares if the control action from both the algorithm is the same, if positive then the control is implemented. If both algorithms report for a different control then the control action from the KFA algorithm is implemented according to its priority and SEMG control is discarded.

Control priority of the parameters is evaluated and fed into the embedded software. The priority of flexion angle is kept more than the EMG signals since speed variations can be mapped more accurately using the former as compared to the latter. Moreover EMG signals are random in nature and may vary with change in physiological conditions of the muscles. The real time controller generates control for the pneumatic actuation system based on the level of the signal. The swing phase control is relatively complex as compared to stance phase where just the EMG signal for GA is considered for control in contrast with the former where both the KFA and EMG signal parameters were considered (Fig.5.26).
5.6 Conclusions

The chapter describes the EMG pattern variations of muscles with the help of surface EMG acquisition and analysis. Experiments to acquire EMG with the help of active surface electrode are carried out for selective muscles of lower limb. The SEMG is analysed on the basis of RMS amplitude values and frequency analysis. It is observed that EMG and angular (linear &
rotational) knee data, at higher walking speeds; either on treadmill or level ground, the EMG and joint angle variation is higher which was in accordance with literature.

It was also observed that apart from RMS values, the median frequency values could also be used as a determinant of walking speed. These results are helpful for quantification of prosthetic devices like AK electronic knee where clinical trials are done on ground level with more physical and experimental constraints. A database of EMG activity of lower limb muscles responsible for walking of normal Indian ethnicity population is generated and which can be used as a reference in other diagnosis.

It was observed that SEMG and angular variation for hip and knee joint angles are comparable at same speeds for treadmill and level ground walk. The findings support the usage of treadmill as a viable EMG acquisition tool and helpful prosthetic rehabilitation.

The experiments conducted on healthy individuals to generate the knowledge base showed some interesting patterns which proved crucial in devising the control strategy. It can be well proven here that EMG activity is reciprocatory for each leg in each phase. A new approach in EMG based prosthetic control is applied by recording the, RMS values of muscles VL, BF, GA, and TA of the non amputated leg for the stance and swing phase and for all three paces of walk (Fast, Normal and Slow) for a single individual is used for generation of knowledgebase. The activity of GA muscle was found to be dominant in the stance phase and that of TA in the swing phase. Hence their use for control in the respective phases is justified. A control strategy is generated based on KFA and SEMG both mapped in terms of walking speed for controlling the swing and stance phase of artificial knee joint. As the approach here to prove the concept, the mapping was done for few parameters with respect to gait phases this makes the knowledgebase limited and control output is crude only deciding the walking speed. This approach is useful for developing a low-cost intelligent prosthesis and it surely serves the purpose. With these results applied with a high degree of precision, an intelligent controller system could be developed which can restore the walking pattern of a trans-femoral amputee to its original using the SEMG from the normal leg muscles. There is need to identify the normal ranges of EMG variations of local population. This would help in order to develop EMG based controls of prosthetic devices.