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MMG SIGNAL: RECTUS FEMORIS MEASUREMENT PROTOCOL

USE OF MMG SIGNALS FOR THE CONTROL OF POWERED ORTHOTIC DEVICES: DEVELOPMENT OF A RECTUS FEMORIS MEASUREMENT PROTOCOL

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Abstract

A test protocol has been defined for the purpose of measuring rectus femoris mechanomyographic (MMG) signals. The protocol is specified in terms of the following: measurement equipment, signal processing requirements, human postural requirements, test rig, sensor placement, sensor dermal fixation and test procedure. Preliminary tests of the statistical nature of rectus femoris MMG signals were performed and Gaussianity was evaluated by means of a two-sided Kolmogorov-Smirnov test. For all 100 MMG data sets obtained from the testing of two volunteers the null hypothesis of Gaussianity was rejected at 1%, 5% and 10% significance levels. Most skewness values were found to be greater than 0.0 while all kurtosis values were found to be greater than 3.0. A statistical convergence analysis also performed on the same 100 MMG data sets suggested that 25 MMG acquisitions should prove sufficient to statistically characterise rectus femoris MMG. This conclusion is supported by the qualitative characteristics of the mean rectus femoris MMG power spectral densities obtained using 25 averages.

Keywords: MMG, EMG, rectus femoris, muscle, accelerometer, orthosis, lifting

1 INTRODUCTION

Numerous devices have been developed for assisting individuals who suffer manipulative or locomotive disabilities. It has been suggested [25] that such devices belong to one of two major categories: rehabilitation technologies or assistive technologies. The first category includes external devices whose action helps to modify the structural or functional characteristics of the neuromusculoskeletal system. The second category consists of body-worn devices which assist the movement of disabled individuals.

A device belonging to the second category is the powered lower-limb orthosis developed at the University of L'Aquila, which is presented in Figure 1. It consists of a frame composed of lower and upper aluminium rods, a steel hinge joint and two shells in thermosetting plastic [24]. The orthosis provides a lifting movement which assists when rising from a seated posture. Orthopaedically correct motion is achieved by means of a knee joint and a cable-pulley system. Two McKibben-type pneumatic muscles [23] are symmetrically located on the opposing sides of the leg in correspondence with the tibia. Introduction of compressed air inside the pneumatic muscles allows their shortening, and thus produces the lifting force. The orthosis is currently controlled by means of a sensing circuit formed by an air bladder (which is wrapped around the circumference of the upper leg) and a pressure transducer which detects user intention by measuring the hardness of the rectus femoris muscle. Two control strategies which are based on the air bladder signal have been developed and tested, with interesting preliminary results: an on/off control, with one or more pressure threshold values, and a continuous control. Unfortunately, the control system based on the air bladder sensor has been found to be highly sensitive to both variations in contact pressure with the muscle

and to air losses due to leakage. The research described here was performed in part due to the need to establish a more natural and efficient command system for the orthosis.

[Insert Figure 1 here]

According to some authors, command techniques for assistive devices can be classified into three categories [34]: those based on signals originating from within the muscular-skeletal system, those based on biologically-derived signals not originating from within the muscular-skeletal system and those based on signals derived from kinetic or kinematic measurements of the body.

Signals originating within the muscular-skeletal system include Electromyographic (EMG) signals produced by the electro-chemical activity involved in motor unit activation. EMG signals are normally measured within the muscle tissue itself or at the external derma by means of either needle or surface electrodes [12]. Advantages of EMG include the fact that they provide signals whose amplitude increases as a function of increasing muscle force [5, 12]. Further, EMG provides a measure of command intention since the electrical activity builds prior to actual muscular contraction [12]. Disadvantages of EMG surface electrodes include possible mechanical detachment of the myoelectrodes due to body motion, electrical interference from nearby materials or devices, and interference caused by impedance changes of the skin. Examples of the use of EMG for the control of powered orthotic devices currently include an ankle-foot orthosis [10].

Signals originating within the muscular-skeletal system also include force, motion or volume change measures related to muscles, tissue or tendons. Advantages of such measures include

their relative robustness against external electrical disturbances and their amplitudes which are normally sufficiently high as to not require sophisticated signal conditioning. Disadvantages include the need to assure the correct positioning of the sensor and the possibility that accidental knocks can generate unwanted signals. Examples of the use of these command signals include the Athroscopically Implantable Force Probe [26, 33], a Hall effect device used to control an upper limb prothesis whose output is proportional to muscle dimensional change [14] and a Tendon Activated Pneumatic sensor used to control a hand prosthesis whose output is proportional to tendon sliding movement [6].

Signals originating from within the muscular-skeletal system also include those which are derived from the mechanical oscillations which occur due to the sliding of the muscle fibres during contraction [17]. Mechanomyographic (MMG) signals consist of either acceleration or sound measurements which are made on the outer skin over a target muscle. Like EMG, MMG measures a command intention that is produced voluntarily, but in this case there is a greater time delay due to the physical response time of the muscle fibres. An advantage of MMG is that signal conditioning is less critical than in the case of EMG [8, 19] because neither the measurement of acceleration nor sound is affected by the electrical impedance of the skin. Disadvantages of this method include the possibility of noise signals generated by accidental knocks [37] and the possibility of sensor detachment. There are currently few studies in the research literature which describe the use of MMG for the control of assistive devices. One study [1] has described an externally-powered hand prosthesis which is controlled by means of an MMG signal measured at the wrist extensor and flexor muscles. A second study [37] has demonstrated the usefulness of MMG in 2-function hand prosthesis control signal. In this case, the system is based on a triplet distally recorded, normalised, root

mean square (RMS) MMG signals which are passed through software and hardware modules that emulate the output of a conventional EMG sensing system.

Signals originating within the muscular-skeletal system also include those produced by Near Infra-Red Spectroscopy, which are related to concentration changes in tissue haemoglobin [9, 31]. The technique measures the absorption of near-infrared light radiation caused by the oxygenation of living tissues. Higher muscular effort causes higher oxygen concentration change. An advantage of the technique is its potential usefulness in cases where fatigue onset requires monitoring. Disadvantages of the technique include the need for complex sensors and measurement instrumentation, and the sensitivity to the level of ambient lighting. No examples of the use of these signals for orthotic devices were found in the research literature.

Biologically-derived signals not originating from within the muscular-skeletal system include Electroencephalographic (EEG) signals, recorded from the scalp using surface mounted electrodes. Advantages of this technique include the possibility of obtaining command signals even for patients who are unable to generate muscular activity (such as patients with tetraplegia) and the non-invasive nature of the measurement. Disadvantages include the complexity of interpretation of the signals and the loss of much information due to the filtering action of the skull, tissues and skin. Examples of the use of EEG for the control of assistive devices include an electrical orthosis [20] and a neuroprosthesis [38] which restore hand grasp function for, respectively, patients with tetraplegia and patients with severe physical disabilities (i.e. amyotrophic lateral sclerosis and stroke).

Signals based on kinematic or kinetic measurement of the interaction between the human body and its surrounding environment include a variety of forces, motions and pressures

which arise from contact between the human body and either the assistive device or the natural environment. Such signals have the advantage of being strictly related to human intention and of often being relatively simple from the point of view of sensor technical requirements. Examples of the application of such signals to the control of orthotic and prosthetic devices include on/off switch systems [7], potentiometers [3], mini-joysticks [7], strain-gauges [21] and load cells [27].

When considering which control technique to adopt in the case of the L'Aquila University powered lower limb orthosis the main criteria were taken to be the simplicity of installation and the simplicity of use as a physical component of the orthosis. Review of the relevant literature suggested that MMG was the most appropriate technology and that its potential use as a control signal had been demonstrated [1,37]. The research described here was therefore performed in order to define a protocol for accurately measuring rectus femoris MMG signals in a laboratory setting for the purpose of developing a new control system for the L'Aquila University powered lower-limb orthosis. New research was required to develop a rectus femoris MMG signal measurement protocol because no such protocol was identified in the literature. The research described here investigated the variation of the MMG signal characteristics under muscular loading conditions which would be expected to occur during the use of the powered lower-limb orthosis. The principle statistical property investigated was the RMS value of the signal since it is this value which would be expected to be the basis of the control action of the orthosis [37]. Instantaneous values of the RMS were measured under specific conditions and the statistical convergence of the statistic under multiple repetitions was determined. This paper summarises the rectus femoris test protocol, which is described below in terms of the necessary measurement equipment, the signal processing requirements,

the human postural requirements, the test rig, the placement and dermal fixation of the measurement sensor and the procedure.

2 TEST PROTOCOL

MMG signals originate from the mechanical oscillations produced by activated motor units, mediated and modulated by the muscle-tendon complex, fat and skin. The causes of the oscillations include: (1) a slow bulk lateral movement of the muscle at the initiation of a contraction that is related to the uneven distribution of the contractile elements, (2) the excitation into ringing of the muscle at its own resonant frequency and (3) the pressure waves generated by the dimensional changes of the fibres of the active motor units [18]. Unequal recruitment of the different motor units [36] results in MMG signals whose shape and amplitude can be different for the same level of voluntary contraction, thus they can be classified to fall within the general category of random signals [16]. Examples of rectus femoris MMG signals collected from the same subject for different repetitions of the same level of muscular contraction are presented as Fig. 2.

[Insert Figure 2 here]

Consideration of the design of a powered lower-limb orthosis suggests that a possible muscle group to monitor for control purposes is the quadriceps. In the case of lifting actions such as rising from a chair, the rectus femoris is the main contributor to the motion due to its role in performing leg extension. During the preliminary movement phase, in which a powered orthosis would be expected to provide assistance, the contraction of the rectus femoris can be described as isometric.

A literature review was performed in the areas of MMG [17, 19, 29, 35] and human quadriceps muscle behaviour [30]. A variety of MMG test protocols were identified which adopted different physical and signal processing parameters. The current lack of a standardised approach suggested the need for experimentation to identify optimal choices of the following parameters:

Measurement equipment

- measurement sensor
- signal conditioning system

Signal processing requirements

- sampling frequency
- band-pass filter and cut-off frequencies
- time window

Human postural requirements

- trunk position
- thigh position
- ankle position

Test rig

- system for measuring body-part alignment
- system for aligning the force measurement sensor with respect to the human body
- system for eliminating floor sliding friction
- system for visualising the instantaneous value of muscle contraction

Sensor placement and dermal fixation

- number of transducers
- physical points of measurement
- transducer fixation method

Test procedure

- choice of muscular contraction levels
- choice of muscular contraction time duration
- choice of relaxation time between two consecutive tests
- choice of total number of tests needed for statistical accuracy

A lengthy laboratory investigation involving two human subjects was performed in which the test parameters listed above were varied individually or in groups. The logical criteria adopted for the investigation was that any given choice or setting of the above should lead to MMG signals which were approximately zero and nearly noise-free when under conditions of muscular relaxation, and MMG signals which were strong, relatively noise-free, and similar to published results available in the literature when under various degrees of muscular contraction. The following sections present a detailed description of the protocol which resulted from the experimentation.

2.1 Measurement equipment

While microphones, contact sensors and accelerometers can all be used to produce MMG signals, accelerometers have most frequently been recommended as the most convenient and

accurate measurement method [35]. An Analog Device ADXL 203 capacitive accelerometer (Analog Devices Inc., USA) was chosen for use in this study due to its high sensitivity and low mass. The accelerometer's measurement range is ±1.7g, its frequency bandwidth is from 0 to 2,500 Hz, its sensitivity is 1,000mV/g, its mass is less than 1.0 gram and its geometric dimensions are 5mm x 5mm x 2mm (0.2in. x 0.2in. x 0.08in.). The accelerometer was used to measure the oscillations on the skin surface directly above the rectus femoris. Signal amplification was not required due to the high sensitivity of the accelerometer, however an analogue anti-aliasing filter was adopted. A 16 bit NIDAQ CARD 6036E (National Instruments Corporation, USA) was used for data acquisition as was National Instruments LabView 6.1 software (National Instruments Corporation, USA). All data were stored on an Intel 3.2 GHz Pentium 4 personal computer.

2.2 Signal processing requirements

Preliminary data acquisitions were performed using an anti-aliasing bandwidth of 100 kHz, which was found to be much greater than the bandwidth of the MMG signals. Upon successive acquisitions the bandwidth was gradually reduced until achieving a final value of 200 Hz. The 0-200 Hz low-pass anti-aliasing filter was achieved by placing a 0.027μF condenser in parallel to the signal line, as recommended by Analog Device (Analog Devices Inc., USA).

All MMG signals were acquired in differential mode at a sampling rate of 1kHz and pass-band filtered to the frequency range from 0.5 to 140 Hz by means of a 12th-order recursive Butterworth filter (National Instruments LabView, Measurements Manual). The time data window used for all acquisitions was 4.096 seconds.

2.3 Human postural requirements

In clinical practice the force characteristics of the rectus femoris are normally measured for the case of isometric contraction, which is achieved by blocking the patient's ankle during leg extension [13]. During the test the patient sits on an appropriate horizontal surface, and uses his or her hands to grip the edges of the sitting surface so as to stabilise the upper body position. The clinician then positions the patient's leg so as to achieve a posture defined by a horizontal thigh and an angle of approximately 90 degrees between the thigh and the lower leg. The clinician then grasps the patient's ankle to fix it in position and requests the patient to raise the lower leg using maximal voluntary contraction.

2.4 Test rig

A laboratory test rig was designed and manufactured for the purpose of reproducing the clinical guidelines in the most accurate manner possible. As shown in Figure 3 it consists of a structure which includes a seat which is adjustable in height, a polyester belt to fix the ankle in place and a sliding block which reduces floor sliding friction. The test rig also includes a TS200 load cell (AEP Transducers, Italy) for measuring the extension force of the lower leg (8), a TDS210 digital oscilloscope (Tektronix Inc., USA) (11) and an ELD9105 digital voltmeter (Eldes Instruments, Italy) (14).

[Insert Figure 3 here]

The sitting posture is controlled by means of three markers. Marker 1 is placed on the lateral malleolus, marker 2 on the lateral condylus femoralis and marker 3 on the great trocanter. The height of the adjustable seat is varied so as to bring marker 3 to the same vertical level as marker 2. The measurement of the final height of the seat above the floor is performed using a system (4) consisting of a pointer and a calibrated scale which is accurate to within 1 mm (0.04in.). The horizontal alignment between markers 2 and 3 is measured by means of a horizontally placed 0.82 meter pointer (32,3 in.) and two calibrated vertical scales (5). The maximum error in angular alignment is estimated to be approximately 2 degrees, based on a maximum potential placement error of 3 mm (0.12 in.) for each of the two markers with respect to their respective bone reference points. The vertical alignment between marker 1 and 2 is assured by a plumb-line which is placed on the horizontal beam. In this case the maximum error in vertical alignment is estimated to be approximately 2 degrees for persons with lower leg lengths from 0.350 to 0.420 m (from 13.78 to 16.53 in.). The test rig also incorporates a handle (6) which is gripped by the test subject so as to stabilise the upper body position.

The test rig incorporates a polyester belt (7) which is worn by the human subject around his or her ankle, and which is fixed to the load cell. The load cell (8) is fixed to the test rig frame by means of an attachment point (9) which can be adjusted to be at the same height and lateral position as the ankle of the test subject. During testing, the foot rests on an aluminium reaction block (10) which is greased on its underside so as to reduce the effect of horizontal friction on the measurement of leg force. This eliminates, practically, possible ankle moments.

2.5 Placement and dermal fixation of the measurement sensor

The frequency content of MMG signals has been shown to depend upon the strength of fixation of the sensor to the outer skin of the test subject [2]. In order to understand this aspect of the test protocol, the accelerometer was fixed to the skin surface by means of bi-adhesive tape as shown in Figure 4 and tests were performed with two subjects in which various knocks and movements were made for a period of 45 minutes. A rectangular piece of tape was used which had a thickness of 0.19 mm (0.007 in.) and which was cut to a size which was 3 mm (0.12 in.) longer on each side than the accelerometer. MMG acquisitions were made at various points during the sequence to establish if signal characteristics remained unaltered. The quality of the acquired signals suggested that bi-adhesive tape was sufficient to guarantee adequate coupling for a period of at least 45 minutes.

[insert Figure 4 here]

The frequency content of MMG signals has also been shown to depend on the physical location along the muscle group [4]. Positions along the centre line of the muscle, and in particularly near the belly of the muscle, have been shown to produce maximal magnitudes of the MMG signal [4]. MMG magnitude has been found to decrease with increasing distance from the centre point of the muscle [4]. These considerations suggested the adoption of the centre point of the muscle as the location which promised the strongest and most easily interpreted signals.

The frequency content of MMG signals is also expected to depend on the physical characteristics of the sensor system. The coupled system consisting of the target muscle, the fat layer, the skin, the mounting tape, the accelerometer and the moving mass of the

accelerometer cable can be considered to act as a multi-degree of freedom mechanical oscillator. From studies of the mechanical response of accelerometers which are mounted to skin layers [15], and from general theory of mechanical oscillators [22], it was considered prudent to reduce the mass of the tape, of the accelerometer and of the accelerometer cable, and to fix the accelerometer to the skin using the strongest possible bond (the greatest possible coupling stiffness). Further, it was considered important to limit the spatial extent of the bi-adhesive tape to approximately the area of the accelerometer so as to avoid large areas of tape modifying the boundary conditions of the muscle and skin motion.

2.6 Procedure

The procedure consists of first a measurement of the force developed during maximal voluntary contraction (MVC), followed by the application of the accelerometer and the recording of MMG measurements relative to fixed percentages of MVC. The steps followed to measure the force of MVC are:

Step 1: the test subject is requested to sit in the test rig.

Step 2: the test subject is asked to put on the polyester ankle belt, which is then fixed to the load cell.

Step 3: the three markers are positioned and body-part alignment is verified and tuned.

Step 4: the MVC test is explained to the subject verbally.

Step 5: preliminary MVC measurements are attempted to familiarise the test subject with the measurement.

Step 6: the test subject is requested to attempt to extend his or her leg with the greatest possible force for 5 seconds while grasping the test rig for support. The maximal force measured by the load cell (8) is read from the screen of the digital voltmeter (14).

Step 7: the test subject is requested to repeat step 6 a total of four additional times, for a total of 5 measurements, which are averaged to determine a mean value of MVC. A one-minute rest period is observed for each repetition of step 6.

After MVC measurement the accelerometer is placed at the centre-point of the muscle and the MMG signals are acquired. An oscilloscope (11) is used to monitor the load cell output voltage to achieve the target value of MVC. A target reference level is set at an established percentage of the test subject's previously recorded mean MVC force. During each test the subject is required to start from rest then reach the target value as quickly as possible, then hold the target value for 1.5 seconds before returning to rest. The steps followed to measure MMG are:

Step 1: the centre point of the rectus femoris is localized by asking the subject to perform a leg extension at MVC and by searching for the point of maximum muscle expansion.

Step 2: the centre point of the rectus femoris is cleaned, and where necessary shaved, in order to facilitate the adherence of the bi-adhesive tape and accelerometer.

Step 3: a rectangle of bi-adhesive tape is applied to the centre point of the rectus femoris, then the accelerometer is secured to the tape.

Step 4: the MMG test is explained to the subject verbally.

Step 5: preliminary MMG measurements are attempted to familiarise the test subject with the measurement.

Step 6: an MMG recording is made of the acceleration data contained in a 4.096 second time window (1.5 seconds at rest, a contraction of 1.5 seconds and the remaining time at rest) relative to a leg extension at a target percentage of MVC. During the recording the subject achieves the target leg force while maintaining his or her upper body stabilised by means of the test rig handle supports.

Step 7: the test subject is requested to repeat step 6 a fixed number of times. A two-minute rest period is observed for each repetition of step 6.

3 EXPERIMENTAL TESTS ON RECTUS FEMORIS MMG SIGNAL STATISTICAL BEHAVIOUR

3.1 Experimental conditions

Two healthy male volunteers aged 27 and 30 years participated in the tests after signing an informed consent form and after declaring that they suffered no history of neuromuscular or orthopaedic disease. Table 1 presents a set of basic anthropometric parameters of possible relevance to the production of the measured MMG signals which were measured prior to testing. Each volunteer was asked to perform a set of 50 MMG measurements following the rectus femoris MMG protocol defined above. The established contraction level for all tests was 50% of MVC.

[insert Table 1 here]

3.2 Signal Gaussianity

Classical methods of vibration analysis often assume that the measured data is both stationary and Gaussian. Such processes are completely described by their Power Spectral Density (PSD), which characterises the distribution of vibrational energy in the frequency domain [16]. The testing of whether a data set is Gaussian or not, and thus of whether it is fully described by its PSD or not, can be achieved by means of the Kolmogorov-Smirnov test [28] which compares a given data set with a standard normal distribution. This non-parametric test can be used to define the confidence level at which the data can be considered Gaussian.

When deviations from Gaussian behaviour are expected, three global signal statistics are often used to quantify the size and extent of the deviation [11]. The first is the RMS value σ. For a zero mean process such as the MMG signals acquired in this study, the RMS value can be expressed in terms of the number of data points L and the sample time step Δt as

$$\sigma = \left\{ L^{-1} \sum_{j=1}^{L} x^2(j\Delta t) \right\}^{1/2} \tag{1}$$

The second is the skewness λ, which is the third normalised spectral moment, and which is defined as the average of the instantaneous values $x(j\Delta t)$ cubed. For a zero mean process the skewness can be expressed as

$$\lambda = L^{-1}\sigma^{-3} \sum_{j=1}^{L} x^3(j\Delta t) \tag{2}$$

while the third statistic which is often used to quantify the deviation from a Gaussian stationary model is the kurtosis γ, which is the fourth normalised spectral moment, which is sensitive to outlying data. For a zero mean process the kurtosis can be expressed as

$$\gamma = L^{-1}\sigma^{-4}\sum_{j=1}^{L} x^4(j\Delta t) \tag{3}$$

For a Gaussian stationary process the RMS value will quantify of the overall amount of signal energy, while the skewness should be close to 0.0 and the kurtosis should be close to 3.0. The greater the deviation from either 0.0 or 3.0, the greater the deviation from a Gaussian model.

A question of interest regarding rectus femoris MMG signals is whether or not they are Gaussian. In order to clarify this point, 50 MMG signals were recorded from each of the two volunteers. A two-sided Kolmogorov-Smirnov test was performed by means of the MATLAB version 6 software package [32]. The null hypothesis was that of a standard normal distribution while the alternative hypothesis was that of a non–normal distribution. For all 100 MMG data sets the null hypothesis was rejected at 1%, 5% and 10% significance levels. The nature of the deviation from Gaussianity is illustrated in Figure 5, which presents the distribution of skewness and kurtosis values determined from the 100 MMG data sets. Most skewness values are greater than 0.0, while all kurtosis values are greater than 3.0.

[Insert Figure 5 here]

3.3 MMG signal statistical convergence properties

According to the weak law of large numbers [28] a sequence of independent and identically distributed random variables with a finite common mean $\mu < \infty$ and partial sum given by $S_n = X_1 + X_2 + ... + X_n$ will provide a running mean value S_n / n which converges to μ. The question of interest is therefore not that of convergence, but rather, that of the rate of convergence.

A statistical convergence analysis was performed to establish the number of repetitions of the MMG acquisition required to obtain stable ensemble average statistics. The statistics chosen for the analysis were the RMS and the kurtosis. They were determined for each signal of the ensemble of 50 MMG acquisitions performed for each of the two test subjects. The resulting RMS and kurtosis values were then randomly ordered so as to reduce any systematic effects due to subject training or due to muscle fatigue. The running mean was then determined for each data series as shown in Figure 6.

[Insert Figure 6 here]

A box drawn in dashed lines in figure 6 indicates the point at which the data series first convergences to a value for which an interval of ±5% of the value contains the remaining data of the complete series (all 50 data measurements). The value of ±5% was considered appropriate since it is a commonly adopted tolerance in engineering and medical science [4, 19, 29]. As suggested by Figures 6a and 6b the RMS required only 7 or 5 MMG measurements to reach ±5% of the terminal mean value of the complete series, depending on the test subject. As shown instead in Figures 6c and 6d, a total of 24 or 25 MMG measurements were required to reach the same statistical convergence in the case of the

kurtosis value. From the statistical convergence results obtained for the two test subjects it was concluded that 25 acquisitions should prove sufficient to adequately characterise rectus femoris MMG data.

The possible sufficiency of 25 data acquisitions was also supported by frequency domain analysis of the MMG signals. As in the case of the global statistics, the PSD was determined for each MMG data acquisition, then the sequence order was randomised so as to reduce the effect of any deterministic relationships with time. Figure 7 presents the mean power spectral densities obtained by averaging the PSDs of the first 2, 3, 6, 12, 25 or 50 MMG signals. The mean PSD obtained from 25 averages is qualitatively similar to the mean PSD obtained using all 50 available signals. Characteristic peaks are present in both PSDs at 3.5 and 10 Hz.

[Insert Figure here 7]

4 DISCUSSION

The test protocol defined here was found to be easily learned by the test participants, and the requirement of achieving the fixed percentages of MVC did not produce noticeable muscular fatigue over the course of the 25 measurements. Dermal fixation of the accelerometer by means of bi-adhesive tape proved adequate for periods of time exceeding the minimum required to perform the 25 measurements. As had been previously reported by Cescon et al. [4] the position of the accelerometer on the muscle was found to have an important effect on the amplitude and spectral characteristics of MMG signals, therefore the protocol included provided sensor positioning guidelines.

Based on the research of Kitazaki and Griffin [15] the mass of the sensor used in the protocol is not felt to produce harmful low-pass filtering effects on the measured MMG signals. The possible effect of the mass of the adipose and skin tissue which is located between the muscle and accelerometer was not, however, quantified in the current study. The mechanical response characteristics of these tissues would be expected to produce some amount of low-pass filtering of the MMG, but no quantification of this filtering action was found in the research literature and none was possible in the current study.

A further mechanical property which was not possible to quantify as part of the current study was the effect of external loads which act upon the rectus femoris muscle. When integrated into an orthotic device the accelerometer might be expected to find itself in contact with components of the orthosis which create normal and/or tangential forces. Unfortunately, no research literature was found which quantified the possible effect of external force on the MMG signal, thus further research in this area would appear important.

The MMG signals acquired for the rectus femoris muscle in the current study are consistent with those reported by previous investigators [4,17,19,35,36] for other muscle groups. As shown in Figure 6, the time domain analysis suggested that the MMG signal amplitudes for volunteer 2 were lower than those of volunteer 1. Frequency domain analysis suggested that the greater part of the energy of all the MMG signals was below 50 Hz. Volunteer 2, who had a relatively average muscular structure, was characterised by both a reduced frequency bandwidth and reduced PSD amplitudes than his counterpart who had undergone sports training. This result agrees with that of Orizio [17] who suggested that muscular power training induces a shift of the PSD towards higher frequencies.

The test protocol defined here is a first step towards the development of an assistive lower limb orthosis. The protocol will be used to measure the rectus femoris MMG signals of individuals who suffer muscular weakness [18] due to multiple sclerosis, muscular dystrophy or old age. Such measurements are necessary prerequisites to orthosis development because the authors have found no MMG measurements for disabled individuals in the research literature. The statistical analysis of the rectus femoris MMG signals seems to suggest that the RMS value can be used as a control criteria for a lower limb orthosis, in a manner similar to that adopted for the prosthesis arm described by Silva [37]. In the lower limb application the MMG signal might be expected to suffer from some mechanical noise caused by movements of the wearer's body or of the orthosis itself. It is hypothesised, however, that it will prove possible to separate those frequency components from the base MMG signal, or to adopt a sensor technology [39] which minimises the difficulty.

5 CONCLUSION

A test protocol has been defined for the purpose of measuring rectus femoris MMG signals. The protocol is specified in terms of the following: measurement equipment, signal processing requirements, human postural requirements, test rig, sensor placement and dermal fixation and the test procedure. The protocol will serve as the basis for characterising rectus femoris MMG behaviour for the purpose of using this signal for the control of orthotic devices such as the L'Aquila University power lower limb orthosis.

Preliminary tests of the statistical nature of rectus femoris MMG signals were performed. The Gaussianity of rectus femoris MMG was evaluated by means of a two-sided Kolmogorov-Smirnov test. For all 100 MMG data sets obtained from the testing of two volunteers the null hypothesis of Gaussianity was rejected at 1%, 5% and 10% significance levels. Most

skewness values were found to be greater than 0.0 while all kurtosis values were found to be greater than 3.0.

A statistical convergence analysis performed on the same 100 MMG data sets suggested that only 7 or 5 measurements (dependent on the human test subject) were required to stabilise the mean root mean square value of the data set to within ±5% of the terminal mean value of the complete series. The same convergence analysis also suggested that 24 or 25 measurements were sufficient to achieve a mean kurtosis value which was within ±5% of the terminal mean value of the complete series. It was concluded that 25 MMG acquisitions should prove sufficient to statistically characterise rectus femoris MMG. This conclusion is supported by the qualitative characteristics of the mean power spectral densities obtained using 25 averages.

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FIGURE LEGENDS

Fig. 1 University of L'Aquila powered lower-limb orthosis

Fig. 2 Time histories of five MMG signals collected from the same subject (volunteer 1). The test consisted of first 1,5s at rest, followed by 1,5s of isometric contraction at 50% of maximal voluntary contraction, then 1s at rest

Fig. 3 Test rig: (1) marker 1; (2) marker 2; (3) marker 3; (4) height adjustable seat; (5) markers alignment measure system; (6) handle; (7) inextensible belt; (8) load cell; (9) load cell attachment point; (10) sliding block; (11) oscilloscope; (12) personal computer; (13) data acquisition board; (14) voltmeter; (15) accelerometer

Fig. 4 Dermal fixation of the accelerometer by bi-adhesive tape

Fig. 5 Skewness and Kurtosis values determined for each of the 50 contraction MMG time histories. Data sets (a) and (c) are from subject 1 while sets (b) and (d) are from subject 2. Each original time history contained 1,500 data points

Fig. 6 Ensemble mean RMS and Kurtosis values determined from the group of 50 isometric contraction MMG time histories. Data sets (a) and (c) are from subject 1 while sets (b) and (d) are from subject 2. Each original time history contained 1,500 data points

Fig. 7 Mean PSD obtained using (a) 2, (b) 3, (c) 6, (d) 12, (e) 25 and (f) 50 averages. Each individual PSD was computed using a real FFT, a Hanning window and a data block size of 4,096 data points.

TABLES

Table 1 Anthropometric data of the subjects

	Volunteer 1	Volunteer 2
Musculature [type]	evidence of sports training	average
Age [years]	27	30
Mass [kg] (lb.)	70 (154.3)	76 (167,5)
Height [m] (in.)	1.780 (70)	1.800 (70.8)
Sitting Height [m] (in.)	0.423 (16.6)	0.476 (18.7)
Leg Length [m] (in.)	0.387 (15.2)	0.407 (16)
Thigh Length [m] (in.)	0.432 (17)	0.430 (16.9)

FIGURES

Figure 1

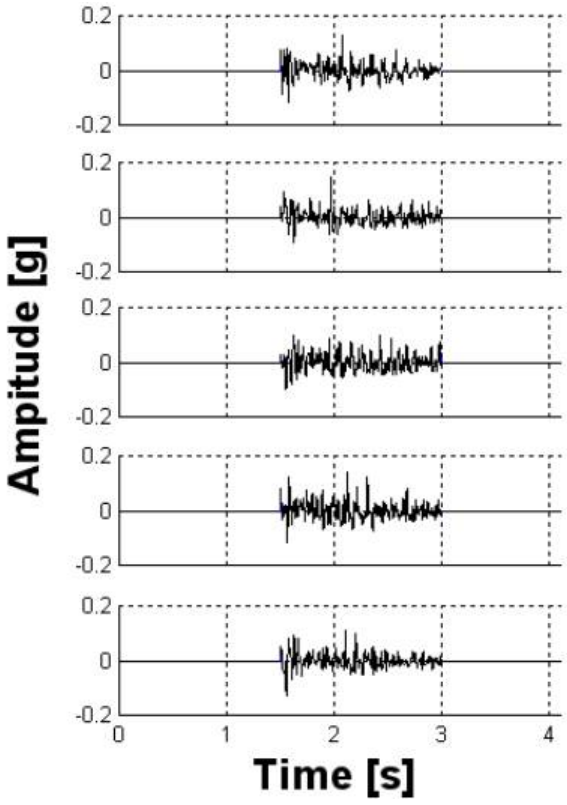


Figure 2

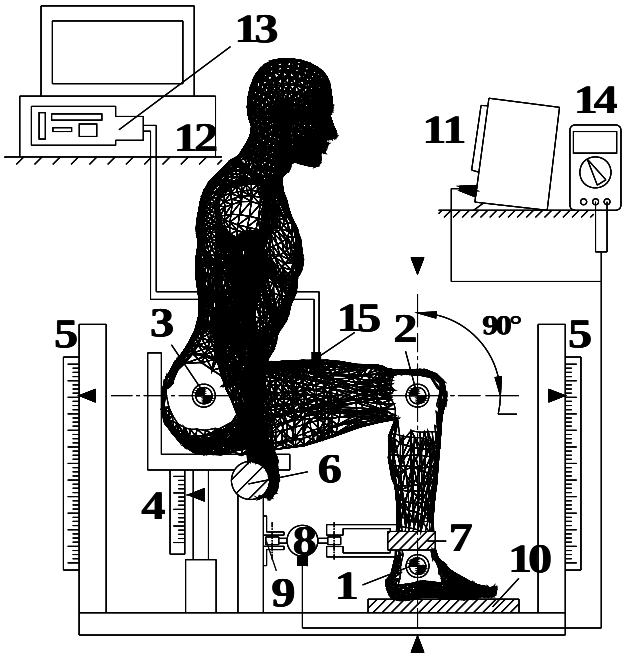
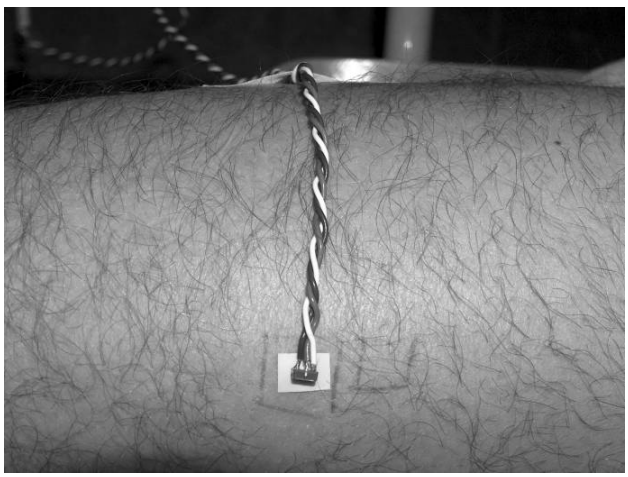


Figure 3



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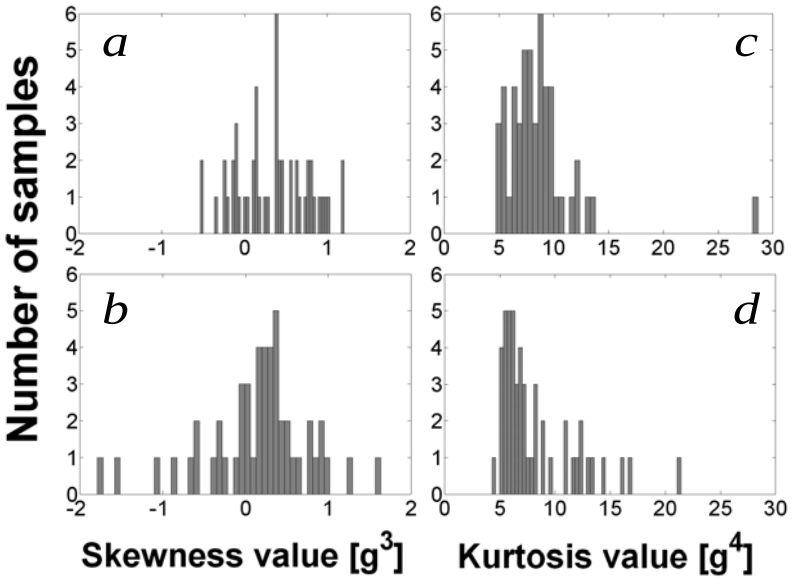

Figure 5

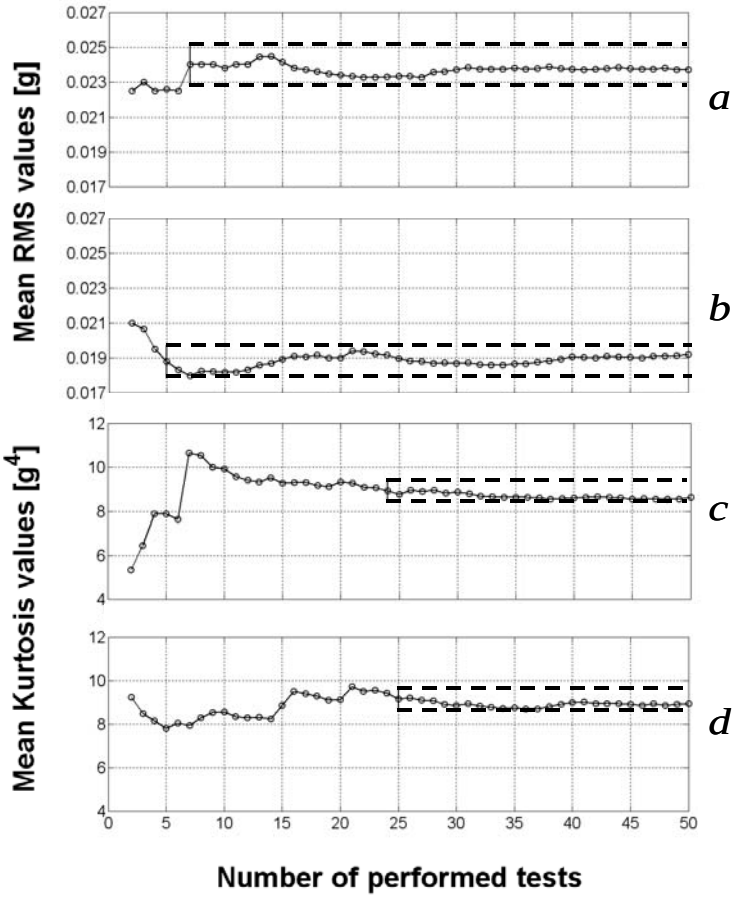

Figure 6

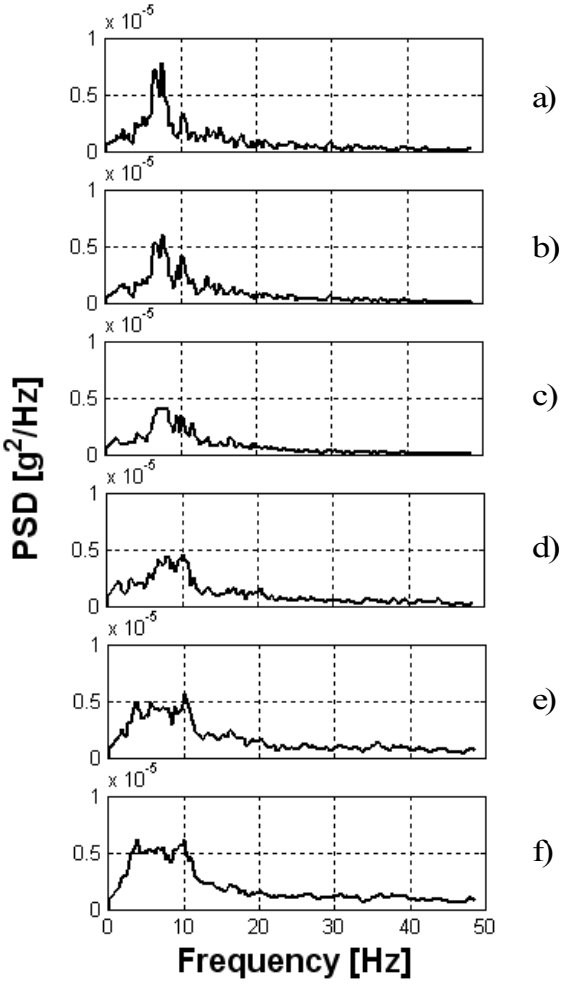

Figure 7