Muscle activity analysis with a smart compression garment

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Abstract

Analysis of muscle exertion while exercising gives insight into an individual’s balance, technique and activity performance. Smart Compression Garments (SCG) function as a low-cost material pressure mapping system integrated within consumer compression apparel capable of assessing both muscle use, and limb positioning. The SCG used for testing contained 5 sensors capable of measuring pressure between fabric and skin above key muscle groups of the lower limbs with marker-based video analysis to determine Knee Flexion Angle. The SCG was calibrated through voluntary contractions of target muscles, where surface pressure range and EMG data allowed for the quantification of exertion levels whilst the participant performed leg extension and flexion activities. Each sensor measured a viable range of pressure relative to the exertion level for each muscle group with a strong repeatable nature and correlation to muscle activation load. Additionally, analysis of the muscle loading variation of the quadriceps and hamstrings whilst walking on a treadmill at low speed was shown to match pre-established gait activation behaviour. Results support that a SCG with low-cost integration of piezoresistive materials has considerable promise in determination of muscle loads and potential injury conditions for the purpose of athlete training support.

1. Introduction

Unlike many other medical injuries, almost 50\% of all sports-related injuries are preventable [1]. The risk of injury and the associated complications reduce the participation and performance across all levels of sport. The muscles and joints of the lower limbs (i.e. Hamstrings, Cruciate Ligaments) are particularly susceptible to

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overload/strain injuries during sporting activities. At a recent conference presentation to leading participants within the Australian sports technology industry, Doctor Peter Larkins, arguably the Australian Football League’s (AFL) most prominent medical consultant highlighted the impending need to solve the major medical problems experienced within elite sport [2]. He placed significant importance on the need to address the ever increasing rate of lower leg musculoskeletal injuries sustained by players at all levels of the code, a statistic that continues to grow each year despite the high financial investments to counteract this trend [3]. It has become apparent that current injury mitigation techniques, employed medical staff and clinical tests are failing to stem the rising injury rate within the elite sporting code. There is an important need to provide a novel and improved solution, not one just for the elite, but also for the sub-elite who are active at least once per week in sporting activities (11.7 million in Australia alone [4]). These participants do not have direct access (both logistically and financially) to the equivalent level of support seen in such developed codes as the AFL, participants whom both during and after their participation sustain a higher percentage of musculoskeletal sporting injuries. This research paper continues the preliminary work in the development of a solution to these problems through a Smart Compression Garment (SCG), a wearable system that directly monitors the pressure response experienced on a compression garment during both muscular and skeletal limb activity [5]. This is capable through the direct development and integration of a low-cost pressure mapping system into a physical garment. Recognition of gross muscular activity is achieved for exercise analysis through the measurement of muscular surface pressures and limb positioning as a means to mitigate injury potential through uneven or overload straining of the activated (or non-activated) muscles. The foreseen benefits realised from research into a SCG stretch beyond the domain of professional sport, to encompass multidisciplinary enrichments to healthcare and rehabilitation sectors.

2. Materials and Methodology

2.1. The Smart Compression Garment

A previously developed SCG prototype by the research team [5] was utilised for its capability of mapping pressure changes over key target muscle groups in the right leg of a wearer. Five of the integrated material pressure sensors were utilised for the test measurements, where each sensor was constructed in-house from a piezoresistive polymer (Rmat2a, RMIT material code), with an individual sensing area of 4cm². Calibration of the sensors was accurately determined ($R^2 > 0.97$) through the correlation of changes in electrical conductance to an applied mechanical load, as measured by a commercial force transducer (Kistler 9317B sensor, Kistler Switzerland).

Skins A400 long tights, an off-the-shelf consumer compression garment, was utilised as the core garment with the sensor placement positioned above the quadriceps (Rectus Femoris, Vastus Lateralis, Vastus Medialis), and hamstrings (Biceps Femoris, Semitendinosus) muscle groups. Measurement of the knee flexion angle ($\theta_{KFA}$) was achieved through the use of marker-based video analysis using the Kinovea visual tracking software (Kinovea.org) and a video camera (GoPro Hero4 Silver, 120fps, 720p, corrected for lens distortion). An electronics module positioned in the waistband provided data collection of the system. The module consisted of a lithium-polymer battery, microcontroller (Teensy 3.1, PJRC.com) with the necessary voltage divider circuits, and a Bluetooth unit for wireless tethering to a computer where data visualisation and logging were performed.

2.2. Estimation of Muscle Force

By selecting an activity where the desired behaviour of the motion is known, an inverse-dynamic approach was utilised to determine the forces involved, as such the muscle force in the quadriceps and hamstrings could be estimated with reliable accuracy. This allows correlation of the corresponding muscle force to that of the pressure variation experienced above the skin at the muscle surface as measured by the SCG. As pressure mapping of the muscles relies upon pressure measurements to infer resultant motion, it functions as a forward-dynamic system, so to assist in validation of results Electromyography (EMG), another forward-dynamic system, was incorporated into the testing methodology.
The basic motions of leg extension and flexion exercises were selected for force derivation, where only the quadriceps or hamstring muscles were activated, and subsequently evaluated. Simple extension or flexion of the knee was performed from a seated (extension, quadriceps loaded) or standing (flexion, hamstrings loaded) position with a weighted shank (5kg mass at ankle). The analysis of the system allowed for the resultant moment about the knee to be calculated, where as it was necessary to account for the complex nature of the forces in the knee joint, earlier derivations by Fuss [6] on knee joint mechanics were used to determine the resultant force required of the activated muscles. Crucial to these calculations was the determination of the $\theta_{KFA}$; this was done through the analysis of video footage with the virtual goniometer function in the Kinovea software package. Distance measurements were taken for the length from the knee’s rotation centre to the centre of mass of the shank ($d_{leg}$) and added weight ($d_{mass}$). This allowed for the further calculation of the moment acting about the knee ($M_{knee}$); a function of the effective moment arm that is dependent on $\theta_{KFA}$, and the estimated total combined mass of the leg ($m_{leg}$) based on the participant’s proportions [7] and the added weight ($m_{mass}$).

$$M_{Knee} = m_{leg}d_{leg}\cos(\theta_{KFA}) + m_{mass}d_{mass}\cos(\theta_{KFA})$$ (1)

As the patella acts as a force multiplier between the quadriceps tendon and patellar ligament [6], the mechanical advantage of the patella ($MA_{Patella}$) at a given $\theta_{KFA}$, along with the moment arm of the patellar ligament ($d_{PL}$), was determined to resolve the knee moment to that of the force produced by the quadriceps (Equation 2). The resultant force produced by the hamstrings through the moment arm of the hamstring tendon ($d_{HT}$) was also determined (Equation 3).

$$F_{Quadriceps} = MA_{Patella} \times \frac{M_{Knee}}{d_{PL}}$$ (2)

$$F_{Hamstrings} = \frac{M_{Knee}}{d_{HT}}$$ (3)

Both relationships resolve to a 3rd order polynomial ($R^2 = 0.999$) that allows for the calculation of the required force in the quadriceps or the hamstrings with respect to the $\theta_{KFA}$ (between 0-90º), holding true where no co-contraction of the quadriceps and hamstrings occurs. For example the force in the quadriceps is solely responsible for the moment about the knee joint when performing quadriceps-only extensions. Equations 4 (extension) and 5 (flexion) are representative of the resultant curve for an adult male (90kg/180cm) performing the test activities.

$$F_{Quadriceps} = 0.0015\theta_{KFA}^3 - 0.2386\theta_{KFA}^2 + 0.3296\theta_{KFA} + 879.86$$ (4)

$$F_{Hamstrings} = 0.0005\theta_{KFA}^3 - 0.0817\theta_{KFA}^2 + 14.792\theta_{KFA} + 0.2663$$ (5)

### 2.3. Placement of EMG and Pressure sensors

To improve the accuracy of the EMG results, the electrode placement was optimised through a 5-point signal analysis test to determine best placement. Comparing the ratio of peak magnitude and median frequency of all five of the measurement sites provided an indication of the optimal location to place the electrodes [8], where signal to noise ratio was maximised, and crosstalk from nearby muscles minimised (Figure 1). The localisation testing was performed on all of the target muscle groups. Following the determination of optimal placement sites, the subject was fitted with wireless EMG electrodes (Zero-wire Cometa Systems, Italy) for each muscle.
Pressure measurements are a result of the mechanical deformation of the muscle, as such are not prone to the electrical crosstalk of neighbouring muscle groups that must be overcome in EMG placement. Achieving a maximal pressure differential drove placement optimisation; as such placement (where possible) was selected to be above the target muscle belly’s centre when under maximal contraction. This aided in maximising the pressure reading and further improving the signal to noise ratio of the measurements.

3. Results and Discussion

3.1. Knee Extension-Flexion tests

An averaged dataset for each muscle group was taken, values were normalised between the maximum and minimum range of measured results and plotted against the measured $\theta_{KFA}$ for both tests. This allowed for direct comparison of both pressure and EMG signals to that of the calculated force loading of the muscle from the performed activity. A curve was fitted to each dataset to define the behaviour of the measured data over the changes in $\theta_{KFA}$ (Figure 2).

The fitted correlation behaviours of both sets of sensor data display a non-linear relationship to that of the calculated muscle load (Equations 4 and 5) for both the quadriceps (Figure 2a) and hamstring (Figure 2b) loading tests, following a clear trend to follow with the increase in muscular loading. The quadriceps measurement for EMG and pressure from leg extensions was fit respectively with a logarithmic ($R^2 = 0.95$) and 4th order polynomial ($R^2 = 0.78$) curve. Hamstring loading required a 4th order polynomial fit for both EMG ($R^2 = 0.94$) and pressure ($R^2 = 0.78$)
0.93) measurements from knee flexion. The presence of a hysteresis loop within the measured pressure data was attributed to the viscoelastic behaviour of the compression garment’s elastic structure; where separate curves could be linked to the loading and unloading segments of the activity. Further measurement into the deformation behaviour of the fabric under repeated loading would need to be conducted to reduce the spread of measured data present in the hysteresis.

3.2. Locomotion tests

Further to the data presented in Figure 2, the relationship between the measured surface pressures to the calculated force was derived for both quadriceps and hamstrings using a 3rd order polynomial fit. The determined forces from the leg extension-flexion tests were used to analyse the variation in muscle force during locomotion. A participant was tasked with walking at 4kph on a treadmill whilst the surface pressures above the target muscles were monitored. Figure 3a displays the averaged surface pressure of each muscle during the stance phase of the walking stride for 29 gait cycles.

Fig. 3. (a) Muscle surface pressure variations and knee flexion angle during stance phase of walking stride, (b) Resultant calculated muscle forces for quadriceps and hamstring muscle groups, (c) Key knee flexion angles during gait cycle
Utilising the relationships determined through the extension-flexion tests, the mean values for the Quadriceps and Hamstring muscle groups were used to calculate the estimated forces during the gait (Figure 3b) where key changes in $\theta_{KFA}$ as components of the stride during the gait were considered (Figure 3c). At initial contact of the foot (1) very little muscle activation is needed, however as the foot becomes loaded with the body weight, $\theta_{KFA}$ increases resulting in the quadriceps being engaged to control the loading against the movement of the treadmill belt (2). The consistent driving force of the stride from the hamstrings is clearly visible throughout the gait, peaking just after heel-raise (3), lessening through to toe-off (4). These findings follow the accepted understanding of walking biomechanics [9]; however one notable consideration is that the results were obtained whilst walking on a treadmill and the missing acceleration of the body would need to be taken into account.

Another important consideration if these methods are to be used for dynamic (concentric and eccentric) contractions is that with the isokinetic movement of the limb segment, the point of both baseline and maximal muscular pressure will change. A basic example of this can be seen in the behaviour of the Biceps Brachii, where on contraction of the muscle and flexion of the elbow, the point of maximal pressure shifts proximally along the muscle. This behaviour is also present during the movement of the relaxed muscle belly mass during flexion of the arm by an external force, resulting in a false reading of muscle activity above the baseline pressure. Further investigation will be necessary to measure the dynamic changes in baseline pressure due to different flexion angles during non-exertion of the muscle. Ongoing developments in the SCG will require additional pressure sensors along the muscle body, and the signal processing altered to adjust for detection of this moving pressure centre.

4. Conclusions

Each pressure sensor successfully measured a viable range of exertion for each muscle group investigated. The summed data collected from the pressure and EMG signals allowed for confident estimation of muscle exertion levels of the participant during the stationary and dynamic exercises. Although only preliminary relationships were drawn between the two forward-dynamic systems (EMG and Pressure mapping), the correlation technique was largely used for comparison of the measured results to that of the inverse-dynamic approach taken to determine the muscle loads. Preliminary correlation of pressure to force estimation during walking was successfully shown through the measurement of muscle pressure throughout the stance phase of the gait. Further research of the smart garment will look into the relationship of these two systems in greater detail, with a focus upon improved fit accuracy along with the inclusion of a matched biofeedback signal capable of alerting the wearer to the condition of the muscles, such as the loading strain and co-contraction activity. This research continues to show that further development into the SCG concept promotes an innovative and smart solution for wearable sports technology, aiding in the training, performance assessment and physical welfare of participants in physical activities.

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