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Design and development of the sEMG-based exoskeleton strength enhancer for the legs

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Abstract

This paper reviews the different exoskeleton designs and presents a working prototype of a surface electromyography (EMG) controlled exoskeleton to enhance the strength of the lower leg. The computer aided design (CAD) model of the exoskeleton is designed, 3D printed with respect to the golden ratio of human anthropometry, and tested structurally. The exoskeleton control system is designed on the LabVIEW National Instrument platform and embedded in myRIO. Surface EMG sensors (sEMG) and flex sensors are used coherently to create different state filters for the EMG, human body posture and control for the mechanical exoskeleton actuation. The myRIO is used to process sEMG signals and send control signals to the exoskeleton. Thus, the complete exoskeleton system consists of sEMG as primary sensor and flex sensor as a secondary sensor while the whole control system is designed in LabVIEW. FEA simulation and tests show that the exoskeleton is suitable for an average human weight of 62 kg plus excess force with different reactive spring forces. However, due to the mechanical properties of the exoskeleton actuator, it will require an additional lift to provide the rapid reactive impulse force needed to increase biomechanical movement such as squatting up. Finally, with the increasing availability of such assistive devices on the market, the important aspect of ethical, social and legal issues have also emerged and discussed in this paper.

Keywords: leg-exoskeleton; electromyography based exoskeleton; LabVIEW myRIO; ethical, societal, and legal concerns.

I. Introduction

Assisted exoskeleton technology seems to be still in the development stage, and needs to be improved to meet the individual needs [1]. Some examples of this exoskeleton are Berkeley Lower Extremity Exoskeleton [2], Raytheon XOS2 [3], Exosuit [4][5], DARPA Soft Exosuit [6], Ekso Bionics [7] and many more [8]. These full body exoskeletons all have some limitation with their design, such as the lack of appropriate power supply suitable to their target specifications. Specialised exoskeleton suits come in many varieties which are used as an aid for medical rehabilitation [9][10], industrial [11], military [12] and commercial [13]. These specialised exoskeletons allow individuals with any lower limb weakness including those who are paralyzed below to move and mimic the biomechanical movement of walking. Any technology that can reduce casualties or enhance one's survival in a harsh environment and open new strategical advantages will always find itself in military use. The concept of specialised exoskeletons is specific to one job but limits the energy consumption to a great extent. On the other hand, the whole body's exoskeletons require a far greater power supply as the exoskeleton is needed to carry the weight of other parts of the suit that might not be used [14]. Such technology may not be popular compared to a humanoid robot in the future, as it is specifically designed for medical and military use and is also expensive and far from generalizing for the general public. It is also impractical for daily life use as it is designed for a harsher environment than the civilian population or is made to help and strengthen one's need for biomechanical movement. Military concept exoskeleton is generally more developed according to the harsher environments [15]. Potential commercial uses are also considered in this review for completeness [16].

A good example of exoskeleton suit that is used medically is the Lifesuit prototype. A man called Monty K. Reed, broke his back due to a parachute accident, created a Lifesuit I prototype in 1986 [17].

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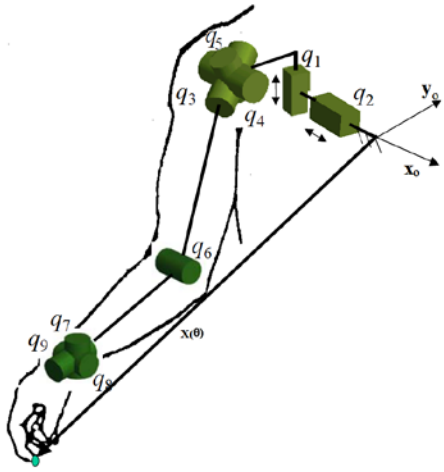


Figure 1. 9-DOF of the human upper limb [72]

His idea of a powered exoskeleton emerged when he was reading Robert Heinlein's spaceship during his time in the hospital while recovering from his injuries. Reed then demonstrated his findings at the university of Washington Engineering Day event and had set a world record for the 8-inches high jump and a land-speed distance record for walking 5 km in powered exoskeletons in 90 minutes at Saint Patrick's Day Dash 2005 [17]. The LS12 Lifesuit prototype used to manage the record also caused some disadvatages to the pilot user. Over time the material used to make the Lifesuit had worn out and became loose, causing minor injuries to the pilot's left outer thigh. The previous prototype also caused some minor injuries when the pilot conducted the experiments. However, the potential risk had been removed and improved with successors [18]. Finding initial risks early within the project can be beneficial as the design can be improved. The lifesuit prototype exoskeleton framework became more ergonomic and more user-friendly as sophisticated systems are improved and added, for example, using the pneumatic power supply, pneumatic actuator and handheld controllers [18]. The handheld controller concept has pro and cons as the user's hand is being mastered but this gives the pilot some manual control over the pneumatic actuators.

Mechanical robotic rehabilitation suits can be divided into the upper limb and lower limb usage [19]. Exoskeletons for upper limb have a shared structure that mimics the human upper limb (Figure 1 and Figure 3). Since the exoskeleton is attached to several upper limb locations difference in a human size, it makes it difficult for the robot to adapt [20]. The upper limb exoskeleton can also be arranged to help certain muscles during rehabilitation by regulating and algorithmic combination that adapts to the forces applied by the exoskeleton to the end user's arm. Most of the exoskeletons work around the biomechanical mechanism of the human flexion elbow movement and shoulder spherical movement [21]. On the other hand, some research on exoskeletons has also included the wrist movement and hand grasping movement [22][23]. Some examples of the upper exoskeletons are Armin 3 and IntelliArm [24]. These exoskeletons have an integrated design 3-DOF to help shoulder depression and elevation movements. The Medarm's exoskeleton has included depression, elevation, retraction and protraction actuator system, while other designs have utilized passive DOF to support the ankle. Passive DOF helps the ankle to move and allows greater freedom of movement, and this minimizes the generated actuation force that is given at the joint [25][26][27]. Lower limb exoskeletons (Figure 2 and Figure 4) focuses more on the ankle rehabilitation. The most common problems addressed in ankle rehabilitation studies is the gait pattern of the patient as the exoskeleton systems manipulate the applied force to improve the gait pattern of the end user [28]. The generalised design of robotic devices provides the actuated motion that affects the foot plantar flexion and

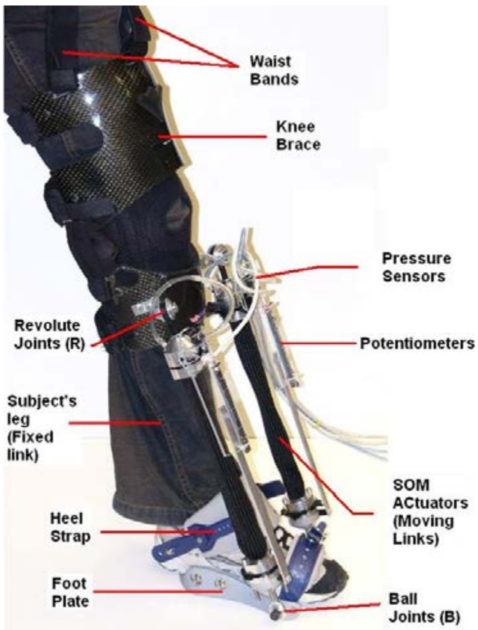


Figure 2. Robotic gait trainer [73]

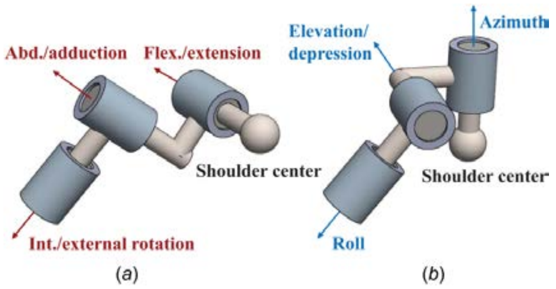


Figure 3. Two rotation conventions for the glenohumeral joint model: (a) flexion–abduction–rotation and (b) azimuth–elevation–roll [29]

dorsiflexion. On the other hand, some devices include passive and/or controlled inversion and eversion movements. Stiffness control of an actuator is commonly used rather than actuators that provide a massive amount of assistive force, but this is mainly due to the biomechanical movement of the ankle joint [29].

Military categorised exoskeletons are more generalised to the entire human body (for example, Raytheon XOS clothing), unlike the medically categorised exoskeletons that are specific to certain key elements of the human body that focuses on say for example only the ankle and not the whole of the human body [30]. The Raytheon Sarcos's XOS2 robotic suit is roughly 50 % more energy efficient than the XOS1 and weighs around 95 kg. The structure is built with high strength aluminium and steel alloy and utilizes actuators, controllers and sensors to perform the required task [31]. The exoskeleton can take a heavy object with a ratio of 17:1, and this is due to the high-pressure hydraulics used, but this again increases the overall weight of the exoskeleton itself. The system analyses the user's limb movements and range awareness so it does not cause damage to itself. This prevents damage from unwanted movements such as sneezing and coughing. The motors have multiple speeds to overcome and produce the appropriate speed and power. Although, XOS2 is more energy efficient than its predecessor but it still has the limitation of the power source. The only power source provided to the system is through a wire tether that connect to the outside power supply. Supplying it with an expensive on-board battery can be violated on the battlefield and can cause friendly casualty [32].

Figure 4. Anklebot [28]

This paper proposes a low-cost and reasonably simple exoskeleton design with a focus on only assisting the user's lower body. The paper is organized into six sections. Section I introduces the exoskeleton designs in general. Section II provides a brief description of exoskeleton aspects, and especially surface EMG (sEMG) control, which is used as a primary sensor in the proposed design. Section III details the proposed lower body one leg sEMG-controlled exoskeleton. Section IV discusses the evaluation/testing of the proposed system. Section V discusses the significant aspects of ethical, social and legal facets in new robotic technologies. Section VI concludes the paper with a brief summary.

II. EMG-based Exoskeleton Control

There are several methods for moving and exoskeletons manoeuvring, however, the most sought-after concept is the use of electromyography (EMG) [33][34][35]. EMG signals can be classified into two types: intramuscular EMG signals, detected from inside of the muscles; and surface EMG signals (sEMGs), detected from the skin surface [36]. EMG-based exoskeletons are usually designed with muscles that are easily accessible from the skin surface. For this reason, sEMG electrode circuit is used in this work. The sEMG-based system works by recording and processing the myoelectric signals from the user so that they can communicate and control the actuators. sEMG signals of flexor Digitorum Superficialis from the finger flexure and Pollicis Longus from the tumb flexure are commonly used as control (actuation) signals. Among the muscles that are part of the upper/lower body limb exoskeleton [37][38], the muscle signals mentioned above are commonly used to create and control an exoskeleton or assistance robotic device that involves hand movements. This is due to the grasping movement, lower noise and the potential ergonomic value, as electrodes can be mounted on the forearm [39]. A better understanding of human anatomy, electrode placements, and the basic principles of muscle contraction can be found in Peter Konrad *et al.* research [40]. Placement of the electrodes from the main voluntary muscles enables control to the specialized exoskeleton. Usually using voluntary EMG muscle signals that are partners with the right part of an exoskeleton creates far more natural movement with respect to the biomechanical movement of the human body. Exoskeleton or auxiliary robots can also work without the correct paired muscle such as the case with amputated personnel [35]. Different muscle types can be trained and used to mimic the EMG signal required for natural movements. For example, a person without fingers can make a robot to capture an object by using the EMG from different muscle group that does not interfere with other signals [39][41][42].

Zaheer *et al.* [43] discuss the sensor site for ideal electrodes placement based on the results of the normalised motor unit and skin thickness. These concepts of the ideal electrode placement sites are located between the centre mass and the muscle

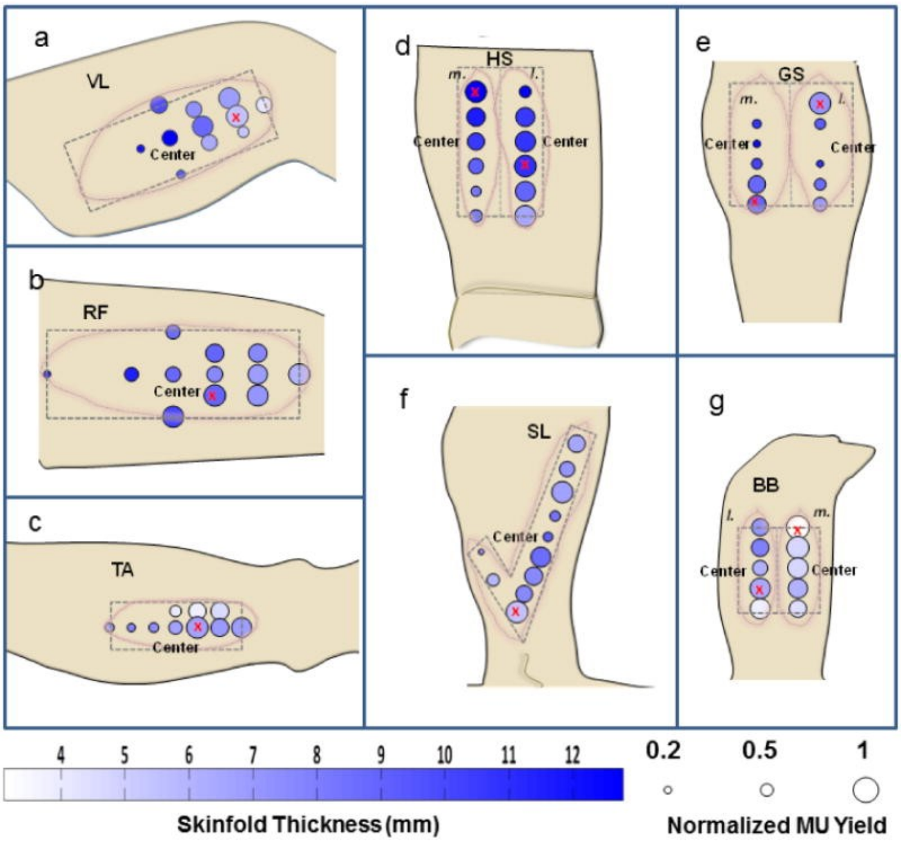


Figure 5. Electrode placement locations from the seven tested muscles topographically mapped by the normalized MU yield per sensor site with increasing circle sizes reflecting greater yields. Average skinfold thickness is indicated by the hue of the color. The values for each muscle are as follows: (a) Vastus Lateralis: the normalized MU yield ranges from 0.3 - 0.9 and the skinfold ranges from 4 to 12.6 mm; (b) Rectus Femoris: the normalized MU yield ranges from 0.3 - 0.8 and the skinfold ranges from 5.9 to 12.4 mm; (c) Tibialis Anterior: the normalized MU yield ranges from 0.4 - 1 and the skinfold ranges from 3.3 to 6.7 mm; (d) Hamstrings Medial: the normalized MU yield ranges from 0.4 - 0.9 and the skinfold ranges from 7.8 - 11.5 mm; Hamstrings Lateral: the normalized MU yield ranges from 0.5 - 0.9 and the skinfold ranges from 6.4 to 12.5 mm; (e) Gastrocnemius Medial: the normalized MU yield ranges from 0.3 - 0.9 and the skinfold ranges from 6 to 12.6 mm, Gastrocnemius Lateral: the normalized MU yield ranges from 0.3 - 1 and the skinfold ranges from 6 to 12 mm; (f) Soleus: the normalized MU yield ranges from 0.2 - 0.9 and the skinfold ranges from 5.4 – 9 mm; (g) Biceps Brachii Medial: the normalized MU yield ranges from 0.7 - 1 and the skinfold ranges from 3.1 to 6.7 mm, Biceps Brachii Lateral: the normalized MU yield ranges [43]

tendency area (Figure 5 for a mapped area). The signal to noise ratio of the detected sEMG signal correlates with the motor unit yield. The signal to noise ratio is inversely related to the muscle fibres and thickness of the subdermal cell tissue between the sensor and muscles [43]. However, the concept of ideal electrode placement varies in different muscle groups [44] as well as from one subject to the other. Inter-subject and intra-session variation is common knowledge in EMG study [36][45]. Subcutaneous fat has also been understood to inhibit sEMG signals [46][47], resulting in a lack of sEMG compared to the invasive EMG needle method [48]. However, the location of this ideal sensor sites by F. Zaheer *et al.* [43] is slightly different from the preferred sensor site of the kinesiology EMG studies [49][50]. The general kinesiology EMG studies are performed using stem electrodes with the intention of acquiring global muscle activities [50].

This study also confirms that reasonable motor unit results can be obtained from almost anywhere in the central mass of the selected muscle group. On the other hand, the results that muscles have localised regions that provide greater motor unit yields are likely related to variations in the EMG signal resulting from the subdermal tissue throughout the muscle surface, and the quality of electrode contact of the sensor and the skin. This also shows evidence that correlates the direct relationship between the motor unit results and the signal to noise ratio of the EMG signal. The poor sources of the sEMG signal are likely due to increasing distance between the electrode sensors and the muscle due to the subdermal tissue, which reduces the sEMG signal amplitude. However, based on some of the ideal electrode placement site results, the relationships of decreasing signal to noise ratio and increasing subdermal network seems inconsistent. Therefore, some other factors may influence the motor unit results such as the muscle innervation zone.

Studies also show that the sEMG signal read in the skin area near the innervation zones produces signals with lower amplitude, resulting in a lower signal to noise ratio due to the cancellation of the action potentials moving in the opposite directions [42]. In summary, the sEMG signal near the muscle innervation seems to have higher frequencies and lower amplitudes. Every area of the human body provides either adequate or poor EMG signals that can vary from person to person. However, certain biomechanical movements of the muscles have preferred sites that provide richer motor unit results. They are generally located between the centre of the muscle mass when contracted and the tendinous area. Another ideal electrode placement is in the area of the skin with the thinnest subdermal tissues. Therefore, the electrodes placement will be located in the area as shown in Figure 5 of vastus lateralis and rectus femoris.

III. Proposed sEMG based Exoskeleton

A. Overall design and software

The development of exoskeleton size depends on anthropometry [51] i.e., the physical measure of human size. For the proposed design, the leg to body height ratio of 49 % with respect to the average male size of 175 cm is considered. The research also includes ±1, 2 or 3 mean value bases of the 49 % comparison to different racial origins. Thus, the most appropriate exoskeleton size is based on the average human height. Smaller exoskeletons can be made to adapt to the average size of the female. Therefore, the size of the exoskeleton is made on the basis of 85.75 cm due to the 49 % of the male average of 175 cm.

The proposed exoskeleton (Figure 6 to Figure 9) consists of flex sensors and actuators that are controlled by using the NI myRIO (LabVIEW) [52] based control system (Figure 10). The start-up sensors used is from the BITalino hardware and software tools kit [53]. BITalino hardware and software are specifically designed to read body signals such as electrocardiography (ECG), sEMG and many more; and has a configurable sampling rate of 1, 10, 100, and 1000 Hz [54]. Flex sensors are used in the development of the exoskeleton as a secondary controller while maintaining the BITalino sEMG sensor [55] as the primary sensor. The sEMG sensor cannot be removed and provides bipolar differential measurement. The raw sEMG signals should be filtered using appropriate techniques such as Savitzsky-Golay (SG) [56], or advanced techniques as Recurrent Quantum Neural Network [57][58] and many others [59][60][61]. In the current work, raw sEMG signals from the BITalino are sampled at 100 Hz, filtered and refined using the SG convolution filter concept. Flex sensors are installed at the top of the knee joint to read the user's posture, create a condition for the control system software, and appear to provide information with a good level of precision, reliability and repeatability [62]. The default analogue value that is read then sent to a state condition that provides Boolean control for the system. The state condition is calibrated to fit the user's sensitivity preference over the exoskeleton, and provides the user's posture state. Depending on the posture, different numeric polynomial sequences and side points are provided for the SG filter LabVIEW program. Memory shape materials can be used as actuators for exoskeleton systems because it does not require external or onboard functioning energy, thereby increasing energy efficiency [63][64]. There are numerous forms of memory materials that can be used, but for the proposed exoskeleton, a tension spring is used as it mimics kinetic movement properties of the pneumatic air muscle when contracting to support the user when squatting and storing the energy. The stored energy is subsequently used to support the user to squat up. A mechanical ratchet and pawl (designed in-house) (Figure 6, Figure 7) is used to hold down the extension of the tension spring from releasing the stored potential energy.

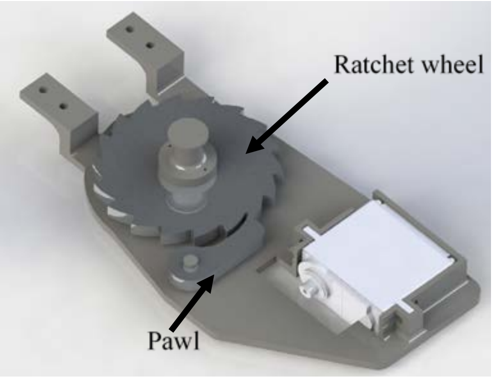


Figure 6. Ratchet and Pawl

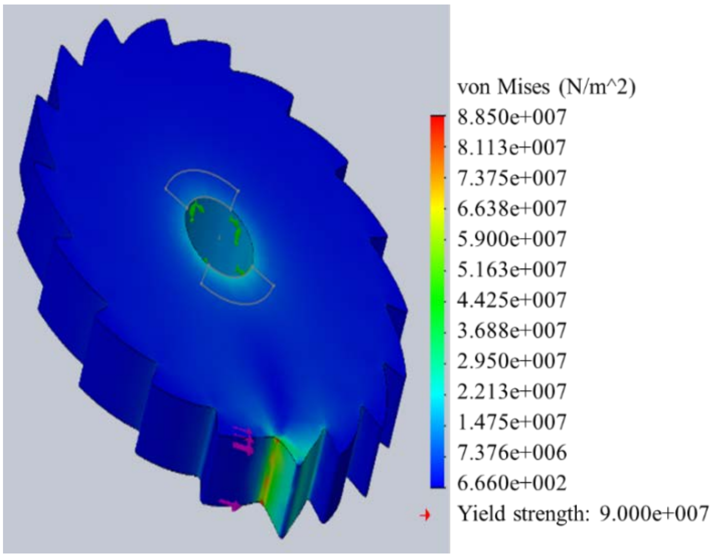


Figure 7. FEA simulation of ratchet gear

Figure 8. CAD prototype of the exoskeleton

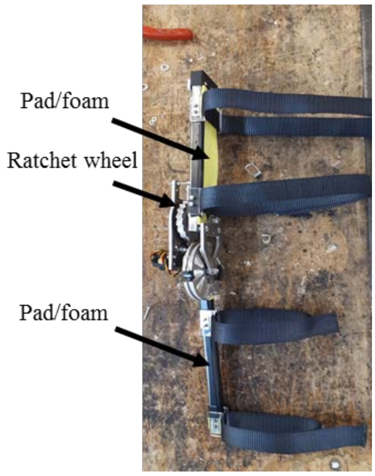


Figure 9. Physical prototype of the exoskeleton

The CAD design (Figure 8) incorporates the rapid prototyping of a grey 3D printer. The ratchet turns as the user squats down and is held by the mechanical pawl. The design concept is to control the reactive force of a spring. The pawl is controlled using a servo motor. The position of the servo motor changes according to the user proclaimed by the system after user calibration. The exoskeleton structural framework is made using aluminium metal to stop potential electrical problems. The good advantage of using aluminium for developing the exoskeleton is due to the machinery available to be used and also the anti-rusting nature. The metals is cut using a water jet cutter that immerses the material in an aqueous environment.

Carbon fibre is planned as the final material to be used for commercial development of the exoskeleton. The red spring is installed in-between the aluminium frames to concentrate the spring strength and share the load between the frames (Figure 8). The complex 3D parts that are manufactured using the 3D printer had weak tensile strength, therefore a material change is needed. Improving the previous prototype design, the revised design incorporates layered sheet build due to the water jet cutting machine is only able to cut flat aluminium material. The 1060-H14 grade aluminium alloy is used as the main body material in the production of the exoskeleton as it is widely recognized for its excellent corrosion resistance, high durability and highly reflective blue/silver appearance (Figure 9). The strap has been added to attach to the exoskeleton, and is locked and joined using Velcro to allow different leg sizes. The yellow soft foam padding is added in a place where the exoskeleton will make surface contact with the user. This creates a soft feeling for the user instead of cold metal and also improve the ergonomic design.

When the parameters set for the sEMG signal are triggered, a Boolean output is given. The condition parameters are set to check the value provided by the flex sensor. The result is two Boolean outputs telling the control system that the user's posture must stand or sit on the base of the angle range of the knee. The Boolean output results are sent to the state condition to control the servo position (Figure 10).

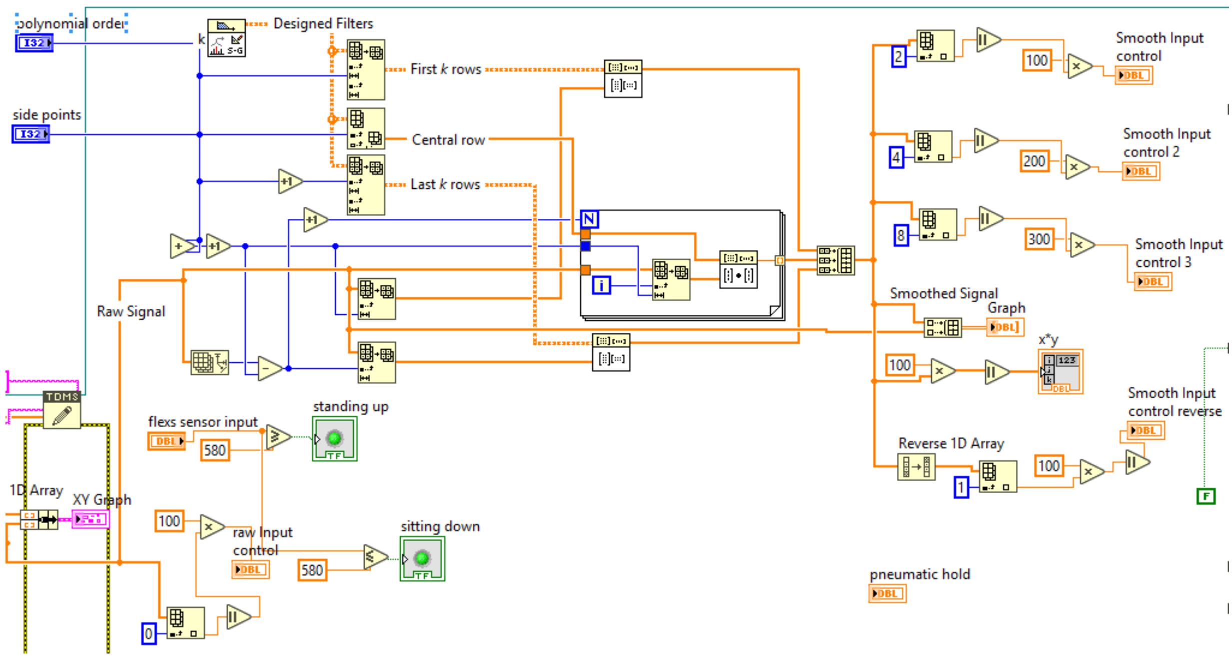


Figure 10. NI LabVIEW control system model

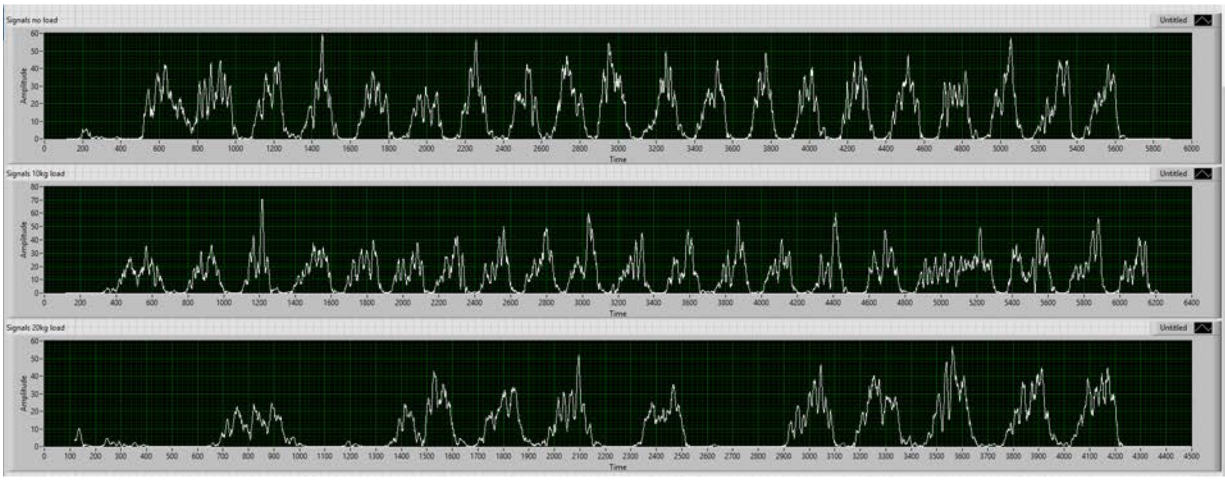


Figure 11. Squat sEMG signal of the quadricep with different spring load *x*-axis [100 = 1 second] *y*-axis [100mV]

IV. Testing and Evaluation

A. Structural test

Ratchet is the only mechanical part that holds the reactive force of the spring potential energy. The 1060-H14 aluminium alloy has a shear modulus of 2.63+010 N/m^2. The force applied on Finite Element Analysis (FEA) simulation of the ratchet gear is set on 607.6 N. The average human weight is 62 kg = 607.6 N. To ensure the system can handle enough spring force to carry the weight of the average human body plus the excess force of overweight users. The FEA simulation shows the highest 8.850e+007 von Mises of the ratchet gear is way below the shear modulus of the 1060-H14 aluminium alloy (Figure 7).

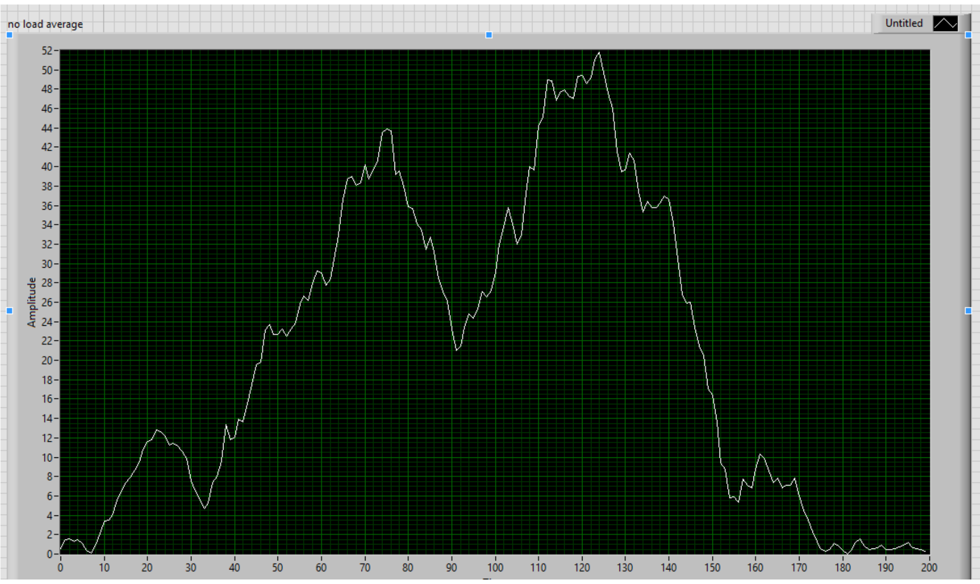
B. sEMG signal based on different spring reactive force

'No-load signal' is the exoskeleton fitted on the user while adjusting the reaction force of the spring to eliminate the weight of the exoskeleton. This establishes the initial basis to show that the exoskeleton can compensate for its own weight without providing support value. Both '10 kg signal load' and '20 kg signal load' are defined as additional reactive force with the additional negative reactive force required to nullify the weight of the exoskeleton (Figure 11). To create a simple average of quadricep sEMG activity while squatting requires some categorization. 10 samples close to the specification and classification are used to create this average value of the sEMG signal shown in Figure 12. The classification of data works by collecting the array of amplitude values from the start of the squat to the end of the squat in a certain amount of time and between breaks. The rest time between each repetition of squat is 4 seconds which is double the time of the 2 second full squat. Once the classification of the data is done, each data set is layered on top of one another to find the most suitable one.

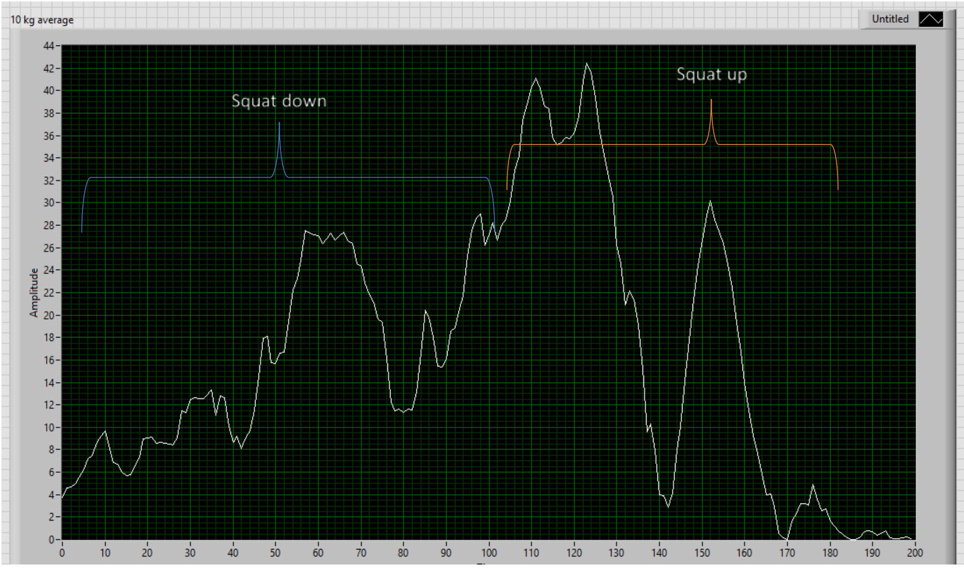
Based on the sEMG signal graph of 'no-load signal' shown on Figure 12(a), it shows; four amplitude spikes: the first and second amplitude spike is located at 0 to 90 *y*-axes when the user is squatting down, as expected due to the quadriceps and hamstring muscle support the body to descend. Once the user reached the target, squat down the legs muscle pass most of the load to the gluteus maximus as shown on the sEMG amplitude which drops at 90 to 105 *y*-axes. On the other hand, the third sEMG amplitude spike up is located at 105 to 132 *y*-axes, this is due to the user squatting up. The body requires a massive workload impact force to move the body weight against gravity. The muscle relaxes as it reaches its baseline point/standing, as shown at 140 to 200 *y*-axes. Once the user stands up, the weight load is transferred parallel from the leg muscle on the bones to the foot.

The comparison of the average sEMG signal with different reactive spring force shown in Figure 12 shows; exoskeleton supports the user well as the strength of body weight pushes the legs down due to squatting. There is an overall lower average sEMG amplitude as the reactive force provided by the spring actuator increases. As the reactive force of the spring increases, the time needed for the user to squat down also increase as the downward force is damped by the spring. On the other hand, the squat up process time is reduced as the stored potential energy inside the spring helps the user to stand up. Therefore, as the spring reactive force increases, the working time of a full squat increase in the process. The actuator memory form material has a good property of absorbing the body weight strength. In the first half of the sEMG signal, where the user squats down, the sEMG signal indicates a lower sEMG amplitude value as the spring reactive force increases.

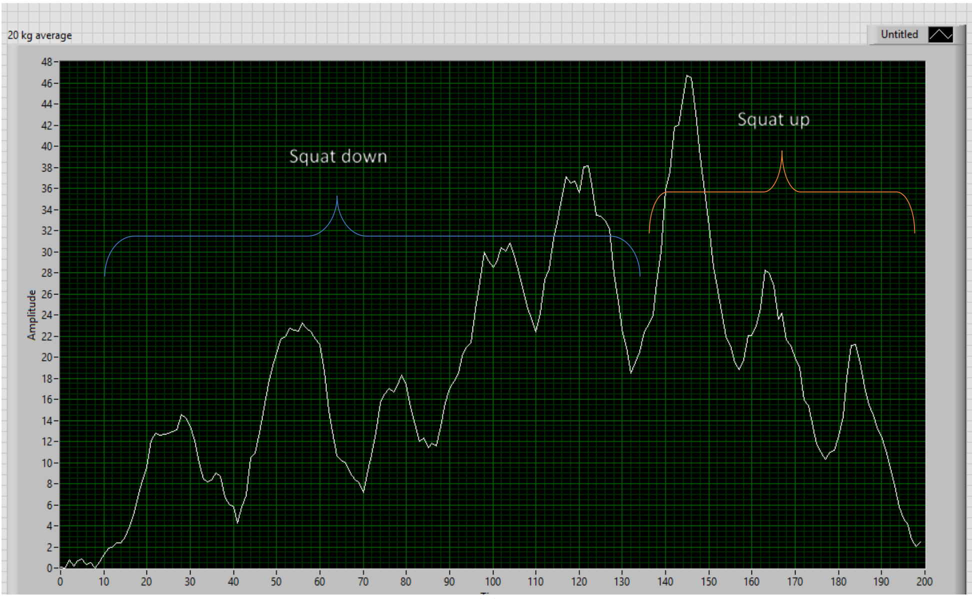
Due to its mechanical properties, the memory shape actuator fails to provide a quick impulse reactive force required to increase the squatting biomechanical movement. An additional actuator is required to compensate for the additional reactive force to increase the squatting biomechanical movement. Adding more shape memory foam to the system can provide a bigger reactive force impulse. However, the process flow chart would change to accommodate the biomechanical time and movement that needed to convert more kinetic force from squatting down to potential energy stored in the spring shape of the memory.



(a)



(b)



(c)

Figure 12. sEMG signal with varying load reactive forces (a) No load reactive force; (b) 10 kg reactive force; (c) 20 kg reactive force. x-axis [100 = 1 second], y-axis [100mV]

V. Ethical, Societal, and Legal (ELS) Aspects in Wearable Robotics

According to Salvini [65], in the coming years, the western societies will have population aged of 60 more than younger people and, to make things even worse, the family caregivers are no longer willing to look after their older relatives, thus obliging them to use wearable robots. This is where assistive devices such as the one discussed here can play an important role. However, with the increasing availability of such assistive devices, an important aspect of ethical, social, legal and standardization aspects have emerged [66][67][68][69]. A comprehensive and a very recent work on ELS issues in Wearable Robotics, identifying relevant values and ethical, philosophical, legal and social concerns related to the design, dissemination and practical use of wearable robots can be found in Felzmann *et al.* work [69]. There are several other works in this field such as that by Greenbaum *et al.* [70], which talks about the specifics related mainly to exoskeleton designs. Greenbaum *et al.* also raises an important open question as a way of dealing with the high costs of exoskeletons in relation to social justice of access/affordability for all who need it (especially with the increasing ageing population), as well as the dependence on expensive technologies that eventually occurs. It seems that for the wider society, exoskeletons and other technological enhancement raise much longer and complex questions that will force human to redefine how human themselves are being perceived [70]. Calo in [71] examines very well and probably for the first time what is the meaning of the introduction of equally transformative new technologies for cyberlaw and policies regarding integrating robotics and such new technologies.

VI. Conclusion

The sEMG-based exoskeleton concept of using a spring as a mechanical actuator works well but it is limited to the energy a spring can store. The memory shape material actuator has a good property of absorbing the body weight force while supporting the user to squat down. On the other hand, the time a person takes to squat down increases as the downwards force is dampened by the spring, slowing down the full squatting action. The amplitude value of sEMG control signal does not have a repetitive arrangement value over time as the muscle contraction varies with minute changes across the environment and the user. In addition, the sEMG electrode placement varies from subject to subject. The entire area of the leg muscle provides sEMG signals that can be decomposed to produce the firing instances and shapes of several motor unit. However, muscles have preferred a place that provides richer motor unit yields. They are generally located between the centre of the abdominal muscle and the muscle tendon area. These sites are associated with regions where the easily measured skinfolds have the least thickness. Therefore, the required amplitude control is set to a lower range to trigger the system to change the servo position. The squat down process shows the greatest change with the different sEMG signal and increased spring reactive force. However, minute changes occur with sEMG signal of squat up. On the other hand, the implantable surgical subdermal electrode implant may provide reliable data as it will no longer be interfered by the skin and the outer layer fat. There is room for further improvement in using the tension spring as a mechanical actuator such as, optimizing the ideal reactive force of the spring and increase the amount of either the tension spring or longer extension range of a spiral spring. The spring(s) can be charged up to the maximum after two or more squat down then releases the stored potential energy in one squat up. Another improvement is to use various types of advanced actuators to create hybrid exoskeleton consisting of mechanical and pneumatic components. Nevertheless, the proposed sEMG driven mechanical exoskeleton is proven to help users with squatting biomechanical movement, however, it will require further improvement.

Declarations

Author contribution

M. Cenit conceived the original idea. V. Gandhi supervised and revised the work critically. Both authors read and approved the final paper.

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Conflict of interest

The authors declare no conflict of interest.

Additional information

No additional information is available for this paper.

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