# Soft Wearable Augmented Walking Suit with Pneumatic Gel Muscles and Stance Phase Detection System to Assist Gait

## Abstract

Lower limb of the human body is responsible for human locomotion and maintain a good quality of life. However, there are many cases of muscle fatigue or injuries due to the stressful work environment, aging and work that involve walking a long distance. Therefore, there is a need for walking assistive suit which can unload muscle activation during walking and reduce the chances of lower limb muscle fatigue. In this paper we discuss the development of lightweight and wearable Augmented Walking Suit (AWS) using Pneumatic Gel Muscle and its actuation control using lower limb pose detection mechanism by considering human gait cycle. The objective of this assistive suit is to reduce required muscle effort of posterior and anterior muscle during the swing phase of the gait cycle thereby making it easier to move forward. To evaluate the effects of the suit we tested this suit with random subjects and record surface EMG (sEMG) of 8 primary lower limb muscles for two level of assistive forces. The evaluation was done based on the sEMG signal envelope for each subject for a different level of assistive forces and the statistical difference in percentage maximum voluntary contraction (\%MVC) of 8 primary lower limb muscles active during the gait cycle. In our result, we found that all subjects showed no change or a statistically significant reduction in muscle efforts due to assistive suit for all the muscles responsible for swing phase of the gait cycle.

## Introduction

Ability to move uninterrupted is one of the critical function of human body. It is one of the reasons for enjoying a good quality of life by enabling one to be independent for performing a variety of daily tasks. However, there are many instances such as aging, accidents and longer and more stressful working conditions that result in muscle fatigue and injuries making it difficult to walk by affecting the quality of life of the individual. Such situation can be avoided or addressed using exoskeletons or wearable assistive devices. Muscle activation pattern of human gait is dynamic, and changes as the motion or intent are changed, but the basic pattern of gait cycle is same for all. While developing AWS we considered factors such as nature of work area, age, flexibility to use in outside environment, lightweight, portable, easy to use, reduces muscle efforts during walking and no impact on normal gait cycle. With increasing elderly population, stressful work condition devices like these will play a significant role in improving the quality of life. L. Garçon et al. \cite{1} in his review mentioned there are large requirement assistive devices for mobility for people such as elderly, disabled and healthcare staff for various tasks involved in daily life. Among various lower limb assistive devices, there exists tradeoff between autonomous actuation, wearable, lightweight and affordability. HAL \cite{2} which enable walking easier for elderly and rehabilitation post stroke or accidents. Wearable agri robot \cite{3} designed for supporting farming activities and reduce muscle fatigue, it supports body posture and reduces the muscle fatigue. Walking assist device with body weight support system \cite{4} for augmenting walking and assistive squats motion required for pick and place tasks in the various work environment. RoboKnee \cite{5} is one DOF exoskeleton designed to support human locomotion such as walk and stair climbing. Plantarflexion assist exoskeleton \cite{6} is designed to reduce the metabolic cost of walking.

These devices are divided into segments such as healthcare, disability support and augmenting locomotion. These devices augment human motion significantly, but its use in outside environment is limited especially in agriculture and factory settings. For augmented walking wearable, lightweight, portable, easy to use and reduce muscle fatigue, these criteria are essential and together missing in assistive devices discussed above. To solve this problem previously, we developed a lightweight, low powered pneumatic gel muscle (PGM) \cite{7} as shown in Fig \ref{fig:pgm}. PGM can generate force with 60 kPa air pressure which is not possible in McKibben pneumatic artificial muscle (PAM) \cite{8}. It is also structure in a way to be stitched to fabric or fix using velcro tapes; this makes it easy to design the assistive suit. Fig \ref{fig:pgmelongationratio} shows the relation of supplied air pressure, generated force and maximum elongation as a percentage of resting length.

In \cite{9} we devised the concept if Unplugged Powered Suit for walking assist using the advantage of PGM and gait cycle. The actuation control of PGM was designed by attaching pump at the heel of a shoe. This configuration was able to generate minimal assistive force for walking. However, the challenge of this configuration changes in shoe design and placement of pumps in the shoe for generation of assistive force.

In this paper, we discuss the design and control of AWS, which improves on Unplugged Powered Suit (UPS) by keeping human gait in the loop by using gait cycle identification system for generating assistive force. In section \ref{methodology} PGM and its force characteristics, biomechanics and human gait detection system and design and configuration of the Augmented walking suit is discussed. In section \ref{Evaluation}, we discuss the evaluation criteria, experiment method setups, results of the lower limb surface EMG (sEMG) evaluation for two levels of assistive force with the comparison of average gait sEMG envelope for all subjects and statistical analysis. Section \ref{discuss} presents the discussion, conclusion and future works.

## Methodology

### Pneumatic Gel Muscle

PGM is a particular type of PAM designed to be driven by low air pressure. Figure 1 shows schematics and real prototype of the PGM. It has a resting length of 30 cm, maximum contraction length of 25 cm and maximum elongation length of 45 cm. Construction of PGM includes an inner tube made of a special styrene-based thermoplastic elastomer to improve the flexibility, and an outer protective mesh. McKibben PAM has rubber or silicon-based rubber tubes covered with protective mesh; these tubes need more air pressure to inflate whereas in case of PGM can generate force with air pressure as low as 50 kPa up to 300 kPa as reported by [7]. The flexible design and working with low air pressure makes it more suitable choice for development of wearable assistive suits as compared to McKibben PAM who have higher force generating capacity but requires large air pressure. Figure \ref{fig:pgmelongationratio} shows elongation ratio of the PGM as measured by \cite{7} it shows the force-generating capacity of the PGM and elongation length for various level of air pressure. In the experiment, the one end of PGM is fixed, and test load is added to another end. Whereas in AWS both ends of the PGM is fixed and stretched, in this case, the force generating capacity of the PGM changes. This change is not measured in \cite{7}, therefore we conducted an experiment to measure the force generated by PGM for stretched and un-stretched condition and different air pressure. The supported range of air pressure is 50 kPa to 300 kPa. Fig \ref{fig:pgmtest} show experiment setup, where one end is connected to load cell, and at the other end air source is connected through Panasonic ADP5161 air pressure sensor. The experiment is conducted for two cases unstretched and stretched to 45 cm. Figure 3 shows the measured force profile for two conditions in both cases PGM shows linear force generation characteristics which is modeled as a linear equation as described in equation 1 and 2 with their respective $R^2$ values. These models exhibit similar force generating behavior when used in AWS configuration. These characteristics can be used for controlling assistive force generated by PGM when in AWS.

### Biomechanics of Gait Cycle

The design and control of the AWS is based on human walking, i.e. gait cycle and depends on how we walk. The gait cycle is divided into three major phases, i.e. stance phase, double limb support phase and swing phase. The stance phase is responsible for weight acceptance, and load transfer to support swing phase of the contralateral limb, Figure \ref{fig:gait} shows a schematic block diagram of the gait cycle. In stance phase muscle activation of tibialis anterior (TA), rectus femoris (RF), vastus medialis (VM), vastus lateralis (VL), soleus (SOL), medial gastrocnemius (MG) and lateral gastrocnemius (LG) is observed. These muscles are responsible for heel strike till toe off in the stance phase. In the double limb support phase, the limb going in stance phase support the forward locomotion of the contralateral limb going in swing phase. In this phase both the limbs are on the ground for about 10\% of the one gait cycle. In this phase SOL, LG, MG and RF muscles are active and responsible for the limb going in swing phase. In the swing phase limb makes forward movement and RF, VL, VM, biceps femoris (BF) are major muscle contributors of this phase.

Apart from the difficult muscle activation foot position and orientation also changes. In stance phase foots orientation start from the heel strike then flat foot, heels off and ends with toe off. We used this information for gait detection and actuation control of the AWS.

AWS Design and Actuation Control Mechanism

AWS is designed to use human motion to detect and provide the assistive force. In the section \ref{gaitcycle} we talked about using foot orientation in stance phase and the respective motion in the contralateral limb. To use this information an FSR-406 pressure sensor was placed in the shoe to detect flat foot, which is when the assistive force is to be applied to the contralateral limb as it goes into the swing phase. FSR placement is shown in Fig \ref{fig:fsrsole} and Fig \ref {fig:aws} shows subject wearing AWS assistive suit with controller, battery, air tank in backpack. Figure \ref{fig:awssystem} shows control mechanism of the AWS suit with FSR-406 sensor based stance phase detection mechanism and actuation control of the PGM. It is a continuous process of proportional (P) control where Arduino Uno board monitors the FSR sensor data to identify stance phase in the gait cycle. Detection of the gait cycle triggers actuation mechanism of the PGM on the contralateral limb. For actuation control, we used Kaganei G010E1 3/2 normally closed solenoid valve. FSR sensor data is continuously monitored for switching ON/OFF solenoid valves. This system is realized using following equation

\begin{equation}\label{kevalue}

E = R - Y

\end{equation}

\begin{equation}\label{uvalue}

U = kpE

\end{equation}

where $E$ is error signal, $R$ is calibrated threshold value of the FSR sensor, and $Y$ is the analog value of the FSR sensor, $U$ is input to the solenoid valve and $kp$ is the P-gain.

This switching controller is designed to detect stance phase of the gait cycle and generate assistive force only during walking thereby avoiding unwanted actuation in a stationary state. The supplied air pressure is directly proportional to the assistive force. Therefore, air pressure control is done through pressure regulator attached to the compressed air tank.

### AWS Performance Evaluation through Muscle Activation Pattern of Lower Limb Muscles

AWS is designed to reduce muscle efforts during walking by using PGM to provide the assistive force. Required assistive force can be increased or decrease by regulating supplied air pressure. In our experiment performance of AWS was evaluated for the different assistive force. Walking involves a combination of muscle activation dynamics of both anterior and posterior lower limb muscles. These changes are recorded using sEMG signals of eight major posterior and anterior muscles which contribute to the gait cycle. We measured TA, SOL, MG, LG, RF, VM, VL and BF, these are the most accessible and prominent muscles of the lower limb and collectively support gait cycle. The performance of the AWS is measured based on the statistical difference in the sEMG recorded between when the subject is not wearing AWS and two levels of assistive air pressure.

### Experiment Protocol

For practical evaluation of the assisted gait, we need to measure minimum three full gait cycles \cite{10}. In our experiment, we recorded sEMG for ten full gait cycles. It was done by asking subjects to walk 15 m straight by maintaining the walking speed during all experiments. For recording sEMG and FSR sensor data, we prepared a backpack as shown in Figure 7 which includes, AWS controller circuit, P-EMG devices for recording sEMG, laptop (this laptop was remotely operated to log EMG data), portable battery (required for AWS controller and P-EMG device). The total weight of the backpacks is 6 kg.

Total three experiment was performed, the first experiment conducted to record sEMG data for normal gait cycle. The second experiment was conducted by wearing AWS which includes waist support belt, knee support, PGM, solenoid valves, shoe with FSR sensor, air tank with pressure regulator and backpack as described above. In this experiment gait performance was measured without air supply, it was done because the PGM has its own elasticity which provides minimum assistive force. In the third experiment, we measured gait performance by supplying 60 kPa air pressure and the fourth experiment was conducted to measure gait performance when supplied air pressure is 100 kPa. Three iterations of each experiment were conducted to perform statistical analysis of the sEMG evaluation using two-sample t-test method.

Seven subjects participated in the experiment. Information was shared with all the subjects before the experiment. During the experiment, subjects could relax or take a break to avoid muscle fatigue because of carrying heavy backpack during experiments.

### Results

Four experiments were conducted with five subjects to record sEMG of eight major muscles in the right lower limb of all subjects. The recorded sEMG was rectified with iEMG, 2nd order low pass filter with cut off frequency of 100 Hz, 2nd order high pass filter with cut off frequency of 40 Hz using P-EMG plus tool for P-EMG device. Figure 8 and 9 shows comparison of muscle activation for all 4 experiments, the graphs also show stance phase detection for both legs based on the recorded foot sensor data, it was also used for segmenting gait cycles and calculating average gait signal for each experiment. The portion of the graph highlighted in the green is stance phase detection on the left leg which provides assistive force on right leg as it transitions from stance phase to swing phase. The portion of the graph highlighter in the blue shows the gait phase during which effect of assistive force is observed. This graph visualizes the difference in the sEMG signal envelope for normal gait when AWS is not worn and 2 levels of assisted gait with AWS. Significance of changes in sEMG were quantified by running two sample t-test to calculate statistical difference and p-value for significance of the difference in the normal and assisted gait signal for all muscles for all subjects. Figure \ref{fig:tagraph}-\ref{fig:bfgraph} shows averaged \%MVC data for each subject for 3 experiments and their significance individual muscle. Table \ref{ptvalues} shows result of the two sample t-test.

For TA, observation of average sEMG enveloped shows reduction in peak value and sEMG envelop for 3 subjects and the \%MVC comparison shows significant difference in normal and assisted gait ($p-value < 0.05$ and $p-value < 0.01$) except for subject 1 who shows no change at all. For SOL two subject showed significant change in \%MVC ($p-value < 0.01$). For MG no change is observed from the sEMG signal envelope and \%MVC data shows two subject have significant change ($p-value < 0.05$). For LG 3 subjects showed significant difference ($p-value < 0.05$ and $p-value < <0.01$) whereas two subject show reduced \%MVC but non-significant. RF shows significant change for all subjects ($p-value < 0.05$ and $p-value < 0.01$), observation of sEMG shows reduction in signal envelope and peak value for assisted gait. For VM and VL \%MVC shows significant difference between normal and assisted gait ($p-value < 0.05$ and $p-value < 0.01$) whereas subject 3 showed no change in \%MVC of both muscles. For BF subject 1 and 3 shows increased in the sEMG signal peak during terminal swing phase and \%MVC shows significant reduction for assisted gait with 60 kPa assistive pressure ($p-value < 0.05$).

### Discussion

In this paper we discussed development of soft wearable Augmented Walking Suit designed to reduce muscle effort during walking. This suit uses only one PGM for each lower limb for augmenting walking gait. Control of the assistive force is performed based on pressure sensor installed in the shoe. This sensor detects stance phase from heel touch to flat foot, this detecting mechanism helps trigger air valves to generate assistive force for the contralateral limb in swing phase. By placing sensors in both shoes gait phase of the individual limb and contralateral limb can be identified. Performance evaluation of AWS was done based on the statistical difference in the average \%MVC of sEMG signal between normal and assisted gait measured for 7 subjects. From the results we can find that use of AWS has reduced muscle activation pattern during experiment especially in the swing phase as designed.

During swing phase of the gait cycle RF, VM, VL, BF, TA and LG all these muscles show showed significant difference in the \%MVC for most of the subjects. Soleus showed increased in the \%MVC for subjects in case of assisted gait using AWS with no air supply, soleus muscle is active during pre-swing (toe off) phase of the gait cycle. We believe the reason for this is the placement of PGM which creates flexion torque at knee during terminal stance where soleus is responsible for toe off and knee extension during initial swing phase.

### Conclusion

In this paper we developed Augmented Walking Suit and PGM actuation control based on stance phase detection system. Results of performance evaluation experiment showed statistical significant reduction in muscle activation of lower limb muscle. The current mechanism provides assistive force for 10\% to 15\% of the gait cycle during swing phase. The current configuration is lightweight, portable and easy to use. In future work, we plan to devise full gait detection system for detail control over muscle activation, this will allow us to add more PGM in the suit for detailed control over gait cycle and improve augmentation factor of AWS while keeping it lightweight and portable.