Soft Wearable Augment Walking Suit with Pneumatic Gel Muscles and stance phase detection system to provide assistive force

Chetan Thakur[1] Kazunori Ogawa[1,2] Toshio Tsuji[1] Yuichi Kurita[1]

***Abstract*— Lower limb of human body is responsible for human locomotion and maintain good quality of life. But there are many cases of muscle fatigue or injuries due to stressful work environment, aging, working involve walking long distance. Therefore, there is a need of walking assistive suit which can unload muscle activation during walking and reduce the chances of lower limb muscle fatigue. In this paper we discuss about development of lightweight and wearable Augment Walking Suit using Pneumatic Gel Muscle and its actuation control using lower limb pose detection mechanism by taking human gait cycle in consideration. The objective of this assistive suit is to reduce required muscle effort of posterior and anterior muscle during swing phase of the gait cycle thereby making it easy to move forward. To evaluate the effects of the suit we tested this suit with random subjects and record sEMG of 8 major lower limb muscles for three level of assistive forces. The evaluation was done based on the sEMG signal envelop for each subject for different level of assistive forces and statistical difference in %MVC of 8 major lower limb muscles active during gait cycle. In our result we found that all subjects show reduced or no significant changes in muscle efforts due to assistive suit for all the muscles responsible for swing phase of the gait cycle.**

I. INTRODUCTION

Ability to move uninterrupted is one of the important function of human body. It is one of the reason for enjoying a good quality of life by enabling one to be independent for performing variety of daily tasks. But the there are many instances such as aging, accidents, longer and stressful working conditions results in muscle fatigue and injuries making it difficult to walk by affecting a quality of life of 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 is changed but the basic pattern of gait cycle is same for all. While developing our Augment walking suit 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 significant role in improving the quality of life. L. Garon et al [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 [2] which enable walking easier for elderly and rehabilitation post stroke or accidents. Wearable agri robot [3] designed for supporting farming activities and reduce muscle fatigue, it support body posture and reduces the muscle fatigue. Walking assist device with body weight support system [4] for augmenting walking and assistive squats motion required for pick and place tasks in various work environment. RoboKnee [5] is one DOF exoskeleton designed to support human locomotion such as walk and stair climbing. Plantarflexion assist exoskeleton [6] is designed to reduced the metabolic cost of walking.

All these devices are divided in segments such as health care, disability support and augmenting locomotion. These solutions solve the problem in human locomotion due to disability or aging to great level but use in outside environment is limited specially in agriculture and factory settings. For augmented walking wearable, lightweight, portable, easy to use and reduce muscle fatigue, these criteria are important and together missing in assistive devices discussed above. To solve this problem previously we developed a lightweight low powered Pneumatic Gel Muscle (PGM) [7] as shown in Fig 1. PGM can generate force with 60 kPa air pressure which is not possible in traditional pneumatic artificial muscle [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. Figure 2 shows relation of supplied air pressure, generated force and maximum elongation rate as percentage of resting length.

In [9] we devised the concept if Unplugged Powered Suit for walking assist using advantage of PGM and gait cycle. The actuation control of PGM was done by attaching pump at the heel of a shoe. This configuration was able to generate minimal assistive force for walking. But the challenge of this configuration is change in shoe design and placement of pumps in the shoe for generation of assistive force.

In this paper we discussed the improved design and control of Augment Walking Suit by keeping human locomotion in loop by using gait cycle identification mechanism for generating assistive force. In section II PGM and its force characteristics, biomechanics and human gait detection system and design and configuration of the augment walking suit is discussed. In section III we discuss about the evaluation criteria, experiment method setups, results of the lower limb sEMG evaluation for different force levels with comparison of average gait sEMG envelop for all subjects and statistical

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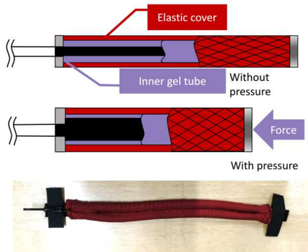

Fig. 1. Pneumatic Gel Muscle schematic Diagram and Real Product

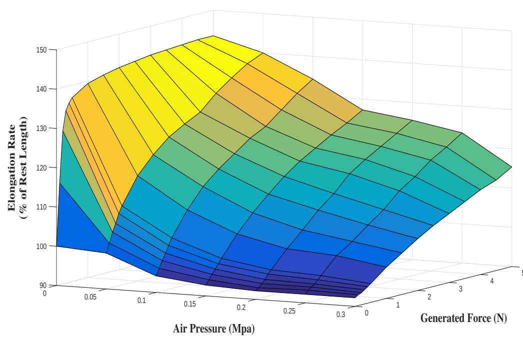

Fig. 2. Pneumatic Gel Muscle's elongation and force generation characteristics [7]

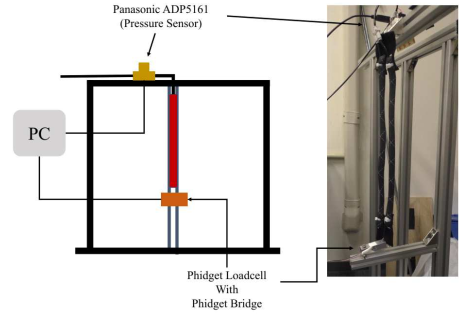

Fig. 3. Force profile test setup

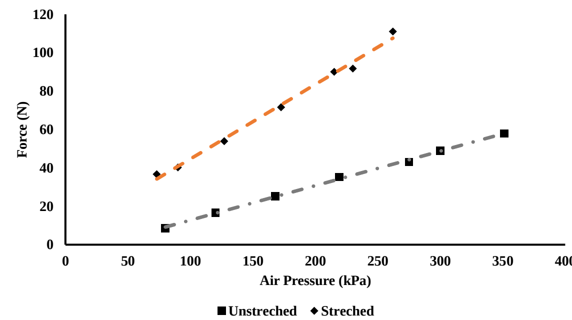

Fig. 4. PGM force profile for unstretched and stretched condition

analysis. Section IV depicts the discussion, conclusion and future works.

II. Methodology

A. Pneumatic Gel Muscle

PGM is a special type of PAM designed to be driven by low air pressure. Figure 1 shows schematics and real prototype of the PGM. It has resting length of 30 cm, maximum contraction length of 25 cm and maximum elongation length of 45 cm. Construction of PGM includes inner tube made of special styrene-based thermoplastic elastomer to improve the flexibility and outer protective mesh. Traditional PAM have 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 traditional PAM who have higher force generating capacity but requires larger air pressure. Figure 2 shows elongation ratio of the PGM as measured by [7] it shows 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 other end to. Whereas in AWS both end of the PGM are fixed and stretched, in this case the force generating capacity of the PGM changes. This change is not measure in [7], therefore we conducted an experiment to measure 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 3 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. 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 PG shows linear force generation characteristics which is modeled as 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.

$$y = 0.1799x - 5.1983; (R^2 = 0.993) \tag{1}$$

$$y = 0.3883x + 5.8899; ; (R^2 = 0.9985) \tag{2}$$

B. 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 5 shows schematic block diagram of the gait cycle. In stance phase muscle activation of tibialis anterior (TA), quadriceps femoris i.e. rectus femoris (RF), vastus medialis (VM), vastus lateralis (VL), soleus (SOL), medial gastrocnemius (MG) and lateral gastrocnemius (LG) is observed. These muscles are responsible from heel strike till toe off in the stance phase. In the double limb support phase the limb going in stance

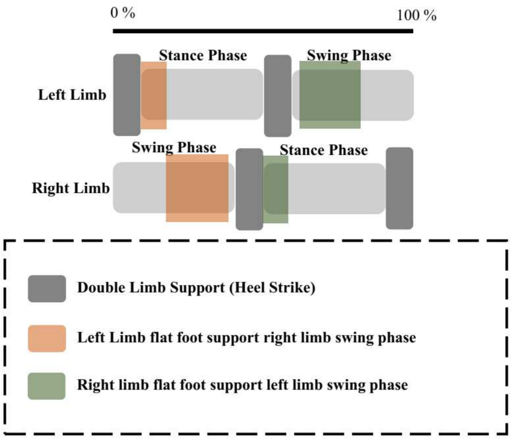

Fig. 5. Gait Cycle Classification

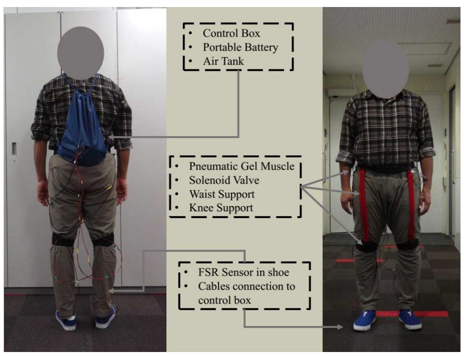

Fig. 6. Augment Walking Suit

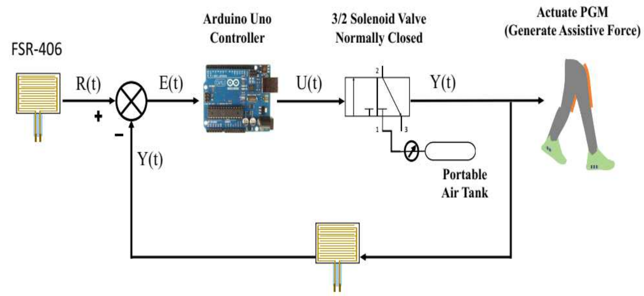

Fig. 7. AWS Control mechanism

Fig. 8. FSR-406 sensor placement in shoe

phase support the forward locomotion of the contralateral limb going in swing phase. In this phase both the limbs are in 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 complex muscle activation foot position and orientation also changes. In stance phase foots orientation start from heel strike then flat foot, heels off and ends with toe off. We used this information to process and use it for gait detection and actuation control of the AWS suit.

C. AWS Design and Actuation Control Mechanism

AWS is designed to use human motion to detect and provide assistive force. In the section II-B we talked about using foot orientation in stance phase and the respective motion in contralateral limb. To use this information we used FSR-406 pressure sensor was installed in shoe and placement was selected to detect flat foot which is when the assistive force to be applied for the limb going in swing phase. FSR placement is shown in Fig 8 and Fig 6 shows subject wearing AWS assistive suit with controller, battery, air tank in backpack. Figure 7 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 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 corresponding 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

$$E = R - Y \tag{3}$$

$$U = kpE \tag{4}$$

where E is error signal, R is calibrated threshold value of the FSR sensor and Y is 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 stationary state. The supplied air pressure is directly proportional to the assistive force therefore, air pressure control is done though pressure regulator attached to the compressed air tank.

III. AWS Performance Evaluation though Muscle Activation Pattern of Lower Limb Muscles

AWS is designed to reduce muscle efforts during walking by using PGM to provide assistive force. Required assistive force can be increased or decrease by regulating supplied air pressure. In our experiment performance of AWS was evaluated for different assistive force. Walking involves 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

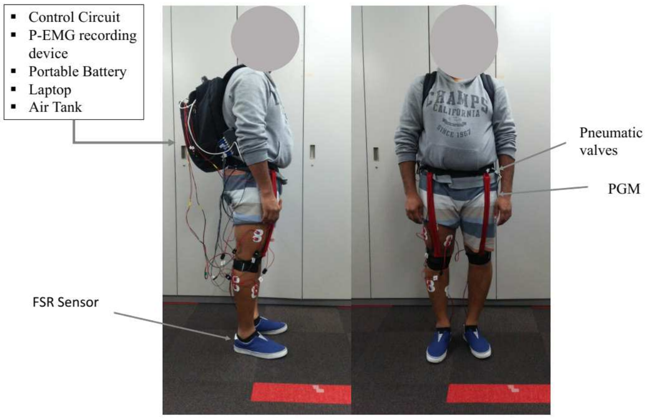

Fig. 9. Experiment setup, subject wearing AWS, electrodes and backpack

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 normal gait and variation of assisted gait.

A. Experiment Protocol

For effective evaluation of the assisted gait we need to measure minimum 3 full gait cycle are needed [10]. In our experiment we recorded sEMG for 10 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 device for recording sEMG, laptop (this laptop was remote operated to log emg data), portable battery (required for AWS controller and P-EMG device). The total weight of the backpack is 6 kg.

Total 4 experiment was performed, first experiment conducted to record sEMG data for normal gait cycle. 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 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 paired t-test method.

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

B. Results

Four experiments were conducted with five subjects to record sEMG of eight major muscles lower limb. 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 which will transition 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 and 3 levels of assisted gait with AWS. To significance of this changes in sEMG was measured 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 12 .. 19 shows averaged %MVC data for each subject for 4 experiments and their significance individual muscle. Table I 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 ($p - value < 0.05$ an $p - value < 0.01$), 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 ($p - value < 0.05$).

IV. Discussion

In this paper we discussed development of soft wearable Augment 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 limb in swing phase. By placing sensors in both shoes gait phase of the individual

		Subject 1		Subject 2		Subject 3		Subject 4		Subject 5	
Muscles	**Experiment**	**p-value**	**t-value**	**p-value**	**t-value**	**p-value**	**t-value**	**p-value**	**t-value**	**p-value**	**t-value**
TA	**Suit No Assist**	0.074	-4.219	0.002	14.861	0.062	6.025	0.013	6.062	0.052	5.073
	Suit 60 Kpa	0.074	-4.219	0.064	4.928	0.009	1.618	0.124	8.743	0.036	7.222
	Suit 100 Kpa	0.174	1.646	0.018	17.362	0.057	10.541	0.004	9.930	0.032	5.572
SOL	**Suit No Assist**	0.152	1.373	0.245	0.838	0.061	-5.137	0.025	-12.503	0.028	4.067
	Suit 60 Kpa	0.215	-1.246	0.075	2.289	0.111	2.738	0.160	-1.821	0.077	4.077
	Suit 100 Kpa	0.259	-0.944	0.182	1.552	0.072	4.341	0.072	-4.322	0.021	4.726
MG	**Suit No Assist**	0.112	2.723	0.151	1.382	0.098	3.155	0.174	1.646	0.078	4.011
	Suit 60 Kpa	0.126	-2.387	0.456	-0.124	0.017	18.882	0.008	38.250	0.336	-0.567
	Suit 100 Kpa	0.389	-0.365	0.170	-1.688	0.077	4.075	0.021	14.927	0.290	0.656
LG	**Suit No Assist**	0.067	2.455	0.041	3.264	0.489	-0.035	0.010	7.078	0.100	3.066
	Suit 60 Kpa	0.140	2.127	0.040	3.328	0.249	1.004	0.033	3.715	0.093	3.337
	Suit 100 Kpa	0.057	-2.697	0.205	1.036	0.253	0.984	0.102	1.858	0.003	13.068
RF	**Suit No Assist**	0.003	98.343	0.131	2.290	0.045	7.000	0.012	27.374	0.169	-1.698
	Suit 60 Kpa	0.003	99.310	0.041	7.739	0.002	16.878	0.004	10.742	0.061	5.115
	Suit 100 Kpa	0.110	-2.770	0.056	5.608	0.020	16.295	0.002	15.687	0.151	1.945
VM	**Suit No Assist**	0.043	7.322	0.072	4.354	0.372	-0.425	0.033	3.668	0.012	-6.300
	Suit 60 Kpa	0.001	23.555	0.002	15.475	0.141	-2.113	0.055	5.714	0.044	-7.138
	Suit 100 Kpa	0.084	-3.682	0.372	-0.427	0.221	1.201	0.098	1.914	0.073	4.312
VL	**Suit No Assist**	0.181	1.560	0.003	14.030	0.378	0.402	0.001	19.288	0.023	-13.859
	Suit 60 Kpa	0.027	4.120	0.016	20.198	0.167	1.724	0.002	17.645	0.000	-39.200
	Suit 100 Kpa	0.095	-3.243	0.473	0.083	0.195	1.423	0.038	3.401	0.006	-50.665
BF	**Suit No Assist**	0.164	-1.763	0.064	4.884	0.371	-0.427	0.058	-5.385	0.084	-3.711
	Suit 60 Kpa	0.150	-1.965	0.073	4.273	0.014	-5.804	0.030	-10.633	0.206	-1.028
	Suit 100 Kpa	0.050	6.339	0.196	1.416	0.398	-0.333	0.064	-4.924	0.146	-2.018

TABLE I

RESULT OF STATISTICAL ANALYSIS

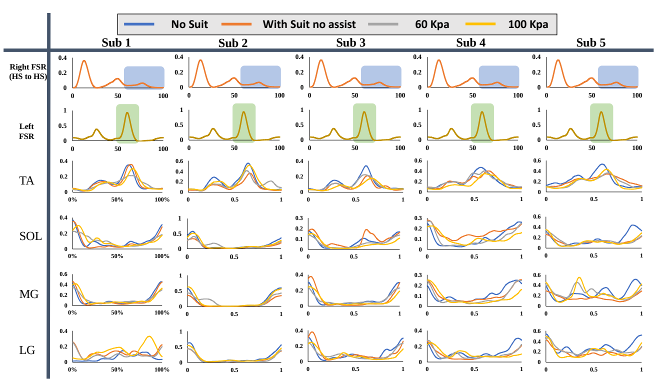

Fig. 10. sEMG of average gait cycle of posterior and anterior muscles below knee and FSR sensor data showing stance phase detection highlighted in green and supported swing phase of contralateral limb is highlighted in blue

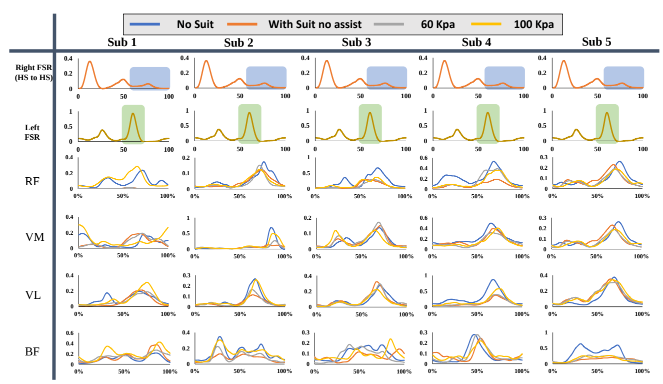

Fig. 11. sEMG of average gait cycle of posterior and anterior muscles above knee and FSR sensor data showing stance phase detection highlighted in green and supported swing phase of contralateral limb is highlighted in blue

limb and contralateral limb can be identified. Performance evaluation of AWS was done based on the stastical difference in the average %MVC of sEMG signal between normal and assisted gait measured for 5 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 rectus femoris (RF), vastus medialis (VM), vastus lateralis (VL), biceps femoris (BF), tibialis anterior (TA) and lateral gastrocnemius (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 preswing (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. Biceps femoris which is one of the hamstring muscle and responsible for hip extension and flexion support showed no significant difference in %MVC.

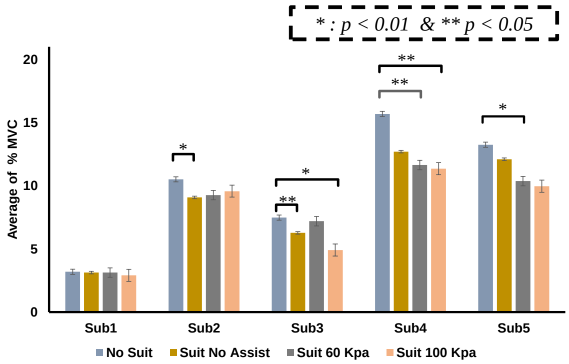

Fig. 12. Average %MVC of TA for all subjects

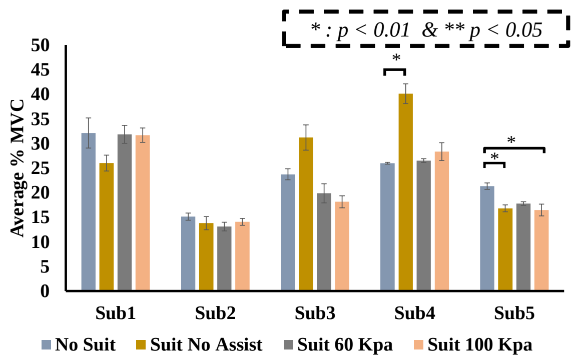

Fig. 13. Average %MVC of SOL for all subjects

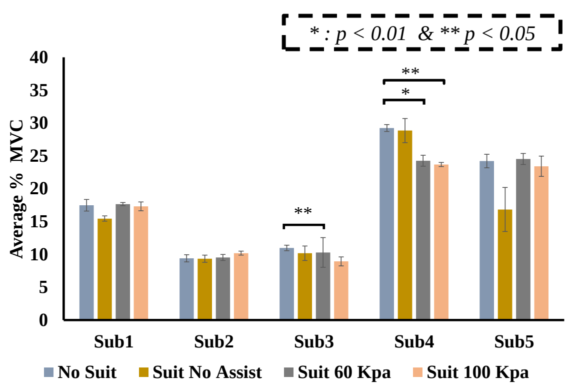

Fig. 14. Average %MVC of MG for all subjects

V. Conclusion

In this paper we developed Augment Walking Suit and PGM actuation control based on stance phase detection system. Results of performance evaluation experiment showed reduced muscle activation of lower limb muscle significantly. 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.

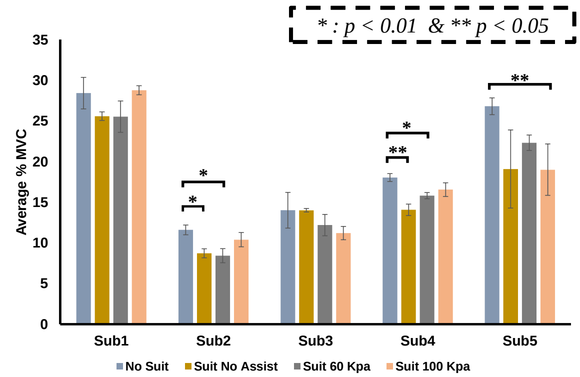

Fig. 15. Average %MVC of LG for all subjects

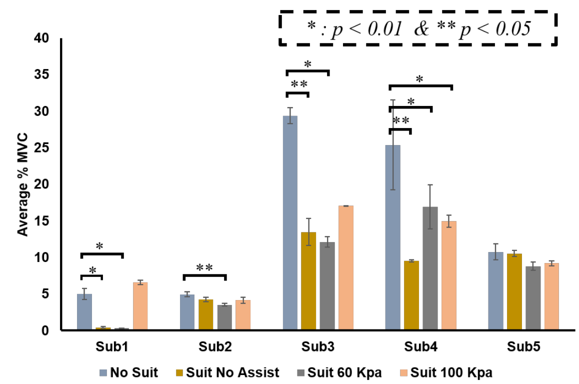

Fig. 16. Average %MVC of RF for all subjects

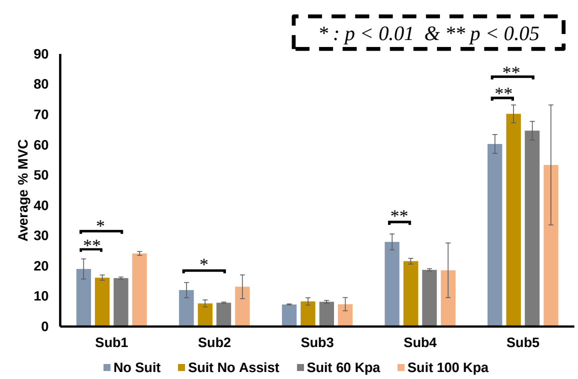

Fig. 17. Average %MVC of VM for all subjects

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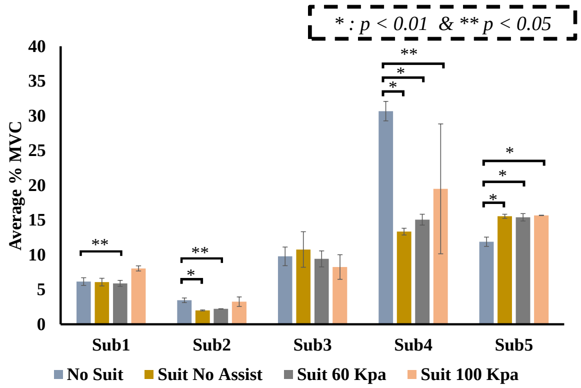

Fig. 18. Average %MVC of VL for all subjects

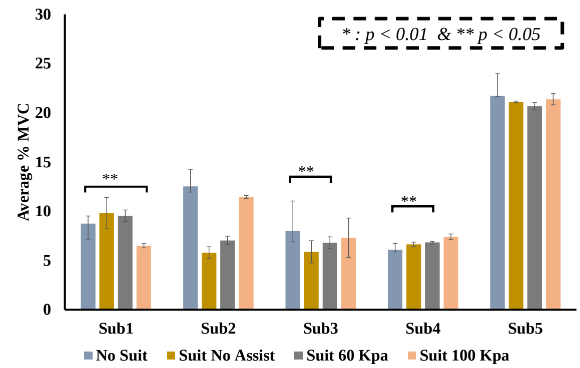

Fig. 19. Average %MVC of BF for all subjects

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