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RESEARCH ARTICLE

Investigation of EMG Parameters for Transtibial Amputees While Treadmill Walking with Different Speeds: A Preliminary Study

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This work involved human subjects or animals in its research. Approval of all ethical and experimental procedures and protocols was granted by the National Medical Research Register Secretariat 37912 and under the guidance of Certified Prosthetist and Orthotist (CPO) of the International Society of Prosthetics and Orthotics (ISPO) Category-2.

ABSTRACT Electromyography (EMG) is the process of acquiring electrical signals generated through muscle activity (contraction/relaxation). Surface EMG deliberates the amount of electrical activity in the musculoskeletal system in a non-invasive way. Under specific conditions and during certain motor activities, this signal is substantially associated with muscle strength. These Signals are used as Control Inputs by assistive devices. The study aimed to investigate the EMG parameters of lower limb muscles (rectus femoris and biceps femoris) in healthy individuals and transtibial amputees walking on a treadmill at different speeds (0.55 m/s, 0.83 m/s, and 1.11 m/s). Ten non-amputee and two amputee subjects participated. Findings reveal significant reductions in EMG signals at slower speeds, emphasizing foot stability. The right biceps femoris exhibits the highest signals average, while the right rectus femoris has the lowest for amputees. The male participants' right biceps femoris muscle showed the greatest signals of average treadmill walking activity at 0,55 m/s (0.0014 V) compared to the amputee individuals' (0.001 V). At (0,83 m/s), male participants (0.0015 V) outperformed amputee subjects (0.0004 V). At (1,11 m/s), male participants (0.0024 V) outperformed amputee subjects (0.001 V). Male participants consistently outperform amputees across speeds. The study suggests the potential application of findings in rehabilitating transtibial amputees on a treadmill, considering distance and maximum speed with a prosthesis. Overall, slow walking pace impacts EMG signals, providing insights for clinicians developing interventions for amputee rehabilitation.

INDEX TERMS Treadmill walking, transtibial amputee, electromyography, slow speed.

I. INTRODUCTION

Restoring the ability to walk is a main objective in the rehabilitation of a transtibial amputee. Prostheses are used to compensate for the limb's loss and help perform daily activities [1]. The rehabilitation tries to improve the functioning of the remaining limbs to near-normal levels. Despite massive progress being made to bring that desire to fruition, they still lack a gratifying degree of enhancement for the lives of afflicted individuals. Much research is still being conducted to reach this objective [2].

Prosthetic limbs were developed to allow individuals with limb loss to ambulate normally again [3]. Lower limb prosthetics are categorised into active and passive prostheses. Passive prostheses depend on inert components to partially iterate amputees' walking movements. In comparison, active prostheses were integrated with an actuator in manufacturing processes to boost and propel the prosthesis and add kinetic

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energy, which improved amputees' gait cycles to be as normal as possible [4].

Electromyography is a technique applied to gait analysis and a variety of other activities. EMG inside a socket offers a technique for predicting and controlling the user's movement. Earlier research used an ANN (artificial neural network) to investigate the role of EMG activity in prosthetic control [5]. The investigation revealed that combining EMG measurements with ankle joint position aids in determining the angle of the ankle throughout the motion. The result assisted in angle prediction, enabling motion control [6].

Many more articles have examined the use of EMG in prosthetic limbs. The study used the Linear Discrimination Analysis (LDA) and Bayesian Information Criterion (BIC) classification algorithms to examine and observe the kinetic pattern obtained from EMG measurements at seven different stages of the gait cycle [7]. Another study reached a satisfying outcome by establishing precise forecasts of foot motion through EMG data acquired by the thigh to operate the prosthesis using SVMs (support vector machines). According to the study, both individual variations and the number of samples exhibit a significant impact on the preciseness of gait cycle categorization, which they discovered using EMG data and SVMs [8].

While walking, the power to regulate the pace of movement is indeed an essential mechanism for controlling the operation of the locomotive to alter environmental requirements in order to improve safety [9]. There is a hypothesised correlation between low speed and dynamic unsteadiness, which has been supported by recent studies revealing that a reduction in speed could lead to an increase in the prospect of destabilising vestibulospinal drive resulting from a reduction in proprioceptive input [10]. Substantial decreases in walking speed could have an influence on the neuromuscular function of a swinging limb [11]. Moreover, at natural speed, the swing leg represents a mostly missile path and walks under passive gravity influence, dramatically diminishing walking speeds that require a very active influence mode to overcome gravity and provide adequate airtime for the swing leg [12].

The normal walking speed of healthy subjects is estimated to be 1.40 m/s (± SD of 0.20) [13]. Amputee walking speeds range from 0.78 m/s for TFA patients of traumatic origin [14] to 1.32 m/s for a group of transfemoral and transtibial amputees [15]. A TFA amputee's normal walking speed was reported to be 0.86 m/s for traumatic causes and 0.6 m/s for vascular causes [16]. Standard studies have examined slow-walking speeds varying between 0.60 m/s and 0.80 m/s [17]. Shiavi et al. [18] and Nilsson et al. [19] documented a slow-walking speed of 0.60 m/s. Murray et al. [20] measured a slow-walking speed of 0.83 m/s, while Winter et al. [21] preferred 0.75 m/s to illustrate slow-walking speed [22]. Observations from previous studies suggest that movement patterns and EMG patterns lessened in amplitude due to a reduction in walking paces [17]. Phasic EMG patterns have been less influenced by slow-walking paces. At a slow walking pace, the phasic EMG timing of ankle muscles persisted and was more compatible [22]. Several studies, however, have shown that phasic EMG patterns at slow walking speeds appear to be more changeable, particularly in the knee and hip, when compared to comfortable self-selected walking paces [21].

The lower leg muscle can adjust to alterations in walking pace by modifying the EMG magnitude and activation timing, as the modifications are considered to be muscular-specific [22]. However, a modicum of EMG pattern variation was defined when amputee subjects nominated slow-walking paces below 0.60 m/s. Even though previous studies in healthy subjects have declared that slow walking of 0.60 m/s demonstrates low amplitude in EMG signals, there is a barely noticeable perturbation in muscle activation. However, there are no statistics on EMG patterns at a slow walking pace of 0.55 m/s for transtibial amputees.

In rehabilitation, patients are mostly assessed and coached by using different walking speeds and an electric treadmill [23]. However, the study aimed to identify the influence of walking speeds (0.55 m/s, 0.83 m/s, and 1.1 m/s) on the treadmill for transtibial amputees and healthy subjects. The outcomes of this study can be used to assist in the rehabilitation treatment of a transtibial amputee, taking into account distance and maximum speed while using the prosthesis.

A. METHODS

1) PARTICIPANTS

Ten non-amputee participants and two transtibial amputees were recruited to take part in this study (Table 1). The recruited participants were free from cardiovascular and musculoskeletal circumstances that could restrict their ability to walk on a treadmill safely. Transtibial amputees have donned their accustomed prosthesis, which they have used for at least six months without further ambulatory assistance.

B. INSTRUMENTATION

Electromyography (EMG) signals were collected from two lower-limb muscles: the biceps femoris muscle (BFM) and the rectus femoris muscle (RFM). EMG signals were recorded using DELSYS EMG Wireless 4-channel sensors from the amputated and sound limbs of transtibial amputees, plus the left and right limbs of healthy subjects. All subjects placed the EMG sensors across the upper-leg muscle belly and along the direction of the muscle fibres via four silver strips. However, to determine the position and direction of the two EMG sensors, the researchers palpated two muscle surfaces while the participants conducted a sequence of spontaneous muscle activations. Regarding the transtibial amputees' upper-limb muscles (BFM and RFM), a series of possible recording points on the skin surface muscle were established by palpating the underlying bone and tissue. We also prevented vulnerable areas of skin and bone protrusions. Two EMG sensors were mounted over the monitoring points on each of the upper-limb muscles, and the transtibial

TABLE 1. Anthropometric measurements and demographic characteristics of the healthy subjects (HS) and transtibial amputees (TA-1 and TA-2).

Variable	Value	P-value
Age (years)	19 ± 10	
Weight (kg)	50 ± 80	
Height (cm)	150 ± 40	
Length of residuum (cm)	23.9 ± 9.9	
Time post-amputation (years)	20 ± 2	
Thigh circumference		
• Right thigh Limb (mm)	50 ± 30	
• Left thigh Limb (mm)	50 ± 30	
• Amputated Limb (mm)	40 ± 6	
• Sound Limb (mm)	44 ± 6	
Sex		
• Males	5 (42%)	0.0002
• Females	7 (58%)	0.00001
Amputated side		
• Right side	0	
• Left side	2 (100%)	0.147
Cause of amputation		
• Electric shock	1 (50%)	0.25
• Motorcycle accident	1 (50%)	
Social situation		
• Lives with a partner	1 (8%)	0.169
• Lives with family	2 (16)%	0.035
• Lives alone	4 (34%)	0.001
• Institutionalized	5 (42%)	0.0002
Comorbidity		
• Mental disease	1 (8%)	0.169
• Stressfulness	2 (16)%	0.035
• Fatigue	1 (8%)	
• Others	2 (16)%	
Walking aids		
• Independent	2 (100%)	0.147
• 1 cane	0	
• 2 canes	0	

(a)

(b)

FIGURE 1. Transtibial amputee subjects with DELSYS Wireless EMG Sensors. Placement of EMG sensors for upper-limb muscles: biceps femoris muscle (BFM) and rectus femoris muscle (RFM). Fig 1(a) Amputee TA-1 (a 29-year-old whose amputation occurred due to a 33-kV electrical shock at age 21) has a comparatively short upper leg and a comparatively large muscle volume. By contrast, Fig 1(b) the amputee TA-2 (a 26-year-old with an amputation due to a motorcycle accident at age 20) has a larger upper leg and shorter muscle volume.

amputees donned their prostheses and walked all over the laboratory to evaluate comfort. No modifications have been made to the amputee prosthesis (Figure 1).

EMG signals were recorded in RFM and BFM by using a treadmill for walking exercise and gait analysis. The treadmill was used as a practical-oriented physical device to gather data and record results. Also, it is useful for rebuilding muscle function following injury recovery. The treadmill is quite involved in clinical research and sports activities [24]. The experiment to capture EMG signals for treadmill walking activity lasted 10 seconds. Each non-amputee participant had three successful trials, and each amputee subject had five successful trials.

C. PROTOCOL

Before the test, the muscles addressed were above the knee: (RF) the rectus femoris muscle in an anterior view and (BF) the biceps femoris muscle in the posterior view. All participants, comprising non-amputee and amputee individuals, were instructed to walk on a treadmill at three different speeds. The positions of the EMG electrodes and the subjects are depicted in Figures 1(a) and (b). Participants completed voluntary warm-up trials in which they attempted to activate the muscles above the knee: the biceps femoris and rectus femoris. For one minute, the subjects walked on the treadmill at a random speed determined by them.

During the test, the participants were asked to rest for five minutes, then walk on the treadmill for one minute, and after that, rest for five minutes, repeating the process for the three different speeds. The participants were asked to walk on a treadmill, starting at 0.55 m/s, following that at 0.83 m/s, and then at 1.11 m/s, respectively. All participants were successfully able to walk at three speeds. The subjects completed the entire walk on the treadmill within two hours. However, five successful trials were conducted and recorded for both amputee and non-amputee subjects, we calculated their average values and used them for the statistical data.

II. RESULTS

The ten non-amputee volunteer subjects had an average age of 19 ± 10 (SD = 3.6) years: female (7/10, 70%), male (3/10, 30%), undergraduate and postgraduate students studying at the University of Malaya, and two transtibial amputee subjects had an average age of 26 ± 3 (SD = 2.1) years: male (2/2, 100%), self-independent worker. Table 1 displays the anthropometric measurements and demographic characteristics of the non-amputee and amputee subjects.

A. FIRST-SPEED (0.55 M/S) WHILE WALKING ON THE TREADMILL

1) RIGHT LIMB

By analysing the right limb at the 0.55 m/s speed, (a) Right Rectus Femoris (AVG), the male average had the most significant mean difference between the average of the subjects (M_{diff} = 0.00045) and a positive correlation (r = 0.245, p = 0.001), and (b) Right Biceps Femoris (AVG), the amputee TA-2 average had the biggest mean difference between the average of the subjects (M_{diff} = 0.000629) and a positive correlation (r = 0.09, p = 0.249). (Table 2, Figure 1).

Right Rectus Femoris T-test (TA-1 = 20.1, TA-2 = 24, Male = 15.9, Female = 21.3), and Right Biceps Femoris T-test (TA-1 = 25.5, TA-2 = 25, Male = 18.6, Female = 43.6). The one-sample t-test revealed a significant difference between the TA-1, TA-2, Male and Female subjects based on the identical observations in the collected sample ($p < 0.001$). The RBF paired samples t-test, TA-1 and TA-2 (t = −20.72, df = 165, $p < 0.001$, M_{diff} = −28.7306003624, ES = −1.6082), Male and Female (t = −0.473, df = 165, p = 0.637, M_{diff} = −9.4036165621, ES = −0.0367).

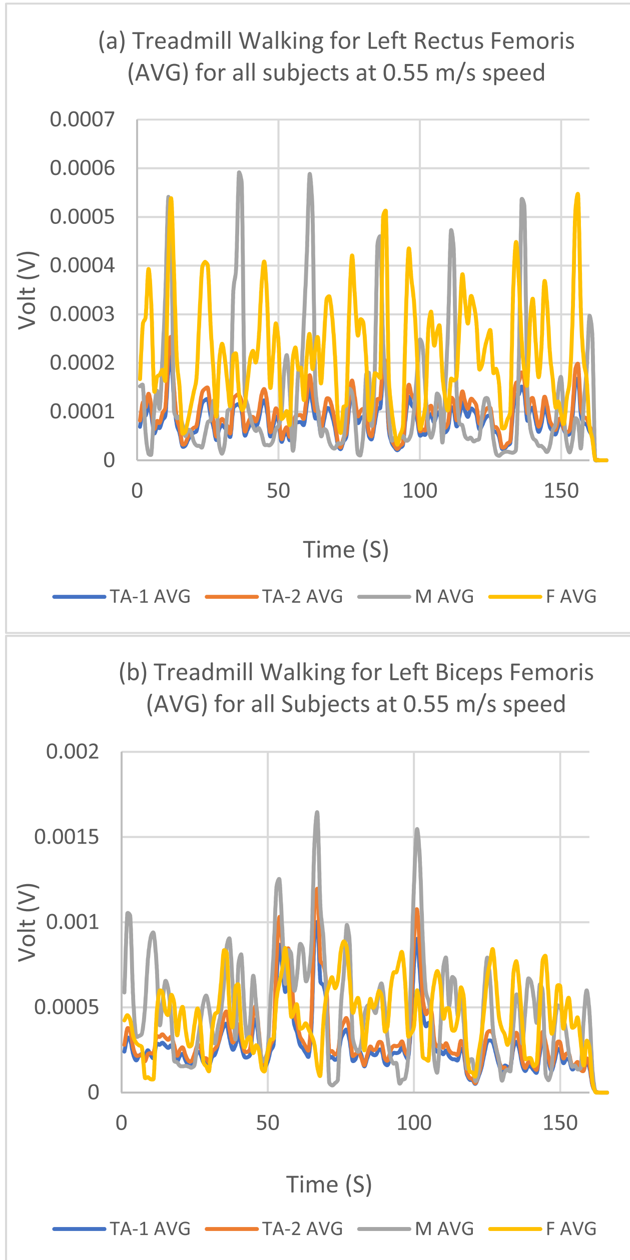
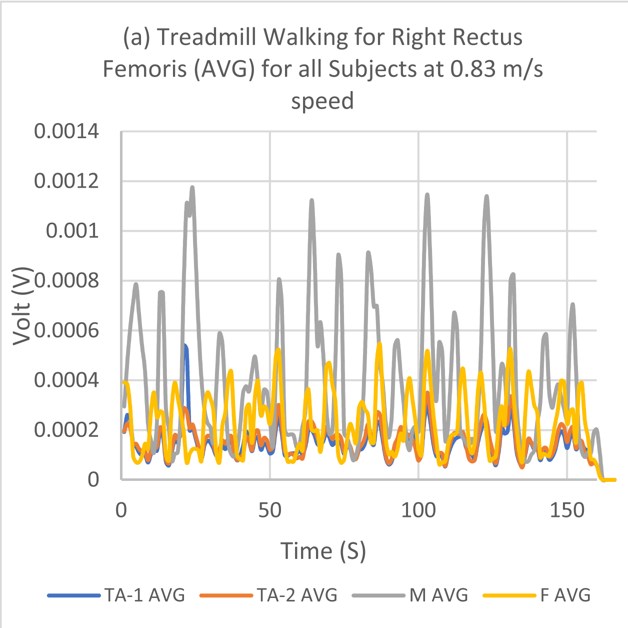


FIGURE 2. (a) The AVG of walking activity on the treadmill at 0.55 m/s speed for right rectus femoris muscles from Male subjects, Female subjects, transtibial amputee 1, and transtibial amputee 2. (b) The AVG of walking activity on the treadmill at 0.55 m/s speed for right biceps femoris muscles from Male subjects, Female subjects, transtibial amputee 1, and transtibial amputee 2.

2) LEFT LIMB

By analysing the left limb at the 0.55 m/s speed, (a) Left Rectus Femoris (AVG), the female average had the most significant mean difference between the average of the subjects (M_{diff} = 0.000213) and a negative correlation (r = −0.019, p = 0.804), and (b) Left Biceps Femoris (AVG), the male average had the biggest mean difference between the average of the subjects (M_{diff} = 0.000481) and a positive correlation (r = 0.08, p = 0.304). (Table 2, Figure 2).



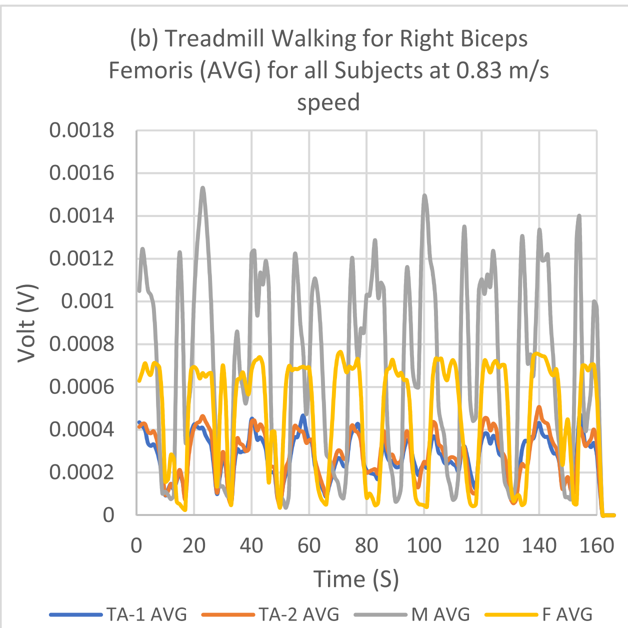


FIGURE 3. (a) The AVG of walking activity on the treadmill at 0.83 m/s speed for right rectus femoris muscles from Male subjects, Female subjects, transtibial amputee 1, and transtibial amputee 2. (b) The AVG of walking activity on the treadmill at 0.83 m/s speed for right biceps femoris muscles from Male subjects, Female subjects, transtibial amputee 1, and transtibial amputee 2.

Left Rectus Femoris T-test (TA-1 = 28.3, TA-2 = 27.9, Male = 11.7, Female = 23.6), and Left Biceps Femoris T-test (TA-1 = 20.2, TA-2 = 19.3, Male = 19, Female = 26.3). The one-sample t-test revealed a significant difference between the TA-1, TA-2, Male and Female subjects based on the identical observations in the collected sample ($p < 0.001$). The RRF paired samples t-test, TA-1 and TA-2 ($t = -25.47$, $df = 165$, $p < 0.001$, $M_{diff} =$ -28.7306003624, $ES = -1.977$), Male and Female ($t = -6.55$, $df = 165$, $p = 0.637$, $M_{diff} = -29.7907302755$, $ES = -0.509$).

B. SECOND-SPEED (0.83 M/S) WHILE WALKING ON THE TREADMILL

1) RIGHT LIMB

By analysing the right limb at the 0.83 m/s speed, (a) Right Rectus Femoris (AVG), the male average had the most significant mean difference between the average of the subjects ($M_{diff} = 0.000388$) and a positive correlation ($r = 0.05$, $p = 0.526$), and (b) Right Biceps Femoris (AVG), the male average had the biggest mean difference between the average of the subjects ($M_{diff} = 0.000677$) and a positive correlation ($r = 0.128$, $p = 0.099$). (Table 2, Figure 3).

Right Rectus Femoris T-test (TA-1 = 26.3, TA-2 = 30, Male = 17.9, Female = 22), and Right Biceps Femoris T-test (TA-1 = 31.3, TA-2 = 29.9, Male = 19.7, Female = 22.4). The one-sample t-test revealed a significant difference between the TA-1, TA-2, Male and Female subjects based on the identical observations in the collected sample ($p < 0.001$). The RRF paired samples t-test, TA-1 and TA-2 ($t = -3.6$, $df = 165$, $p < 0.001$, $M_{diff} = -31.3343866412$, $ES = -0.279$), Male and Female ($t = 7.02$, $df = 165$, $p < 0.001$, $M_{diff} = -0.000165$, $ES = 0.545$).

2) LEFT LIMB

By analysing the left limb at the 0.83 m/s speed, (a) Left Rectus Femoris (AVG), the male average had the most significant mean difference between the average of the subjects ($M_{diff} = 0.000384$) and a positive correlation ($r = 0.153$, $p = 0.049$), and (b) Left Biceps Femoris (AVG), the male average had the biggest mean difference between the average of the subjects ($M_{diff} = 0.00037$) and a positive correlation ($r = 0.052$, $p = 0.508$). (Table 2, Figure 4).

Left Rectus Femoris T-test (TA-1 = 30.9, TA-2 = 29.4, Male = 18.2, Female = 24.3), and Left Biceps Femoris T-test (TA-1 = 30.9, TA-2 = 30.1, Male = 18.7, Female = 23.4). The one-sample t-test revealed a significant difference between the TA-1, TA-2, Male and Female subjects based on the identical observations in the collected sample ($p < 0.001$). The RBF paired samples t-test, TA-1 and TA-2 ($t = -25.47$, $df = 165$, $p < 0.001$, $M_{diff} = -11.9859842991$, $ES = -2.0018$), Male and Female ($t = 0.546$, $df = 165$, $p = 0.586$, $M_{diff} = 0.0000133$, $ES = 0.0424$).

C. THIRD-SPEED (1.11 M/S) WHILE WALKING ON THE TREADMILL

1) RIGHT LIMB

By analysing the right limb at the 1.11 m/s speed, (a) Right Rectus Femoris (AVG), the male average had the most significant mean difference between the average of the subjects ($M_{diff} = 0.000532$) and a positive correlation ($r = 0.182$, $p = 0.019$), and (b) Right Biceps Femoris (AVG), the male average had the biggest mean difference between the average

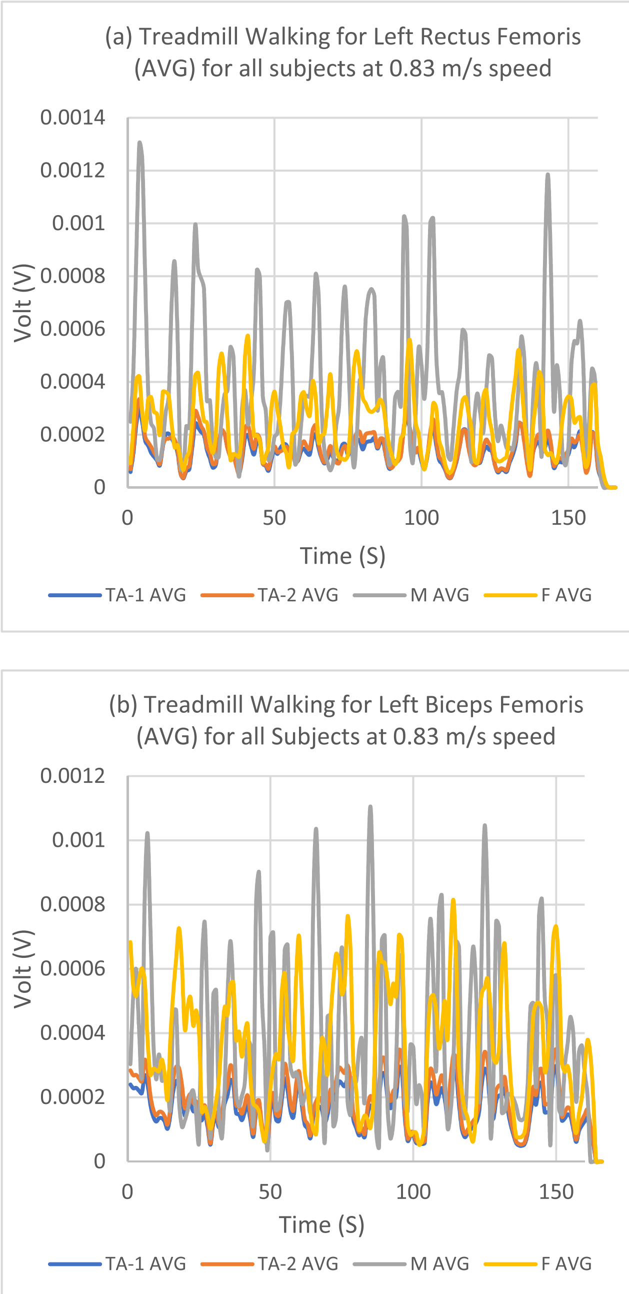


FIGURE 4. (a) The AVG of walking activity on the treadmill at 0.83 m/s speed for left rectus femoris muscles from Male subjects, Female subjects, transtibial amputee 1, and transtibial amputee 2. (b) The AVG of walking activity on the treadmill at 0.83 m/s speed for left biceps femoris muscles from Male subjects, Female subjects, transtibial amputee 1, and transtibial amputee 2.

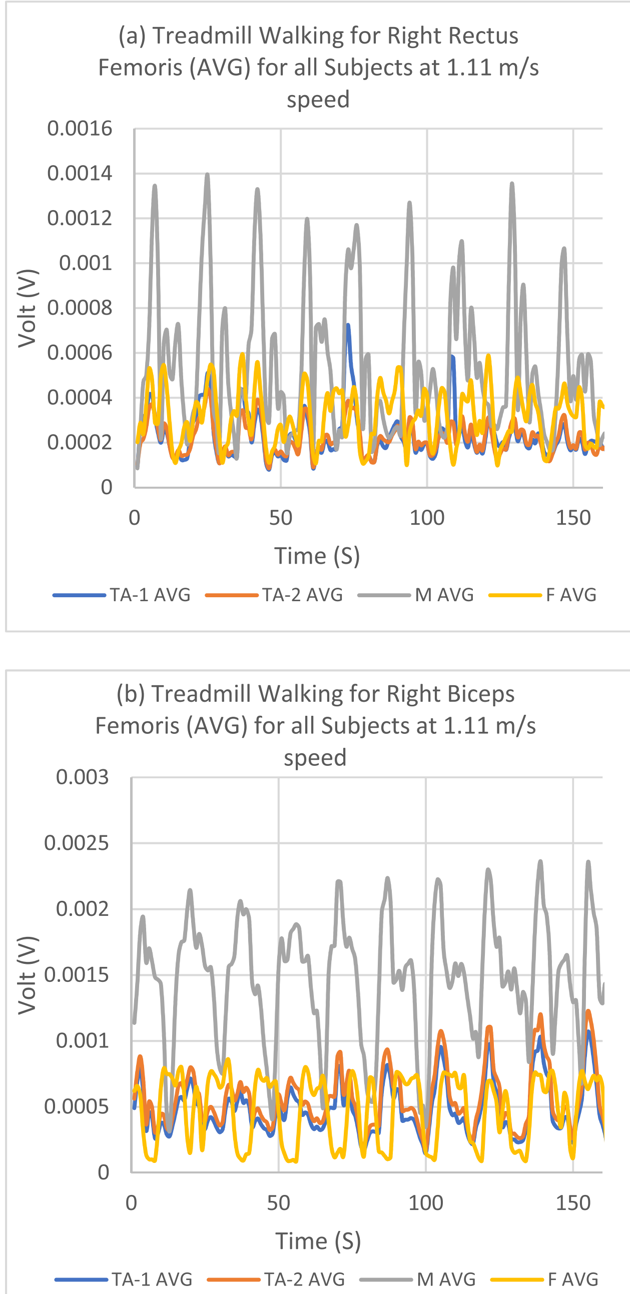


FIGURE 5. (a) The AVG of walking activity on the treadmill at 1.11 m/s speed for right rectus femoris muscles from Male subjects, Female subjects, transtibial amputee 1, and transtibial amputee 2. (b) The AVG of walking activity on the treadmill at 1.11 m/s speed for right biceps femoris muscles from Male subjects, Female subjects, transtibial amputee 1, and transtibial amputee 2.

of the subjects ($M_{diff} = 0.00143$) and a negative correlation ($r = -0.045$, $p = 0.564$) (Figure 5).

Right Rectus Femoris T-test (TA-1 = 26.3, TA-2 = 35.5, Male = 21.1, Female = 31.4), and Right Biceps Femoris T-test (TA-1 = 28.9, TA-2 = 29.4, Male = 36.3, Female = 23.3). The one-sample t-test revealed a significant difference between the TA-1, TA-2, Male and Female subjects based on the identical observations in the collected sample ($p < 0.001$). The RBF paired samples t-test, TA-1 and TA-2 ($t = -30.6$, $df = 165$, $p < 0.001$, $M_{diff} = -26.39287799$, $ES = -2.37$), Male and Female ($t = 21.3$, $df = 165$, $p < 0.001$, $M_{diff} = 0.000962$, $ES = 1.66$).

2) LEFT LIMB

By analysing the left limb at the 1.11 m/s speed, (a) Left Rectus Femoris (AVG), the male average had the most significant mean difference between the average of the subjects ($M_{diff} = 0.000598$) and a positive correlation ($r = 0.106$, $p = 0.175$),

TABLE 2. The mean value and standard deviation of transtibial amputees (TA-1 and TA-2) and healthy subjects (HS) while walking on a treadmill at three different speeds are calculated in percentage.

Mean value and standard deviations of BFM and RFM (0.55 m/s)						
	HS		TA-1		TA-2	
	AVG	SD	AVG	SD	AVG	SD
Left BF	44.4	0.04	0.00304	0.0006	0.00288	0.000552
Left RF	18.5	0.02	0.000903	0.000237	0.00138	0.000476
Right BF	61.3	0.04	0.00323	0.000807	0.00637	0.00293
Right RF	28.9	95	0.000827	0.000336	0.010563	0.0055
TOTAL	153.1	95.1	0.0079923	0.0019808	0.021196	0.009453
MEAN	38.28	23.78	0.0019981	0.0004952	0.005299	0.002363
SD	18.68	47	0.00	0.00	0.00	0.00

Mean value and standard deviations of BFM and RFM (0.83 m/s)						
	HS		TA-1		TA-2	
	AVG	SD	AVG	SD	AVG	SD
Left BF	36.1	0.038	0.00315	0.000572	0.00329	0.000718
Left RF	28.1	0.032	0.000996	0.000286	0.00171	0.000554
Right BF	52.7	0.065	0.012206	0.00909	0.0086	0.00482
Right RF	27.3	0.035	0.000663	0.000129	0.02543	0.022185
TOTAL	144.2	0.17	0.017016	0.010073	0.039029	0.02828
MEAN	36.05	0.043	0.0042541	0.0025183	0.009757	0.0070701
SD	11.79	0	0.00	0.00	0.00	0.00

Mean value and standard deviations of BFM and RFM (1.11 m/s)						
	HS		TA-1		TA-2	
	AVG	SD	AVG	SD	AVG	SD
Left BF	67.5	0.044	0.015206	0.010959	0.00424	0.00221
Left RF	44.8	0.042	0.00117	0.000755	0.0043	0.00382
Right BF	75.9	0.06	0.063855	0.074846	0.00866	0.00806
Right RF	37.7	0.036	0.000733	0.000282	0.085097	0.11911
TOTAL	225.9	0.182	0.080969	0.086841	0.102292	0.133202
MEAN	56.48	0.046	0.020242	0.02171	0.025573	0.0333
SD	18.14	0	0.00	0.00	0.00	0.00

and (b) Left Biceps Femoris (AVG), the female average had the biggest mean difference between the average of the subjects ($M_{diff} = 0.000747$) and a positive correlation ($r = 0.037$, $p = 0.639$). (Table 2, Figure 6).

Left Rectus Femoris T-test (TA-1 = 37.4, TA-2 = 37.4, Male = 23.4, Female = 29.9), and Left Biceps Femoris T-test (TA-1 = 44.1, TA-2 = 43.6, Male = 20.7, Female = 38.4). The one-sample t-test revealed a significant difference between the TA-1, TA-2, Male and Female subjects based on the identical observations in the collected sample ($p < 0.001$). The RBF paired samples t-test, TA-1 and TA-2 ($t = -40.91$, $df = 165$, $p < 0.001$, $M_{diff} = -20.3311095125$, $ES = -3.175$), Male and Female ($t = -7.81$, $df = 165$, $p < 0.001$, $M_{diff} = -10.5238763883$, $ES = -0.606$).

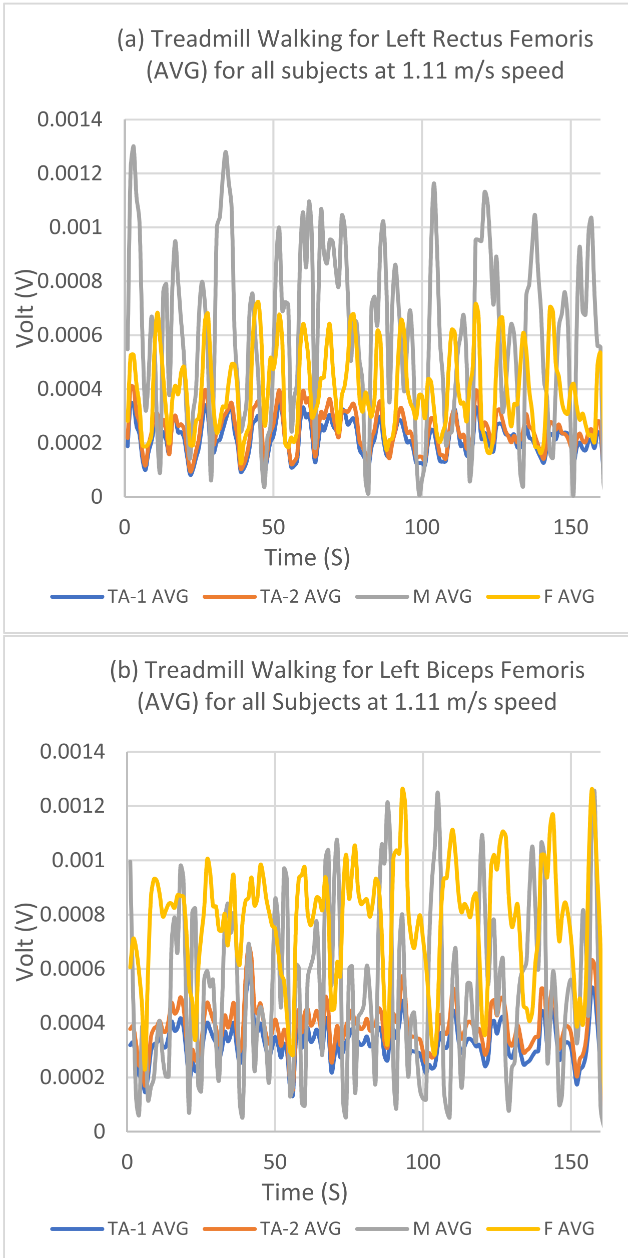


FIGURE 6. (a) The AVG of walking activity on the treadmill at 1.11 m/s speed for left rectus femoris muscles from Male subjects, Female subjects, transtibial amputee 1, and transtibial amputee 2. (b) The AVG of walking activity on the treadmill at 1.11 m/s speed for left biceps femoris muscles from Male subjects, Female subjects, transtibial amputee 1, and transtibial amputee 2.

III. DISCUSSION

The purpose of this study is to investigate the EMG parameters of lower limb muscles (rectus femoris and biceps femoris) muscles for non-amputee and amputee participants while walking on the treadmill at three different speeds. Also, to compare the average values of the right and left (rectus and biceps) femoris muscles between the healthy subjects and amputees. In light of the importance of the rectus femoris muscle, it is in charge of managing the swing phase of gait. To avoid abnormal knee flexion and poor foot posture towards the conclusion of the swing phase, its functions must be continuously watched during the late stance-early swing phase. In addition, biceps femoris activity exhibits the strongest relationship between gait speed and swing length asymmetry index. So, greater biceps femoris action results in greater knee flexion angle and foot clearance [25].

The outcomes of this study have generally demonstrated that the EMG signals of the lower limb muscles appear to be boosted by the different speeds of movement. Variations in the activity pattern have been observed in BFM and RFM, which can be distinctively dependent on particular speed ambits. In transtibial amputees' RF muscles, the function that controls the speed of muscular activity adaptation tends to be fairly consistent at slow speeds. Both transtibial amputees had weak residual limb muscle activation patterns that were closely similar to each other and varied dramatically from healthy subjects.

Table 2 illustrated the mean difference in EMG signals between transtibial amputees and healthy subjects walking on the treadmill at three different speeds (0.55 m/s, 0.83 m/s, and 1.11 m/s), whereas Huang et al. [26] reported the measurement of four speeds of treadmill walking (0.7, 1.0, 1.3, and 1.6 m/s). In this study, we recorded the muscle signals from the amputated and sound limbs of amputee subjects and the right and left limbs of healthy subjects. In another study, EMG signals were recorded from the amputated leg of amputee subjects and the right leg of control subjects. As in earlier studies, we discovered that walking speed had a major influence on muscle activity, with large and systematic increases in EMG peak amplitude as walking speed rose [9], [10], [11], [12], [13]. Lack of activity is a primary element that has a detrimental impact on the muscles, causing them to atrophy. However, their everyday activities have a significant influence on the efficiency of those muscles.

In a previous study [24], data were obtained as participants walked at seven different speeds. Thirty amputees were involved, and 10 trials were held. In our investigation, there were five trials for amputees and non-amputee individuals, with one trial chosen as the strongest signal. The greatest peaks of AVG measures in each individual are shown in Table 2. Thus, based on the electrical current recorded from the muscles while doing treadmill walking activities, these data demonstrated the effectiveness and functioning of the muscles. Another study by Otter et al. [9] found that it is vital to study walking at normal to extremely slow rates in order to overcome the challenges that amputees may face. Little research has been done on treadmill walking at slow speeds [24]. Slow speeds have been reported at 0.06, 0.11, 0.17, 0.22, 0.28, 0.83, and 1.39 m/s. This research included nine healthy people. Based on this study, no amputee participants were engaged [9].

According to the outcomes, at 0.55 m/s speed, the right RFM muscle had the highest AVG signals in male subjects (0.0018 V), while the lowest AVG signals were in TA-2 amputee (0.0004 V). The right BF muscle had the highest AVG signals in male subjects (0.0014 V), while the lowest AVG signals were in TA-1 amputee (0.001 V). The left RF muscle had the highest AVG signals in male subjects (0.0006 V), while the lowest AVG signals were in TA-1 amputee (0.0002 V). The left BF muscle had the highest AVG signals in male subjects (0.0017 V), while the lowest AVG signals were in female subjects (0.0008 V). At 0.83 m/s speed, the right RF muscle had the highest AVG signals in male subjects (0.0012 V), while the lowest AVG signals were in TA-2 amputee (0.0003 V). The right BF muscle had the highest AVG signals in male subjects (0.0015 V), while the lowest AVG signals were in TA-1 and TA-2 amputees (0.0005 V). The left RF muscle had the highest AVG signals in male subjects (0.0013 V), while the lowest AVG signals were in TA-1 and TA-2 amputees (0.0003 V). The left BF muscle had the highest AVG signals in male subjects (0.0011 V), while the lowest AVG signals were in TA-1 and TA-2 amputees (0.0003 V). At 1.11 m/s speed, the right RF muscle had the highest AVG signals in male subjects (0.0014 V), while the lowest AVG signals were in TA-2 amputee (0.0004 V). The right BF muscle had the highest AVG signals in male subjects (0.0024 V), while the lowest AVG signals were in female subjects (0.0008 V). The left RF muscle had the highest AVG signals in male subjects (0.0013 V), while the lowest AVG signals were in TA-1 amputee (0.0003 V). The left BF muscle had the highest AVG signals in male and female subjects (0.0012 V), while the lowest AVG signals were in TA-1 amputee (0.0005 V).

Amputees had a variation in BF and RF muscles at all three different speeds. It is rational to conclude that the amplitude could be reduced if residual limb muscle activity had a practical objective while walking on a treadmill, for instance, to regulate the stability of the controlled prosthesis. In the previous study, Huang and Ferris [26] reported that the loss of proprioceptive or visual stimulation of muscle movement is one reason why amputee subjects may have lost effective volitional regulation of the residual limb muscles. Also, the absent ankle joint fails to produce sensory input at the joint site, resulting in no references to explain the effects of muscle activity [27]. In addition, the attachment of the residual limb muscle to the prosthetic limb is likely to boost voluntary control [28]. Past studies have shown that amputee subjects achieved higher residual limb biceps femoris muscle activation by early posture comparison to healthy subjects, to reinforce the joint of the knee and increase the acceleration of the residual limb [10] [29], [30], [31]. While in this study, TA-2 amputee achieved higher residual limb biceps femoris muscle activation in comparison to female subjects and TA-1 amputee.

A majority of amputee individuals displayed residual lower-limb muscle activation patterns that had been attached to the gait cycle yet were extremely varied between individuals when walking. The patterns of residual lower-limb muscle activation were considerably different from the patterns of healthy subjects [26]. Amputees demonstrated higher inter-subject variability in biceps femoris muscle activation while walking than healthy subjects. In another study, amputees had more gluteus medius muscle activity than healthy participants. Furthermore, amputees showed a distinct biceps femoris activation profile shape compared to healthy individuals [32]. Earlier research has revealed that amputees walk using higher residual limb biceps femoris activation throughout early stance than intact biceps femoris to stabilise the knee joint or boost residual limb propulsion. During walking, the inter-subject variation in biceps femoris activation contour found in amputee subjects implies that transtibial amputees adopt different compensatory muscle recruitment patterns [33]. In other studies, amputees and healthy individuals had low EMG cross-correlation values for the gastrocnemii and tibialis anterior. Notwithstanding the substantial variation in residual lower-limb EMG patterns among amputees, inter-stride variation was comparable to that of healthy subjects. The only muscle exhibiting a substantially higher variance-to-signal ratio in the amputee group than in the healthy group was the gastrocnemius medial head [34]. Plenty of amputees demonstrated strong volitional control over residual lower-limb muscle activity. Many amputees' residual muscle activation patterns were identical to those of healthy subjects during maximal voluntary dorsiflexion and plantar flexion [26]. In previous studies, a number of amputees could distinguish between gastrocnemii and tibialis anterior activity and exhibited coactivation levels comparable to healthy individuals. Throughout volitional maximal activation, other amputees were incapable of distinguishing between gastrocnemii and tibialis anterior activity. The absence of proprioceptive and visual indications of muscle activations possibly contributed to the amputees' loss of strong volitional control of the residual limb muscles [35].

Automatically and regularly, a treadmill belt draws back the standing leg, allowing impulses to flow continuously from the tendon Golgi apparatus, muscle spindle, and cutaneous afferents to the motor pattern generator (PG). Varied environmental data, including vestibular information and visual information, is sent to the spinal PG at the same moment; therefore, various walking patterns regarding the walking cycle and time of foot contact with the surface are anticipated. Individuals' walking patterns modify in response to alterations in the pain, walking surface, and variations in velocity. A higher walking pace increases muscular effort and energy consumption [36]. Furthermore, when walking on a treadmill, the number of cadences and steps is raised, as are the stance phase and step length, causing subjects to employ additional muscles to preserve stability. Walking at fast speeds on a treadmill promotes muscular activation and strengthens muscles. Nevertheless, walking on a treadmill is additionally efficient, regardless of velocity, owing to the higher amount of effort performed by boosting calorie consumption and expediting weight reduction in obese individuals [37].

The residual limb biceps femoris and vastus lateralis muscles showed considerable differences compared to the sound limb throughout all speeds. When comparing the residual limb to the sound limb, the biceps femoris displayed increased activity throughout braking at all speeds, which served to boost the positive hip performance of these individuals in early stances [10]. This observation is compatible with prior amputee research that has found greater residual limb hip power compared to non-amputees throughout the first half of the stance [38]. Earlier simulation and modelling research on healthy subjects walking have revealed the fact that the bicep femoris has the ability to make a positive contribution to propulsion during the stance. As a result, enhanced residual limb biceps femoris activity that occurs in the early stance is expected to provide a greater positive contribution to the anterior/posterior (A/P) ground reaction force (GRF), lowering the net braking GRF in analogised to the sound limb and healthy control limb during the early stance. Higher residual limb biceps femoris activity throughout braking was already seen in amputees walking at average speeds. Biceps femoris compensation happens over many different kinds of speeds [39].

Additionally, at all speeds, the residual limb Vastus Lateralis muscle activity has risen analogised to a sound limb. At moderate speeds, residual limb Vastus Lateralis muscle activity has formerly been demonstrated from early to mid-stance. Without ankle plantar flexors, the Vastus Lateralis muscle adapts over many different kinds of speeds and seems to be required to offer extra body support [40]. At higher speeds, along with risen biceps femoris and vastus lateralis muscle activity, there was also increased activity in the residual limb rectus femoris muscle when braking. The rectus femoris was previously proven in healthy subjects walking to give body support when activated while braking. Therefore, the Rectus Femoris seems to collaborate alongside the Vastus Lateralis muscle to give extra body support [41].

In normal movement, the main objective of the biceps femoris muscles has the power to develop assistance from an early stance to mid-stance. Ankle dorsi-flexors produce assistance in an early stance, and ankle plantar flexors produce assistance in a late stance [42]. Transtibial amputees are likely to offset the loss of assistance from the ankle muscles by trying to recruit muscles above the knee to enhance walking steadiness through posture [43]. The interindividual variation in biceps femoris muscle activation patterns observed in amputees suggests that there were differences in the patterns of compensated muscular mobilisation used by amputee subjects while walking [9].

Getting back or sustaining walking capacity can be difficult for people with lower limb amputations. If a particular walking pace surpasses a reasonable limit, individuals with amputation choose to slow down, lowering metabolic energy consumption. When compared to healthy subjects, those with traumatic amputations lowered their walking speed by 21.6%. Individuals who had a vascular amputation walked much more (45.2%) than individuals who had a trauma amputation. The two sets of amputees chose slower walking speeds than healthy subjects [44]. In light of the absence of the COP modulation mechanism, which tends to be provided by ankle motor functions, individuals with amputation are obliged to pick the stepping approach instead of the in-stance strategy whenever confronted with an outward-directed disturbance on the amputated side [45]. When outward-directed distractions happened at the time of the non-amputated limb's foot strike, subjects in an amputation regulated the centre of pressure and ground reaction force precisely compared to healthy people, showing the usage of in-stance balancing procedures [46]. However, because disturbances happened as the amputated limb approached the stance phase, the individual displayed no in-stance reaction. Amputation patients used the stepping technique and adjusted the posture of the non-amputated limb in the succeeding stance phase to form a cross-step. A response of this size leads to a significantly larger shift in the centre of mass [44], [47].

Gait training on a treadmill increased bioenergetic performance. Lower-limb amputees, on the other hand, demonstrate a less efficient walking pattern, which becomes apparent with bilateral involvement or greater amputation levels. Therefore, when amputee individuals attempt to reduce energy expenditure while walking, such a concern could lead to extra gait consequences, which might involve reduced self-selected speed of walking and reliance on a walking aid, along with gait perversion, demonstrating the importance of boosting bioenergetic effectiveness for amputees [48].

EMG signals at three different speeds during walking on a treadmill are consistent with the literature [13], [24], [49], [50], [51], with some differences between healthy subjects and transtibial amputees in this study. Previous studies have stated that the treadmill gait usually raises the amplitudes of the EMG among the entire lower limb muscles [42], [50], [52], [53]. This study revealed increased EMG amplitudes in the right biceps femoris muscle (BFM) at three different speeds for healthy subjects and amputees while walking on the treadmill. Also, the inter-individual differences in the EMG parameters were large for healthy subjects at slow speeds. Moreover, the fatty tissue layer affects the EMG signal. Yet, the effect of the fatty tissue layer on the EMG signal varies depending on the muscle under consideration [54].

IV. CONCLUSION

This study focused on investigating the EMG parameters for amputee and non-amputee subjects walking on the treadmill at three different speeds and comparing the EMG parameters between amputees and healthy subjects. In this study, EMG signals were recorded from two lower limb muscles: the rectus femoris and the biceps femoris. This study demonstrates that EMG signals were strongly affected by treadmill walking at three different speeds. EMG signals for left RF, right RF, left BF, and right BF were different across (0.55, 0.83, and 1.11 m/s) speeds. EMG signals decreased at slow speeds, likely to assist with foot stability, and occurred due to the mechanical requirements of walking at different speeds. EMG signals in male subjects were higher than in female subjects and transtibial amputees, with a range of 36% to 56%. Assessment of the impacts of a slow walking pace can enable clinicians to develop interventions that target transtibial amputees to rehabilitate on treadmill walking, taking into account distance and maximum speed while using the prosthesis.

DISCLAIMER

The views expressed in the submitted article are of the authors and not an official position of the institutions.

ETHICS APPROVAL

The research was conducted with the approval of the National Medical Research Register Secretariat 37912 and under the guidance of Certified Prosthetist and Orthotist (CPO) of the International Society of Prosthetics and Orthotics (ISPO) Category-2.

DECLARATION OF CONFLICTING INTERESTS

The author(s) declared no potential conflicts of interest with respect to the research, authorship, and/or publication of this article.

ABBREVIATIONS

EMG = Electromyography.
AVG = Average.
HS = Healthy Subject.
TA-1 = Transtibial amputee subject-1.
TA-2 = Transtibial amputee subject-2.
RF = Rectus femoris.
BF = Biceps femoris.
RFM = Rectus femoris muscle.
BFM = Biceps femoris muscle.
TFA = Transfemoral Amputee.
TTA = Transtibial Amputee.
SD = Standard Deviation.
GRF = Ground Reaction Force.
A/P = Anterior/Posterior.

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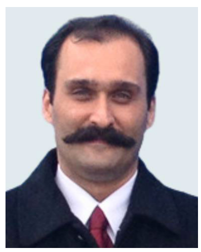

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