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original article

Ground reaction force and electromyographic activity of transfemoral amputee gait: a case series

Força de Reação do Solo e atividade eletromiográfica da marcha de amputados transfemorais: uma série de casos

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Abstract – Ground reaction forces (GRF) and electromyographic activity form a part of the descriptive data that characterise the biomechanics of gait. The research of these parameters is important in establishing gait training and understanding the impact of amputation and prosthetic components on movement during the act of walking. Therefore, this case series describes the GRF and electromyographic activity in the gait of transfemoral amputees. A force plate was used to measure GRF, and an electromyographic system monitored the vastus lateralis, biceps femoris, tibialis anterior, and gastrocnemius lateralis muscles of the non-amputated leg. The average vertical and anteroposterior GRF time-curves, average electromyographic activity, and descriptor variables were then analysed. We observed decreases in vertical and anteroposterior GRF magnitudes as well as in anteroposterior GRF descriptor variables during the propulsive phase in the amputated leg. There were increases in phasic muscle activity and co-activation in the non-amputated leg. We concluded that, during walking, the unilateral transfemoral amputees (who were analysed in this case series) developed lower GRF in the amputated limb and a longer period of electromyographic activity in the non-amputated limb.

Key words: Amputation; Biomechanics; Electromyography; Gait.

Resumo – *O comportamento da Força de Reação do Solo (FRS) e a atividade eletromiográfica formam uma parte dos dados que caracterizam a biomecânica da marcha. O estudo destes parâmetros é importante para a recuperação da locomoção e para compreensão do impacto da amputação e dos componentes protéticos nos movimentos desenvolvidos no andar. Portanto, esta série de casos tem como objetivo descrever a atividade eletromiográfica e a FRS de amputados transfemorais. Para mensurar a FRS, foi utilizada uma plataforma de força e um sistema de eletromiografia monitorou os músculos vasto lateral, bíceps femoral, tibial anterior e gastrocnêmio lateral da perna não-amputada. As médias das componentes vertical e ânteroposterior da FRS, a atividade eletromiográfica e variáveis descritivas foram analisadas. Foi observado uma diminuição da magnitude da FRS vertical e ânteroposterior e das variáveis descritivas da componente ânteroposterior da FRS durante a fase de propulsão na perna amputada. Houve aumento na atividade fásica muscular e co-ativação na perna não-amputada. Pode-se concluir que os amputados transfemorais unilaterais analisados nesta série de casos desenvolveram menor FRS na perna amputada e longos períodos de atividade eletromiográfica na perna não amputada durante a marcha.*

Palavras-chave: *Amputação; Biomecânica; Eletromiografia; Marcha.*

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INTRODUCTION

Amputation of the lower limbs changes the biomechanics of gait[1-3]. Different lower limb inertial properties[4,5] and a limited capacity to generate internal forces and torques[2,6] are two major locomotion problems facing an amputee when using a prosthetic limb. As a result of transfemoral amputation, there is an attenuation of the GRF in the amputated limb (AL)[7]. To control the sudden prosthetic knee flexion, the gait speed slows, the extension of the prosthetic knee is maintained for up to 40% of the support phase[5], and the time sequence of muscle activation remains the same, compared to normal gait[3], but lasts longer. Moreover, the non-amputated leg (NAL) muscles, especially the hip extensors and ankle plantar flexors[2], generate more joint torque and power to move the body forward.

Previous studies have explored the kinematics[5,7-9] and kinetics[1,2,6,8,9] of the amputee gait, but few studies have described the muscles' activation[3]. It is unknown how transfemoral amputation affects muscle activation during walking, nor what adaptations in muscle activation may occur to accomplish the changes observed in the mechanics of the locomotion system with a prosthetic leg. Analysis of electromyographic activity may describe some of the strategies used by the nervous system to adapt to the amputee condition. Several factors affect this adaptation such as: the amputation level[1], the prosthesis type[9], the muscle reinsertion method into the thigh, the anatomical and functional condition of the remaining muscles and nerves[3], gait rehabilitation strategies[10], stump length[11], and how varied were the motor experiences after amputation.

Although the lower limb amputation incidence is not low[12], because of the many differences in their own adaptation processes, the challenges of gathering several amputee participants for gait analysis are formidable. For example, differences in the related features of the amputation (etiology, amount of time, stump length and circumference) and prosthesis (type, socket, foot, time between first amputation, prosthesis placement, as well as the length of time with current prosthesis) all hinder composing a group with similar features. Although the studies seek to analyze homogeneous samples, it is difficult to standardize the prosthetic components used[3,11] or to assemble a group with similar characteristics such as age and time since amputation[1,2], stump length[3,11], or cause of amputation[9]. Thus, one of the feasible strategies for studying amputee gait is to combine selected cases and analyze what they have in common. Such a "within group" analysis may help to understand the biomechanics of the issue.

The objective of this study, then, is to describe, during the walking, the GRF and electromyographic activity of three transfemoral amputees. The experimental hypothesis is that GRF and EMG parameters change according to the time elapsed since amputation.

Such research is important in establishing gait training as well as in understanding the impact of amputation and prosthetic components on movement during walking.

METHODS

The ethical committee of School of Physical Education and Sport at the University of São Paulo (USP) approved this study (protocol No.37), and each participant signed an informed consent. The participants were three unilateral transfemoral male amputees of traumatic origin.

Participant 1 (P1: 13 years old, 1.66 m, 45.5 kg, time since amputation: 7.3 years, level: proximal thigh, prosthesis experience: 1 month) wore an endoskeletal prosthesis with an ischial total contact containment socket with suction suspension, a Tehlin knee (TK4POC, pneumatic control in the swing phase), and a Springlite foot for one month. Because of a short stump, P1 had problems finding a comfortable prosthesis. After several attempts to fit a prosthesis, the participant began a one-month period of gait training. After his amputation, he played table tennis for 2.4 years on crutches, and for one month prior to beginning this study he played while using his prosthesis (training: 3-5 times/week, 2-4 h/day).

Participant 2 (P2: 30 years old, 1.77 m, 79.4 kg, time since amputation: 1 year, level: mid-thigh, prosthesis experience: 6 months) changed his prosthesis one month prior to this study and is currently training with the new prosthesis. The current prosthesis has an ischial containment total contact socket with suction suspension, a Tehlin knee (TGK4000-mechanical control in the swing phase) and a SACH foot. He practiced discus throwing and putting the shot for two years prior to his amputation (training: 3 times/week, 2-4 h/day), and had restarted his training routine one month before beginning this study.

Participant 3 (P3, 17 years old, 1.75 m, 76 kg, time since amputation: 3 years, level: mid-thigh, prosthesis experience: 2.6 years) wore an endoskeletal prosthetic with an ischial containment total contact socket with suction suspension, a Proteval knee (pneumatic control in the swing phase), and an Endolite foot. He was successfully fitted with prosthesis and has completely adapted to walking and running. He has practiced table tennis with his current prosthesis for 2.5 years (training: 3-5 times/week, 2-4 h/day).

Procedure

A piezoelectric force plate (600 x 900 mm, Kistler 9287A), placed in the middle of a 20-m walkway, measured the anteroposterior and vertical GRF. The force plate and the walkway were covered with a 2 x 20 m non-elastic plastic carpet. A raw electromyography (EMG) signal was recorded (Bagnoli-8 - Delsys, Inc., Boston, MA) using a differential amplification (amplified 1,000 times at a 12-bit resolution) with a sampling frequency of 1 kHz. The EMG bandwidth was limited to between 20 and 450 Hz (CMRR < 92 dB, input impedance $>10^{15}/0.2$ ohm/pF). An analog-to-digital converter (A/D DAS – 1600/1400 Series Keithley Instruments, Inc.) with 16 channels and 12 bit resolution was responsible for data synchronization.

Only the non-amputated leg muscle activities were monitored. After the trichotomy, the bipolar active surface electrodes (Ag/AgCl; 1 cm diameter, 1 cm inter-electrode distance) were placed 1 cm away from the motor point[13] of the m. vastus lateralis (VL), m. biceps femoris (BF), m. tibialis anterioris (TA), and m. gastrocnemius lateralis (GL). To locate the motor point the participants were asked to lie down on a clinical table. We then applied electrical pulses using an OMNI Pulsi-901 (Quark, Piracicaba, SP, Brazil) universal pulse generator on the surface of the skin and above the muscle where even the smallest intensity of current would activate the muscle. The monophasic, quadratic, pulsed current was applied with 7 Hz of frequency – the smallest intensity to activate the motor point. A ground electrode was placed over the patella.

The participants walked straight ahead at a self-selected speed. The speed was determined to be the average velocity to cross a 20 m long walkway. The time was measured with a manual chronometer. The average velocities were 0.63±0.03 m/s (P1), 0.84±0.02 m/s (P2), and 1.08±0.08 m/s (P3). The participants were expected to step over the force plate 15 times with each foot; however, for some trials, the participant did not step correctly over the force plate, resulting in discarded data. Ten AL and nine NAL stances were measured for P1, and twelve AL and eight NAL stances for P2. Finally, 10 AL and 15 NAL stances were measured for P3.

Data analysis

The raw GRF was low-pass filtered (2nd order Butterworth 20 Hz, recursive filter) and normalized in relation to the body weight (BW)[14] of the participant. The raw EMG was represented through a linear envelope[15] and calculated in five steps: off-set removal from the raw EMG, full-wave rectified, low-pass filtered (2nd order Butterworth 5 Hz, recursive filter), the data amplitude normalized in relation to its mean[16], and the time base normalization by the stance phase (% SP)[14]. The beginning and end of the stance phase was determined by the vF. The cutting was done visually and determined during the processing of data.

The following variables were calculated (Figure 1a) from the anteroposterior GRF (apF): the braking phase peak (1apF) and its instant (Δt1apF); the propulsive phase peak (2apF) and its instant (Δt2apF); the braking phase impulse (apFBimp); the propulsive phase impulse (apFPimp); the ratio between the impulses (apFBP); and the stance time (Δtstance). The total vertical impulse (vFimp)[14] was derived from the vertical GRF (vF). Both the pulse duration (the first and last instants the EMG intensity was > 25% peak, initial–final% SP Figure 1b), and the peak instant (from 0 to 100% SP, Figure 1b) were calculated from the EMG. We opted to choose a moderate level of muscle action[17] as a parameter indicative of phasic activity during the stance phase. The linear envelope and variables were calculated using a mathematical function (Matlab software).

The figure illustrates the following variables: the braking phase peak (1apF) and its time interval (Δt1apF); the propulsive phase peak (2apF) and its time interval (Δt2apF); the braking phase impulse (apFBimp); the propulsive phase impulse (apFPimp); the ratio between the impulses (apFBP); and the stance time (Δtstance), which was calculated from the anteroposterior GRF. b) The average of the gastrocnemius lateralis from S3, which represents the studied variables. The EMG intensity, which was established to determine the duration of the pulses (25% of the peak), are represented using the dash dot (-.-). The first and second (start and final) pulses are represented using the '*' symbol.

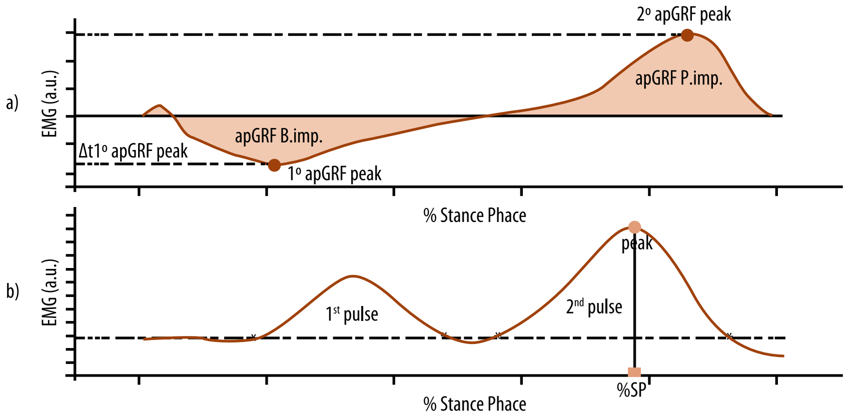

Figure 1. a) The average of the anteroposterior GRF from P3, which represents the studied variables. The figure illustrates the following variables: the braking phase peak (1apF) and its time interval (Δt1apF); the propulsive phase peak (2apF) and its time interval (Δt2apF); the braking phase impulse (apFBimp); the propulsive phase impulse (apFPimp); the ratio between the impulses (apFBP); and the stance time (Δtstance), which was calculated from the anteroposterior GRF. b) The average of the gastrocnemius lateralis from S3, which represents the studied variables. The EMG intensity, which was established to determine the duration of the pulses (25% of the peak), are represented using the dash dot (-.-). The first and second (start and final) pulses are represented using the '*' symbol.

The averages and standard deviations for all those parameters were described and compared across the three participants. Data reported by Rab[18] and Winter[19] were used to compare our results to a pattern of non-amputee walking.

RESULTS

The participants presented different and asymmetrical vF ensemble averages (Figure 2). Therefore, the peaks and inclinations from the vF were not calculated. As a consequence of slow gait and shorter Δtstance, the vertical GRF impulse was lower for the AL (Table 1).

For apF ensemble average curve, all participants presented a biphasic pattern for both limbs (Figure 2); but their parameters were different (Table 1).

During the braking phase, the 1apF and apFBimp were similar for both stances of Participants 1 and 3, whereas in the propulsive phase, the 2apF, Δt2apF, apFPimp, and Δtstance were lower in the AL of all participants (Table 1).

For P2 and P3, the VL was active from the beginning of the stance phase (Figure 3, Table 2) up to 50 and 30% SP (Table 2), respectively. For all three participants (Figure 3, Table 2), the BF activity begins at the weight acceptance (where it reaches its peak), and the TA was active during the weight acceptance and pre-swing. For P2 and P3 (Figure 3, Table 2), the GL presented its first burst (just after the foot-flat phase) to decelerate the tibia rotation, and the second during the propulsive phase. Only for P1, were all muscles active after 60% of the stance phase (Figure 3, Table 2).

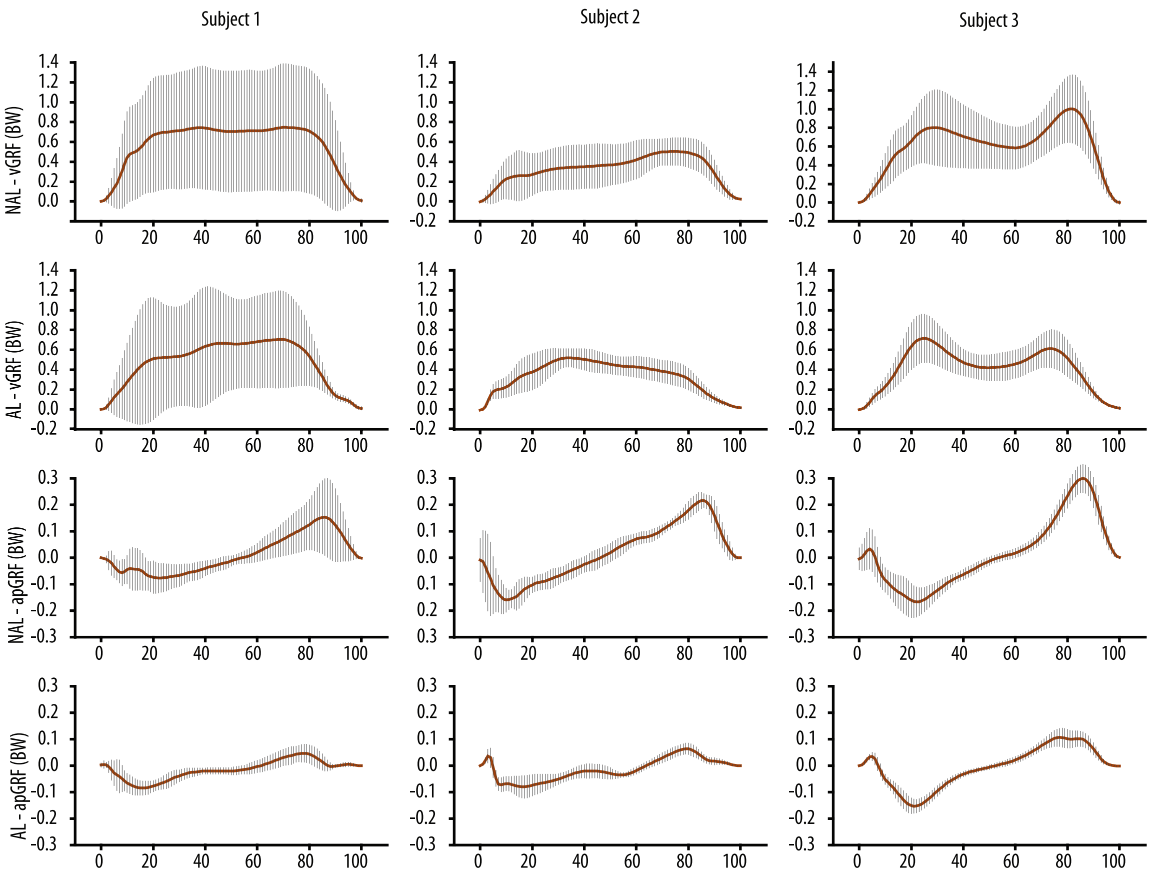

Figure 2. The means and confidence Interval (mean ± 1.96*SD) of the anteroposterior (ap) and vertical (v) GRF [BW (Body Weight)] obtained from the non-amputated limbs (NALs) and amputated limbs (ALs) of Participants 1, 2, and 3.

Table 1. The means and standard deviations of the horizontal (ap) and vertical (v) ground reaction force variables [the braking phase peak (1apF) and its time interval (Δt1apF), the propulsive phase peak (2apF) and its time interval (Δt2apF), the braking phase impulse (apFimp), the propulsive phase impulse (apFPimp), the ratio between the impulses (apFBPimp), the vertical impulse (vFimp), and the stance time (Δtstance)], which was normalized in relation to the body weight (BW) from the non-amputated limbs (NAL) and amputated limb (PL) of Participants 1, 2, and 3.

		AL	NAL			AL	NAL
1apF (BW)	1	-0.07±0.02	-0.09±0.09	Δt1apF (s)	1	0.15±0.04	0.23±0.10
	2	-0.09±0.01	-0.18±0.02		2	0.06±0.04	0.13±0.02
	3	-0.15±0.01	-0.16±0.05		3	0.14±0.09	0.17±0.01
2apF (BW)	1	0.06±0.05	0.17±0.05	Δt2apF (s)	1	0.70±0.10	0.94±0.10
	2	0.06±0.01	0.22±0.01		2	0.67±0.03	0.98±0.02
	3	0.11±0.01	0.30±0.02		3	0.56±0.02	0.69±0.02
ApFBimp (BW/s)	1	0.02±0.02	0.02±0.01	ApFPimp (BW/s)	1	0.01±0.01	0.04±0.01
	2	0.02±0.01	0.04±0.03		2	0.01±0.00	0.06±0.00
	3	0.04±0.06	0.03±0.00		3	0.01±0.00	0.04±0.00
ApFBP	1	4. 13±3.25	1.37±2.48	Δtstance (s)	1	0.88±0.10	1.08±0.12
	2	2.47±0.87	0.60±0.21		2	0.83±0.03	1.11±0.02
	3	2.18±3.16	0.73±0.09		3	0.70±0.02	0.81±0.02
vFimp (BW/s)	1	0.31±0.07	0.71±0.13				
	2	0.29±0.04	0.37±0.09				
	3	0.30±0.05	0.50±0.09				

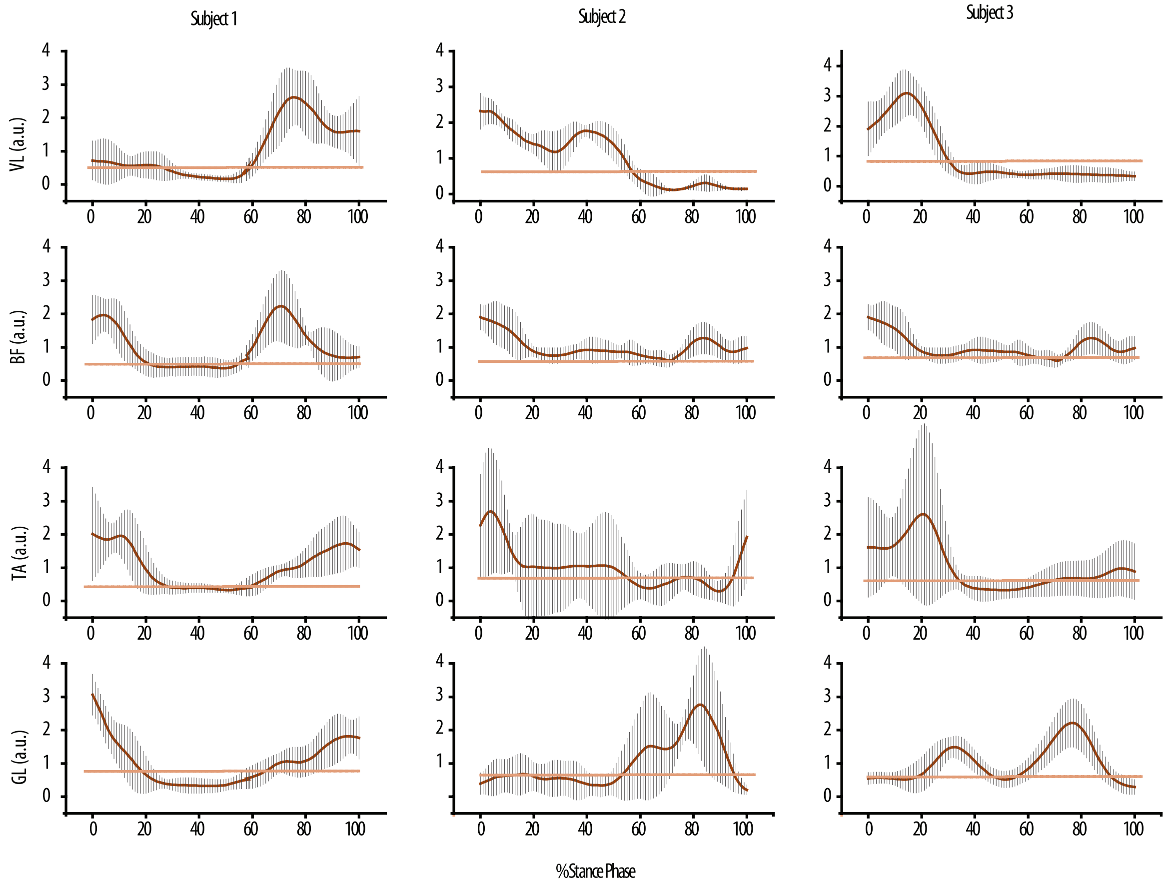

Figure 3. The means and confidence interval (mean ± 1.96*SD) of the VL, BF, TA and GL linear envelopes [a.u. (arbitrary unit)] from the non-amputated limbs (NALs) of Participants 1, 2, and 3.

Table 2. The means and standard deviations of the phasic activity [1st (first pulse %), 2nd (second pulse) and 3rd (third pulse)] and of the time to activation peak (peak) of the VL, BF, TA and GL muscles from the non-amputated limbs (NALs) of Participants 1, 2, and 3.

		1st	2nd	3rd	peak
VL	1	6±11 - 16±6	63±2 – 100		77±6
	2	0 - 50±3			3±4
	3	0 - 30±2			15±2
BF	1	0 - 17±4	58±4 - 90±6		48±34
	2	0 - 59±15	72±3 - 100		3±4
	3	0 - 34±5			10±10
TA	1	0 - 24±3		65±6 - 100	4±6
	2	0 - 44±22	73±4 - 78±3	96±2 - 100	5±5
	3	0 - 32±5		76±13 - 100	13±10
GL	1	0 - 17±4	58±4 - 90±6		71±3
	2	5±4 - 19±2	59±9 - 94±2		76±10
	3	24±4 - 42±2	62±3 - 88±4		77±1

DISCUSSION

The purpose of this case series was to describe the GRF and electromyographic activity of three transfemoral amputees during the act of walking. The combination of dissimilar individual characteristics and their differing gait kinematics led to important adaptations in both the GRF and their phasic muscular activity.

Individual differences among participants and how they coordinate the AL and the NAL might affect vF ensemble averages. The GRF is the most commonly studied external force. This force represents the pattern of acceleration of the whole body center of mass, which is formed by vertical, anteroposterior, and medial-lateral vector-components. During gait, the vF typically has two peaks: the first vF peak occurs immediately after the heel strike and represents the deceleration of segments at the beginning of the stance phase, and the second determines the acceleration upward from the center of mass during the push-off phase[14,20]. Participant 3 was the faster walker, the one to wear his prosthesis longest, and the only one who exhibited the two typical peaks (Figure 2)[14].

The other two participants, on the other hand, who began their gait training only one month prior to the study, presented slower gait speed and asymmetrical vF (Figure 2). The gait is a dynamic activity, and the gait speed is associated with the force applied to the ground. The individual characteristics, such as the condition of the stump, the duration of the amputation procedure, the time it takes to become accustomed to the prosthesis, and the time devoted to gait training[21,22] may all have contributed to the slower gait speed.

As a consequence of slow gait and the shorter Δtstance (Table 1), the vertical GRF impulse was lower for the AL (Table 1), thereby mirroring previous studies[1,23]. The lowest vertical impulse facilitates balance control[24]. Moreover, after amputation, the center of gravity moves nearer to the NAL rather than the AL[1,25]. Lower limb amputees prefer to load their weight on the NAL[1]. After lower limb amputations, the somatosensory input is impaired, and these balance strategies are developed by transfemoral amputees.

The influence of amputation in apF was notable during propulsion. The lack of the plantar flexor muscles and the greater prosthetic knee extension during the weight acceptance stage[5,24] may be responsible for a decrease in forward propulsion. Only in P2, is there an attenuation of 1apF and apFBimp. This behaviour is possibly an attempt to minimize the loading rate. This result is a response to a combination of factors, such as the reduction of walking speed and shorter stance time developed by the participating volunteers.

Muscular activation varied among different participants. The difference suggests that motor strategies account for how comfortable each amputee is with his own prosthesis.

In normal gait[18,19], the major activity and the peak of the VL occurs at the weight acceptance to control the knee flexion and to help the knee extension during the mid-stance. BF activity begins during the terminal swing acting to decelerate the swing leg and continues into the weight acceptance stage. It is at this point that it reaches its peak. TA activation begins at the terminal swing for the ankle dorsiflexion position and controls the foot-flat phase after the heel strike. During pre-swing, the TA activates the ankle dorsiflexion for foot clearance. The GL presented its first burst just after the foot-flat phase to decelerate the tibia rotation. During the propulsive phase, the muscle generates the highest mechanical power to plantar flex.

During weight acceptance, Participants 2 and 3 presented longer VL, BF, and TA bursts; they presented longer GL burst during the propulsive phase. To increase the phasic muscle activity is a motor strategy observed in transtibial[26] and transfemoral[3,8] amputees. The unilateral amputation increases the net joint moments and the power output on the NAL[1]. For example, in non-amputee gait; the GL accelerates the body forward during the terminal stance[19]. The absence of the forward push-off on the prosthetic leg requires a higher amount of power on the NAL[27,28], thereby increasing the duration of muscles bursts. Furthermore, the longer EMG bursts, co-activations, and the reduction of vF are all strategies for achieving balance control. The longer co-activation periods increase the joint stiffness and prevent knee collapse at the load response. However, prolonged co-activation affects the mechanical and metabolic efficiencies of movement which might cause muscular fatigue[18].

The sample size restricted any generalization of our findings. However, it did allow us to explore the individual trends in detail[29]. Because of the uniqueness of each lower-limb amputation and participants' abilities with their custom-made prostheses, this group is likely to exhibit higher inter-individual variability in comparison to adults without amputations.

One other limitation of this study is related to how to determine the beginning and end of the stance phase, which was addressed visually during the processing of data.

For the next study we suggest recording the EMG in the AL. We analyzed the muscles in the NAL because of the difficulty in establishing the correct placement for the EMG electrodes on the stump muscles. A recent study placed the electrodes at alternative locations providing strong EMG signals from both the flexor and extensor muscles of the stump. This placement thereby facilitated their being recorded[30]. This is a strategy that might be followed in future studies.

CONCLUSION

The transfemoral amputees analyzed in this case series exhibited atypical vertical GRF in their AL and reduced anteroposterior GRF during the propulsion phase. The phasic muscle activity in their NAL was increased in comparison to non-amputee walking. The degree of the gait development, the actual speed at which the walking occurred, and the individual characteristics of the studied participants all influenced the results.

Further work is required to establish whether these characteristics are found generally in transfemoral amputees.

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