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Vibration-Based Assessment of Below Knee Prosthesis

Aseel Ghazwan[1*] iD , Muhammed Abdul Sattar[2] iD , Mahmud Rasheed Ismail[3] iD

[1]Department of Biomedical Engineering, College of Engineering, Al-Nahrain University, Baghdad, Iraq
[2]Department of Mechanical Engineering, College of Engineering, Al-Nahrain University, Baghdad, Iraq
[3]Department of Prosthetics and Orthotics Engineering, College of Engineering, Al-Nahrain University, Baghdad, Iraq

*Email: aseel.m.ali@nahrainuniv.edu.iq

Article Info	Abstract
	Human locomotion is achieved through a series of gait cycles. The kinetic analysis of human walking indicates repeated cycles of ground reaction force GRF at the stance phase for each foot. Such a cyclic excitation force can induce vibration that can be transmitted to the human body. In patients using a prosthesis, such as a below-knee Bk prosthesis, the vibration response can cause undesirable effects, leading to discomfort and abnormal gait. In this work, a mathematical model is derived to analyze periodic vibration induced by GRF in the BK prosthesis. In this model, the transfer function of the prosthesis device was evaluated from experimental data with the aid of System Identification SID found in MATLAB software. Fast Fourier transform FFT is used to decompose the periodic GRF excitation and evaluate natural frequencies and responses. It has been found that the excitation frequencies of periodic GRF can affect prosthesis vibration at higher walking speeds and the vibration levels of human organs, including the abdominal, shoulder, lung, and spinal cord regions.

1. Introduction

In the field of below-knee (Bk) prostheses, one of the main concerns in selecting a prosthesis is its response to dynamic Bk behavior, such as impact and vibration. Vibrational analysis has been widely utilized for structural integrity and damage analysis [1]. When considering vibration, the most significant characteristics to examine are the natural frequency, which is influenced by mass, stiffness, and damping. Cappozzo [2] examined the human body's adaptability to vibration stimuli. As walking speed increased, the vertical acceleration spectrum shifted toward organ-resonant frequencies. A synchronized trunk-pelvis movement reduced the head's anteroposterior acceleration and shifted the spectrum to lower frequencies.

This field of research has broad practical applications in engineering, including the clinical effectiveness of whole-body vibration training in subjects with osteoarthritis [3]-[5] and the quantitative monitoring of the osseointegration process [6]-[9]. Vibration-based structural monitoring methods, such as vibration stimulation and acoustic emission analysis, can be used to assess the presence and extent of meniscal tears [10].

However, high-frequency vibrations have been associated with prosthetic joint loosening, low-back disorders, "overloading" disorders, and vascular injuries, where the latter appears

frequency-dependent [11], with changes at frequencies > 30 Hz. Low-frequency vibration may improve motility. Matching the sinusoidal impulse or walking frequency to a vibrational mode improves physiological gait [12]. Lower limb amputees may be more sensitive to vibration. Lower-limb amputees lack some of the muscles that non-disabled individuals use to attenuate high-frequency vibrations from impact forces during locomotion [13]-[16]. The prosthetic limb may exhibit an unusual peak in ground reaction force and loading rate [17]-[20], and a larger fraction of the signal may exceed 25 Hz [21].

The vibration amplitude and frequency of prosthetic components vary, and certain prosthetic features may exacerbate tissue vibration and damage, although they may also enhance mobility and energy return. Therefore, engineering solutions that are adaptable to a range of prosthetic components are required to minimize vibrations. These solutions must be capable of doing so without affecting the prosthetic system's ability to return energy efficiently or produce movement. Chiad et al. [22] investigated the effect of the liner on the vibration of the above-knee (AK) prosthesis. They found that natural frequencies and vibrational amplitudes decrease with the use of a liner and increase compared with the normal leg. Bedaiwi and Chiad [23] investigated the effect of vibration on the BK prosthesis; they aimed to study the effect

of free and forced vibration on the comfort of the amputee. The experimental rig was constructed from a shaking table equipped with a cam mechanism. They found that shifting the liner away from the amputee increased vertical vibration. Furthermore, for varied rotational motor speed and excitation, the maximum frequency of the healthy leg is lower than that of the prosthetic limb and the prosthetic limb without a liner. On the other hand, Jweeg and Jaffar [24] measured the natural frequencies of through-knee prostheses using a hammer test; they reported natural frequencies of 33.2 Hz and 46.8 Hz for the old and design prostheses, respectively. Furthermore, vibration has been shown to increase sensation on the applied limb. Katu et al. [25] investigated the effects of vehicle vibrations. Different vehicle models were used to conduct vibration tests at three speeds and under three road configurations. Fast Fourier Transform (FFT) was used to analyze the vibrational response. The effects of gender on body segments like the head, abdomen, and legs have been considered. Results showed that the prominent effect of vibration occurs in the region between the head and the distal part of the abdomen.

To study whole-body vibration, Kitazaki and Griffin [26] used a servo-controlled electromagnetic exciter with a rigid expander and handrails to determine the free-vibration characteristics of human body parts. In line with this, Kiiski et al. [27] conducted experiments to measure the whole-body vibration. They used frequencies between 10 and 90 Hz and vibration amplitudes between 0.05 and 3 mm as the vibrational source. They found that at frequencies <20 Hz, the human body segment tends to resonate, amplifying amplitudes and frequencies and increasing the acceleration, posing a hazard to a fragile skeleton. Mikhael et al. [28] conducted vibration tests on the participant group using a shaking plate whose amplitude was controlled by a spring-loaded cam driven by a motor operating at 12 Hz. The tests were conducted on a plate on which the participant stood with their feet shoulder-width apart and their hands by their sides. Fairley and Griffin [29] examined 2.5–10 Hz vertical and fore-and-aft whole-body sinusoidal vibration. Their study focused on developing an optimal method for estimating pain caused by vertical and fore-and-aft vibration. They found no bias in experiment outcomes when the test vibration was centred. It was shown that participants under-adjusted the test vibration amplitude, which may be important when calculating similar pain ranges.

Balbinot and Tamagna [30] evaluated bus driver seat transmissibility, with emphasis on comfort and health. Researchers employed accelerometers and a computer system to measure vibrations. Seven bus models were tested on a commercial metropolitan route. The research found that drivers may experience health problems from 4-8 Hz vibrations for 8 hours per day. The chairs exhibited inadequate dynamic performance in the frequency range associated with spinal resonance, hence subjecting the drivers to potential vibration-related issues. It has been demonstrated that human-structure interaction significantly affects structural response [31], and the interaction of walking in vertical and horizontal directions should be considered [32]-[34].

This study aims to evaluate the BK prosthesis transfer function and then determine the anticipated operational frequency ranges that ensure safety. This may be achieved by evaluating the system attributes of the curve connecting the input and output data. The system is exposed to input stimuli to generate an output response. During this process, the input and output signal waves are sampled at equal time intervals. The data is then transformed into the frequency domain using the FFT to obtain the optimal transfer function. The transfer function is used to identify the system's dynamic characteristics, including natural frequency, damping, and gain. Consequently, a comprehensive analysis of the vibration, stability, and control characteristics of the target system may be conducted. System Identification (SID) refers to a software tool used for the purpose of doing such analyses.

2. Materials and Methods

2.1. Mathematical model

Fig. 1 presents the mathematical model of the Bk prosthesis. In view of dynamical behavior, the BK device can be modeled as a mass-stuffiness-damper system with m, k, and c representing the mass, stiffness, and damping, respectively. The periodic GRF, $F(t)$, can impose forced vibration response, $x(t)$, Depending on the Bk transfer function $H(\omega)$.
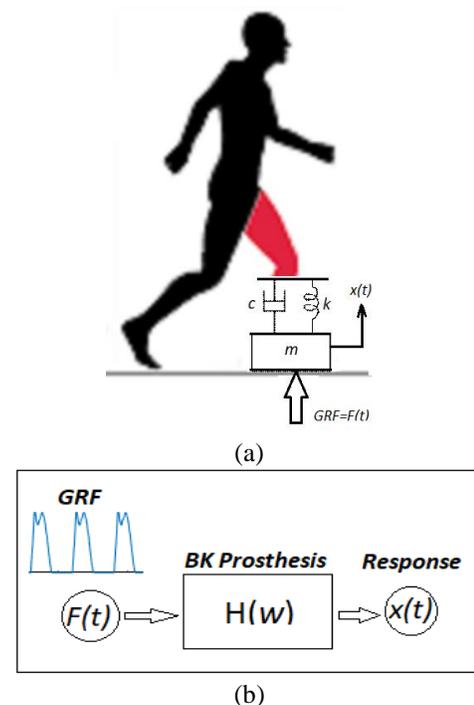

(a)

(b)

Figure 1. (a) Illustration of a forced damped vibration system. **(b)** A mathematical model of BK vibration.

The dynamic model can be written as:

$$M\ddot{x} + c\dot{x} + kx = f(t) \tag{1}$$

Where M, k, and c represent the mass, stiffness, and damping, respectively. Equation (1) can be arranged in a more convenient standard form as follows:

$$\ddot{x} + 2\zeta\omega_n\dot{x} + \omega_n^2 x = f(t)/M \tag{2}$$

The transfer function H_n of the prosthesis can be written as [35]:

$$H_n(\omega) = \left[\frac{1/k}{1 + i2\zeta\left(\frac{n\omega}{\omega_n}\right) - \left(\frac{n\omega}{\omega_n}\right)^2}\right] \tag{3}$$

Where n represents the number of vibration modes, n =1, 2, 3 ….

The excitation periodic GRF, $F(t)$ can be decomposed using the Fourier transform as follows [35]:

$$F(t) = \frac{a_o}{2} + \sum_{m=1}^{\infty} a_n \cos\cos(m\omega t) + b_m \sin\sin(m\omega t) \tag{4}$$

Where;

$$a_0 = \frac{2}{T}\int_0^T F(t)dt$$

$$a_0 = \frac{2}{T}\int_0^T F(t)\cos\cos(m\omega t)\,dt \text{ , and}$$

$$b_0 = \frac{2}{T}\int_0^T F(t)\sin\sin(m\omega t)\,dt$$

Where m is the number of Fourier terms. Finally, the response $x(t)$ can be evaluated from the following [35]:

$$x(t) = f(t)\,H_n(\omega) \tag{5}$$

Combining (3), (4), and (5) gives the following equation for the vibration response [35];

$$x(t) = \frac{a_o}{2k} + \sum_{m=1}^{\infty}\sum_{n=1}^{\infty}|H_n(\omega)|[a_n \cos\cos(n\omega t + \phi_n) + b_n \sin\sin(n\omega t + \phi_n)] \tag{6}$$

Where; $|H_n(\omega)| = \left[\frac{1/k}{[1-(\frac{n\omega}{\omega_n})^2)]^2 + [2\zeta(\frac{n\omega}{\omega_n})^2]^{0.5}}\right]$ and $\emptyset_n = tan^{-1}\left[\frac{2\zeta\left(\frac{n\omega}{\omega_n}\right)}{[(1-(\frac{n\omega}{\omega_n})^2)]}\right] \tag{7}$

In summary; Equation 3 can be used for the evaluation of the BK dynamical parameters, namely, ω_n, ζ and k. Equation (4) for evaluating the excitation frequencies $n\omega$ and (6) for the evaluation of vibration response.

2.2. Participant

This research included the participation of a 47-year-old male who had a transtibial amputation of his left leg 18 months ago as a consequence of diabetes. Table 1 provides a comprehensive overview of the prosthetic limb attributes used by the individual with limb loss.

Table 1. Demographic characteristics of the participant.

Variable	Amputee subject's data
Age (year)	47
Height (m)	1.78
Body weight (kg)	74
Body mass index (kg/m²)	23.4
Prosthetic type	Below-knee prosthesis
Amputated leg	Left Leg

Apart from the amputation, the subject reported no neurological or muscle health issues in the present or the past. The research was approved by the Internal Review Board of Al-Nahrain University, and the subject was selected from the Baghdad Artificial Limbs Center after providing written informed consent.

2.3. Experimental Data

The experimental data were collected at the Prosthetics and Orthotics Engineering Department of Al-Nahrain University. This study is a single-visit experiment. The experimental protocol consisted of two consecutive steps:

The first step is to evaluate the BK prosthetic transfer function. The vibration impact hammer test was used to produce an impact force at the input point. A plastic hammer was used to strike the foot, whilst the voltage obtained from the patches was utilised to determine the vibration modes via analysis of the transmissibility function. The evaluation of the vibration attenuation capacity included subjecting the physical foot to hammer impacts to measure the transmissibility function. The output reaction was detected using the accelerometer. The response was amplified using a charge amplifier. The signals received as input and produced as output were displayed on an oscilloscope and then exported as comma-separated values (CSV) files to the computer. Fig. 2 illustrates the approach used for estimating the transfer function.

Fig. 3 presents the instrumentation and testing used to evaluate the transfer function data using a vibration shaker and an accelerometer. The data from the experiment are fed to MATLAB via System Identification SID tools as shown in Fig. 4.
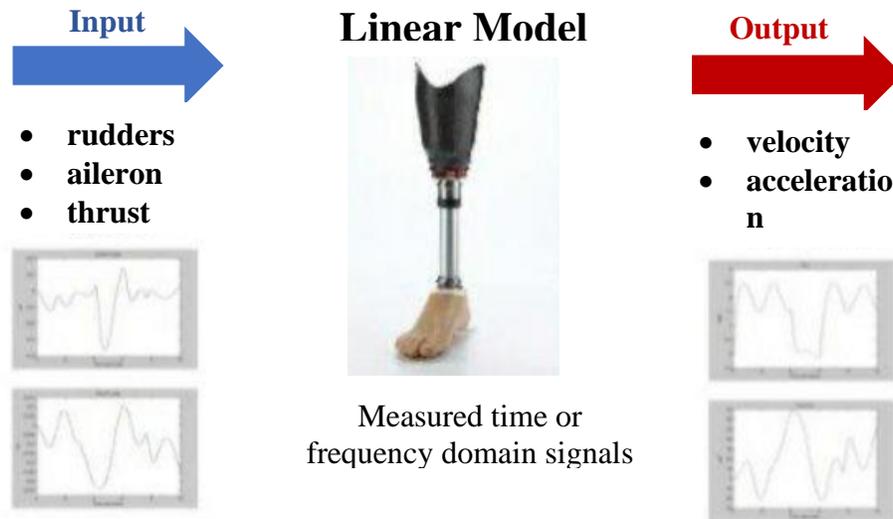
Input

Linear Model

Output

- **rudders**
- **aileron**
- **thrust**

- **velocity**
- **acceleration**

Measured time or
frequency domain signals

Figure 2. Schematic illustration for the approach used for estimating the transfer function.

Figure 3. Vibration test for estimating the transfer function of the prosthesis.

The second step is to evaluate the ground reaction force (GRF) acting on the prosthetic leg. The participant was instructed to walk a 6-meter pathway at their own pace. The GRF was captured using a force platform (AccuGait), which recorded the forces exerted from heel strike to toe-off. The AMTI's Net Force/Bioanalyze software was used to analyse the data from the AccuGait platform at a 600 Hz data set per second, as shown in Fig. 5. The GRF's were normalised to the stance phase of each gait cycle, meaning they were translated into a percentage relative to the subject's body weight. Three typical stance phases were averaged to produce group averages for the values at the first peak, second peak, mid-stance, impulse, and stance time.

3. Results and Discussions

The CSV data from the Hummer test are used to estimate the transfer function of the BK prosthesis. Firstly, the data were exported to the MATLAB workspace, and then system identification SID is employed for this purpose [36].

The estimated transfer function is [36]:

$$H_1(\omega) = \left[\frac{2.127 x 10^{-7}}{1+2i0.2214\left(\frac{\omega}{307.87}\right)-\left(\frac{\omega}{307.87}\right)^2}\right] \qquad (8)$$

From (8), the following dynamical parameters of the BK prosthesis can be easily found as follows:

$\omega_n = 307.87$ rad/s, $\zeta = 0.2214$ and k= 9370 N/m.

Fig. 6 shows the MATLAB SID results. The walking data was saved in *.csv format. MATLAB was used to process motion data for the prosthetic leg, i.e., to decompose the periodic GRF. The difference between model outcomes and the experimental data is evaluated using a goodness-of-fit (GOF) criterion. The best fit with Goodness of fit, SSE= 24.67, Confidence bounds 95%, R2= 0.9981, and RMSE= 0.2331 using the FFT fitting model as shown in Fig. 7.
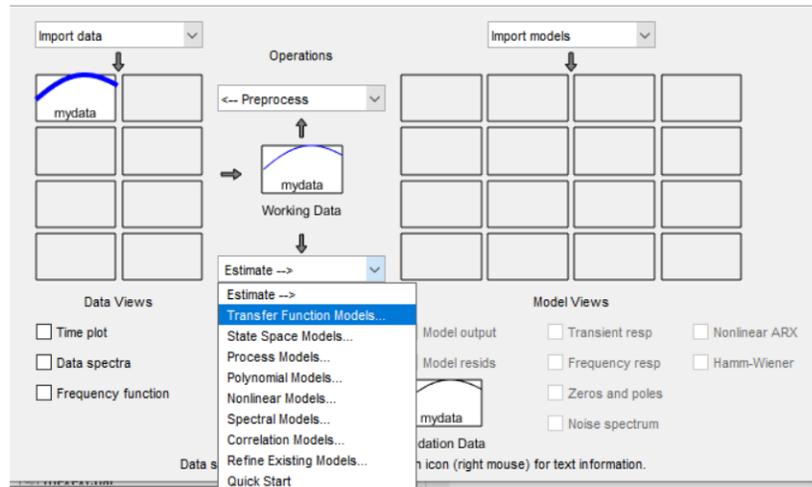
Figure 4. System Identification Window in MATLAB.

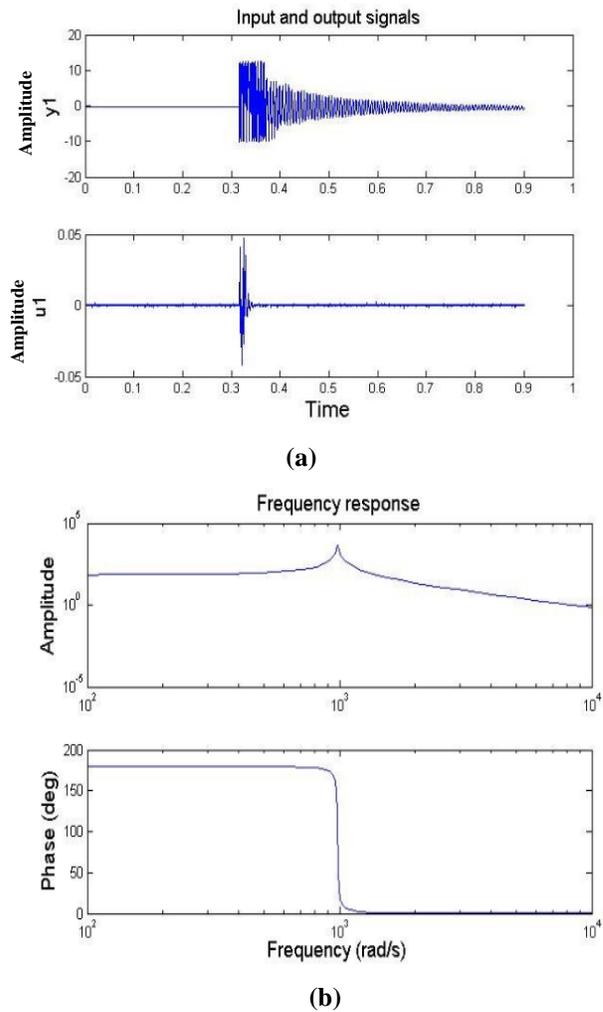**Figure 5.** Participant walks across the FP to collect the GRF.

(a)

(b)

Figure 6. MATLAB SID for BK prosthesis **(a)** Time domain **(b)** Frequency domain.
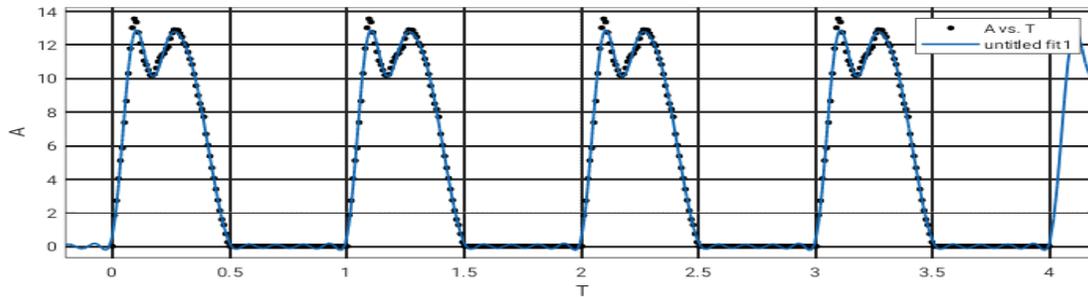
Figure 7. GRF fitting MATLAB model.

The resulting fitting equation for eight terms is:

$$f(t) = a_0 + a_1 \cos(\omega) + b_1 \sin(\omega) + a_2 \cos(2\omega)$$
$$+ b_2 \sin(2\omega) + a_3 \cos(3\omega)$$
$$+ b_3 \sin(3\omega) + a_4 \cos(4\omega)$$
$$+ b_4 \sin(4\omega) + a_5 \cos(5\omega) + b_5 \sin(5\omega)$$
$$+ a_6 \cos(6\omega) + b_6 \sin(6\omega) + a_7 \cos(7\omega)$$
$$+ b_7 \sin(7\omega) + a_8 \cos(8\omega) + b_8 \sin(8\omega)$$

Where the coefficients a_1 to a_8, b_1 to b_8 and ω are given in the following Table 2. The FFT plot of the GRF is further checked by using the SIGVIEW program, taking into account that the frequencies are given in Hz rather than in rad/s. The resulting plot is shown in Fig. 8.

By applying (6), the response of the BK prosthesis can be found as:

$$x(t) = 265.73195 + \sum_{n=1}^{8}|H_n(\omega)|[an \cos \cos (n6.27t + \phi) + b_n \sin \sin (n6.27t + \phi)] \ (9)$$

Where, $|H_n(\omega)|$ and ϕ can be found from (7) according to the BK prosthetic parameters $\omega_n = 307.87$ rad/s, $\zeta = 0.2214$ and k=9370 N/m.

It is important to investigate the effects of prosthesis vibration-induced resonance frequencies on human organs, as shown in Table 3.

Table 2. Fourier Coefficients of periodic GRF.

a_1	a_2	a_3	a_4	a_5	a_6	a_7	a_8	a_0
1.026	-2.24!	0.2242	-0.8761	-0.7314	-0.6809	-0.2629	-0.1351	4.367
b_1	b_2	b_3	b_4	b_5	b_6	b_7	b_8	ω
6.617	0.8597	0.7212	0.9643	0.229	-0.0702	-0.2737	-0.1776	6.27

The fundamental excitation frequency due to GRF from Table 2 is 6.27 or (about 1 Hz), which is far from any resonance effect on human organs; however, the harmonic frequencies for higher modes (nω) are 2,3,4….8 Hz, which can affect many organs (Table 3), namely, the abdominal, shoulders, lungs, and spinal cord regions. These effects originate from vibration modes 5 to 8, which have small amplitudes and will not cause patient discomfort. At fast walking or running gait, excitation frequencies can increase, resulting in more hazardous vibration effects on human organs. Additionally, comparing the GRF excitation frequency with the prosthesis's natural frequency (307.03 rad/s, approximately 49 Hz) indicates that the prosthesis is out of resonance with the excitation frequency.

Figure 8. FFT of GRF obtained from the SIGVIEW program.

Table 3. Resonant frequencies of human organs [29].

Organs	Resonance frequencies (Hz)
Head	20-40
Spinal Column	8
Chest Wall	60
Abdominal	4-8
shoulders	4-8
Lungs	4-8
Hands and arms	20-70
Ocular Globe	60-90
Maxilla	100-200

4. Conclusions

The current study has shown that the frequencies at which periodic GRF are generated may impact the vibration of prostheses at faster walking speeds. Additionally, these frequencies can influence the intensity of vibration experienced by various human organs, including the abdomen, shoulders, lungs, and spinal cord. The analysis reveals two key findings. Firstly, the natural frequency of the BK prosthesis used in the experiment is 307.3 r/s, which is much greater than the excitation frequency of 6.27 r/s. This indicates that the prosthesis can function without the potential for resonance induced by the periodic GRF during walking. In addition, the periodic vibration frequencies of the GRF have a minimal impact on certain amputees' organs, particularly the abdominal region, the shoulder, the lungs, and the spinal cord. Therefore, it is essential to account for the dynamic features of the BK prosthesis to prevent resonance when studying the impact of vibration on human organ systems during fast walking or running.

Acknowledgements

Abbreviations

BK	Below-knee
c	Damping coefficient of the prosthesis
FFT	Fast Fourier transform
GRF	Ground reaction force
k	Stiffness constant of the prosthesis
M	Mass prosthesis
m	Number of Fourier terms
n	Number of vibration modes
SID	System identification
ϕ	Phase angle
ω	Excitation frequency
ω_n	Natural frequency
ζ	Damping ratio

Conflict of interest

The authors have no conflicts of interest to declare.

Author Contribution Statement

Aseel Ghazwan contributed to the conceptualization, analysis of experimental results, and writing of the original draft.

Muhammed Abdul Sattar contributed to the verification of the analytical methods and supervised the findings of this work.

Mahmud Rasheed Ismail contributed to experimental and investigative work. All authors discussed the results and contributed to the final manuscript.

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