Knee Trajectory Modulation for Impact Reducing of Lower Limb Exoskeletons

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Abstract—lower limb exoskeletons have gained considerable interest in the past several years thanks to its effectiveness in rehabilitation and walking assistance. During walking, repetitive and excessive foot-ground impact force makes the patient uncomfortable and even leads to secondary damage. In this paper, we proposed a humanoid method to reduce the Peak of Ground Reaction Force (PGRF) by Knee Trajectory Modulation (KTM). The main idea is to imitate the humanoid passive motion of the compliant leg by modulating the active knee trajectory of the exoskeleton. Different from traditional shock absorption methods by passive mechanical devices, the proposed method enables adjustment of leg stiffness and without bulky mechanical structures. We demonstrated the effectiveness of the proposed KTM in the AIDER lower limb exoskeleton system. Experimental results indicate that the proposed KTM can effectively reduce the PGRF.

I. INTRODUCTION

Lower limb walking assistance exoskeletons are wearable robotic systems that integrate human intelligence and robot power, which have gained considerable interest in past decades. Considering clinical trials have demonstrated the effectiveness of lower limb exoskeletons in rehabilitation applications for Spinal Cord Injury (SCI) patients. Including improvement of bone mineral density decline, common deep vein thrombosis, sores, and various other complications [1][2][3].

To ensure a safe gait, most lower limb exoskeletons are fixed following predefined gaits. However, predefined gaits exhibit rigid leg movements when swing feet repeatedly strike the ground and results in large impact force. As shown in Fig. 1, the foot-ground impact force \(\hat{F}\) acts on the human-exoskeleton system. The force and energy would continuously be transmitted from the plantar surface to the body and ultimately absorbed by soft tissue, muscles, etc. This not only affects the comfort experience of patients, but also seems a risk factor that causes tissue injury and pain [4][5]. On the one hand, SCI patients have no muscular function to reduce the impact force [6], in previous experiments with exoskeleton-assisted walking, able-bodies walk roughly three times faster than patients at approximate vertical ground reaction forces [3]. On the other hand, low bone mineral density [7] implies that SCI patients are potentially more vulnerable to impact force than normal persons. The damage of impact force mainly comes from the maximum force value, i.e. the Peak of Ground Reaction Force (PGRF). Thus, reducing the PGRF by exoskeletons becomes more critical.

Several approaches have been developed to reduce the PGRF. Zelik et al. proposed to reduce the PGRF by trailing ankle push-off just before contralateral foot strike [8][9], which can reduce the vertical
velocity of Center of Mass (CoM) to reduce potential PGRF when feet strike. However, the ankle should be powered to provide net positive work for preemptive ankle push-off [10], which cannot be used for the exoskeleton with passive ankle joints.

A basic idea to reducing the PGRF and absorb shock energy is configuring shock-absorbing soles [11], but shock absorbed by soles is limited because of the limited thickness and damping. In addition, there have been various mechanical devices to address this issue. Jun et al. [12] develop a lower-extremity wearable link mechanism for shock absorption in extreme environments. Jeongsu et al. [13] designed a shock absorption mechanism on the shank for a powered exoskeleton, which can reduce the PGRF effectively with a specific gait. In general, mechanical shock absorption always brings complex and bulky structural, thus limiting their application in exoskeletons. Moreover, it is difficult to adjust the stiffness and damping parameters of the mechanism to adapt to different impacts.

Compliant leg motion plays a central role in human walking [14][15], among which stiffness and damping in human legs are crucial for reducing joint and limb damage on impact [16][17]. Inspired by this, we proposed a walking shock absorption strategy by online Knee Trajectory Modulation (KTM) for lower-limb walking assistance exoskeletons. Firstly, a Shock Absorption Model (SAM) is proposed by modeling the human-exoskeleton as a spring-damping system, which aims at imitating the controllable stiffness of the human leg when feet strike and produce the compliant leg motion. Secondly, a Trajectory Modulator (TM) is proposed to combine the motion of SAM and clinical gait, which can convert the SAM motion to knee motion via leg geometry and converge it to clinical gait after shock absorption. As a result, modulated gaits can be used to reduce the PGRF, and the main contributions can be summarized as:

- The SAM based on the humanoid spring-damping system has been proposed for walking assistance exoskeletons.
- A TM that combines the motion of SAM and clinical gait for reduced PGRF gait planning has been proposed.
- The proposed approach has been verified by the AIDER exoskeleton.

The following sections are organized as follows: Section II proposed the implementation of KTM. In Section III, experimental results on a lower-limb walking assistance exoskeleton are presented. Finally, we concluded the paper in Section IV and suggested future work.

II. METHODS

In this section, KTM will be introduced. The framework of the proposed approach is shown in Fig. 2. Clinical gaits can be collected from healthy subjects walking on level ground by a motion capture system. The CoM and Ground Reaction Force (GRF) information can be computed with sensor data sampled from the human-exoskeleton system. In the KTM, the Feet Strike Detector (FSD) can detect the feet strike moment of the swing leg with GRF data. At the strike moment, the SAM can calculate the spring-damping system based motion of the strike leg. Then
the TM can online modulate the knee trajectories by the motion and the clinical knee trajectories. The clinical hip trajectories and modulated knee trajectories can be applied in joint controllers.

A. Feet Strike Detector

Due to the pilot disturbance, the passive ankle motion of the supporting leg cannot be completely limited during single support phases. Thus the duration of single support phases may be slightly changed, i.e. the strike time of the swing foot is not fixed in walking cycles. Therefore, it is necessary to detect the swing feet strike moment to online modulating the knee trajectories.

The use of the plantar pressure system to collect GRF data and determine the feet strike moment is effective [18][19]. Therefore, we adopt the same approach. Set a pressure threshold $F_t$ in the FSD, the time is determined as the feet strike moment when the GRF is detected to be greater than this threshold.

B. Shock Absorption Model

At the feet strike moment, as shown in Fig. 3, assuming that the exoskeleton is restricted to move in the 2-D sagittal plane. Generally, the CoM of the human-exoskeleton system can be located on the hip [20]. To deal with impact forces transmitted from the ground to CoM, we consider the SAM as a spring-damping system, which shows shock absorption behavior [13]. There is a virtual spring with stiffness $k$ and a virtual damper with damping $c$ between the hip and ankle. The motion of the SAM along the direction from ankle to CoM, and the leg length $x$ should follow the motion of SAM after feet strike, which can reduce the PGRF. The passive ankle joint is not described in detail because it is weakly related to the initial ground contact impact.

Define the time when the swing foot strike as $0_s$. The response of the spring-damping system $\Delta x(t) = x(0) - x(t)$ can be described as:

$$\Delta \ddot{x} + 2\zeta \omega_n \Delta \dot{x} + \omega_n^2 \Delta x = \frac{f_{st}}{m},$$

where the damping ratio $\zeta = \frac{c}{2\sqrt{km}}$, and the natural frequency $\omega_n = \sqrt{\frac{k}{m}}$. Consider the exoskeleton and the wearer as a whole, and the $m$ is the mass of the human-exoskeleton system, the magnitude of the bodyweight force $f_{st}$ is:

$$f_{st} = \frac{mg}{\cos \theta_2} - F_{pre},$$

where $\theta_2$ is the angle between the support leg and the vertical direction, as shown in Fig. 3, and the angle can be calculated by the data of the inertial measurement unit (IMU) in the backpack of the exoskeleton and the hip encoder data. $F_{pre}$ is the preload of the spring, which can adjust the max knee flexion angle. Reference to walking experiments, the variation of knee flexion angle should be limited to about 20 degrees [21]. Therefore, the $F_{pre}$ is introduced to $f_{st}$.

Fig. 4 shows the ideal SAM response $\Delta x(t)$ after
feet strike. The spring and damper can effectively absorb the impact energy, extend the buffering time, and thus reduce the PGRF. When $t = \frac{4}{\xi\omega}$, the leg length is almost compressed to the shortest, and the shock absorption process is completed.

More specifically, the SAM should be over-damped to avoid vibration when the leg strike, thus $\xi \geq 1$. To stabilize the system as quickly as possible, the damping ratio should be selected as small as possible. To improve the bionics of the SAM, The configuration of the stiffness is considered from the study of walking experiments. Hong et al. found that the human leg stiffness $k$ consistently increased as a function of the average walking speed $v$ [22], and the function can be approximated as:

$$k = 12000v + 4000,$$

In the SAM, we introduce a similar stiffness variation model. Then we can calculate the damping based on the stiffness and damping ratio. In this way, the SAM shows humanoid leg compliance when the swing leg strike.

To enable the exoskeleton legs to track the motion of the SAM during shock absorption, the dynamic response of the SAM is calculated. The response $\Delta x(t)$ consists of two parts: the response by the applied bodyweight force $\Delta x_1(t)$ and the response by the contact velocity $\Delta x_2(t)$.

1) **Response by the applied bodyweight force**: The SAM as an over-damped single-degree-of-freedom system, and the response by the applied bodyweight force can be defined as $\Delta x_1(t)$:

$$\Delta x_1(t) = \frac{\Delta}{k} \left(1 - c_1 e^{-\left(\xi - \sqrt{\xi^2 - 1}\right)\omega t} + c_2 e^{-\left(\xi + \sqrt{\xi^2 - 1}\right)\omega t}\right),$$

where $c_1 = \frac{1}{2\sqrt{\xi^2 - 1}(\xi - \sqrt{\xi^2 - 1})}$, $c_2 = \frac{1}{2\sqrt{\xi^2 - 1}(\xi + \sqrt{\xi^2 - 1})}$. Note that the over-damped systems with damping ratios $\xi \geq 1$ to ensure that $c_1$ and $c_2$ can be calculated.

2) **Response by the contact velocity**: Consider the initial conditions when feet strike. The contact velocity $v_0$ is the component of the CoM velocity $V_0$ along the direction of the leg. Where $v_0 = V_0 \cos \theta_1$, among which $\theta_1$ is the angle between the CoM velocity and the direction of the support leg, as shown in Fig. 3, $V_0$ can get from the IMU data. $v_0$ also affects the motion of SAM in eq. (1), the response by contact velocity is defined as $\Delta x_2(t)$:

$$\Delta x_2(t) = e^{-\xi\omega t} \left(c_3 e^{\sqrt{\xi^2 - 1}\omega t} + c_4 e^{-\sqrt{\xi^2 - 1}\omega t}\right),$$

where $c_3 = \frac{v_0}{2\omega\sqrt{\xi^2 - 1}}$, and $c_4 = -c_3$.

3) **Response of SAM**: SAM is a linear system, eq. (4) and eq. (5) can be used to calculate the response of SAM $\Delta x(t)$ and the motion of strike leg $x(t)$ by the superposition principle:

$$\Delta x(t) = \Delta x_1(t) + \Delta x_2(t)$$

$$x(t) = x(0) - \Delta x(t).$$

In this way, the SAM combines the shock absorption performance of the spring-damping system with humanoid stiffness adjustment.

C. **Trajectory Modulator**

To obtain the modulated knee trajectory both simple and comfortable, we combine the movement $\Delta x(t)$ of SAM with clinical knee trajectories $\theta(t)$. The modulated knee trajectory consists of three components: Firstly, we convert the SAM motion to knee motion via leg geometry and get the transformed trajectories $\theta_b(t)$. Secondly, remove the original vibration of clinical knee trajectories after feet strike, and get the vibration removed knee trajectories $\theta_b(t)$. Thirdly, converge the knee motion to clinical gait after shock absorption via response of the reverse applied bodyweight force $\theta_b(t)$. The modulated knee trajectory $\theta'(t)$ can be described as:

$$\theta'(t) = \theta_a(t) + \theta_b(t) + \theta_c(t).$$

We then introduce each part of the modulated knee trajectory $\theta'(t)$, according to the simple geometric relationship, the SAM-based leg motion $x(t)$ can be transformed into the knee motion:

$$\theta_b(t) = \arccos \left(\frac{l_c^2 + l_t^2 - x(t)^2}{2l_t l_c}\right),$$

where $l_c$ and $l_t$ are calf length and thigh length, respectively.

In order to make the standing leg movement as close as possible to SAM after feet strike, we consider remove the original vibration of clinical knee trajectories and replace it with eq. (9), as shown in fig. 5. The vibration of clinical knee trajectories $\theta(t)$ was compressed within the standing phase and get the vibration removed knee trajectories $\theta_b(t)$:

$$\theta_b(t) = \begin{cases} \lambda \theta(t) + \theta(0)(1 - \lambda), & (0 \leq t < \omega\%T_n) \\ \theta(t), & (\omega\%T_n \leq t < T) \end{cases}.$$
where $\lambda$ is the compress factor and can be set as empirical values 0.1. $T_n$ is the gait period. $\omega$ is the duration of compression in the gait period. Note that in order to maintain the continuity of the compressed trajectory, $\theta(0)$ in the original clinical gait needs to be equal to $\theta(\omega\%T_n)$.

Then, to make the standing leg motion return to the clinical knee trajectories $\theta(t)$ after shock absorption, we input a reverse bodyweight force $(-f_{st})$ to the SAM at the end of the shock absorption (when $t = \frac{4}{\xi\omega}n$), and get the reverse trajectories:

$$
\theta_c(t) = \begin{cases} 
0 & (0 \leq t < \frac{4}{\xi\omega}n) \\
-arccos\left(\frac{l_1^2+l_2^2-(s(0)-\Delta s_1(t))^2}{2l_1l_2}\right) & \left(\frac{4}{\xi\omega}n \leq t < T_n\right) 
\end{cases},
$$

finally, use the eq. (9), eq. (10) and eq. (11) to modulate the clinical knee trajectories after touchdown, as shown in eq. (8). In this way, the $\theta'(t)$ combines the motion of clinical gait and human leg stiffness based spring-damping system, which can simulate the motion of the spring-damping system at the beginning of standing phases and return to clinical gait after shock absorption, we can get the modulated knee trajectory $\theta'(t)$ in the support phase.

III. EXPERIMENTAL RESULTS AND DISCUSSION

A. Description of the AIDER

The AIDER exoskeleton system is designed for walking assistance of paraplegic patients, as shown in Fig. 6. The AIDER system is actuated by four DC servo motors in hips and knees, and ankles are passive flexible joints. The exoskeleton system weighs about 25 kg. A distributed control system with four node controllers and the main controller is built for real-time control applications of the AIDER system. AIDER has three kinds of sensors. Joint angle encoders are interpreted in hip and knee joints to measure the joints angle. IMU sensors in the backpack are employed to measure the orientations of the pilot’s trunk and calculate the angle $\theta_1$ and $\theta_2$. The force sensors are installed on the shoes of the AIDER system to estimate gait phases. Pilots can operate AIDER with the button on crutches directly.

B. Experiment setup

A paraplegic patient was recruited to test the proposed approach on AIDER. The movement of his arms are not restricted to ensure that he can keep balance with the help of crutches. We collected some gait trajectories from healthy subjects walking on the level ground by motion capture system Vicon. Parametric values of the experiment are given in Table I. Two experimental clinical gaits with pilot-selected walking speeds were tested, the average walking speeds are about 0.3 m/s and 0.6 m/s, respectively. For each clinical gait, the pilot firstly walked with clinical gaits. Subsequently, walked with modulated gait. For modulated gait, GRF threshold $F_t = 15 N$ and damping ratio $\xi = 1.2$ are set as empirical values. The spring preload $F_{pre} = 600$ was set to avoid excessive knee flexion when KTM working and cause the opposite swing foot to scratch the ground. To observe the changes of the GRF according to the presence of the KTM, all the other settings were kept in the same
condition. The experimental data, such as joint angles and GRF, were recorded in the embedded computer.

| Description         | Value |
|---------------------|-------|
| calf length $l_c$   | 44 cm |
| thigh length $l_t$  | 41 cm |
| pilot weight $m_p$  | 70 kg |
| exoskeleton weight $m_r$ | 25 kg |
| GRF threshold of TE $F_t$ | 15 N |
| damping ratio of SAM $\xi$ | 1.2 |
| spring preload of SAM $F_{pre}$ | 600 N |

### Table I

**Experiment Settings.**

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|---------------------|-------|
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C. Results and discussion

Fig. 7 and Fig. 8 show the GRF with two different walking speeds, among which (a) and (b) represent the GRF of clinical gait (CG) and modulated gait (MG), respectively. The joint angles of the two walking speeds corresponding to Fig. 7 and Fig. 8 are shown in Fig. 9, respectively. For comparison, we aligned the gait period of CG to that of MG.

1) **Effectiveness of MG:** The performance of KTM was evaluated by comparing the PGRF between two gaits. Enormous force duration can be analyzed as standing phases, e.g. 3.7s to 5.5s in Fig. 7 (a). As expected, PGRF often appears near the strike moment, e.g. 3.9s in Fig. 7 (a) and 2.5s in Fig. 8 (a). This is in agreement with that obtained by Jeongsu et al. [13]. Since the CG and the leg motion can not be changed when the shock occurs, and the leg can be considered rigid at the moment of strike. The shock energy was absorbed by the human-exoskeleton body in a moment and leads to excessive PGRF. In contrast, the PGRF when the pilot walked with the MG shows reduced PGRF, as shown in Fig. 7 (b) and Fig. 8 (b). Since the leg motion contains a compliant mechanism when the shock occurs, a part of shock energy was absorbed by leg compliant motion rather than the human-exoskeleton body. Table II shows a comparison among the four conditions by comparing the mean values and standard deviation values, the results suggest that the PGRF after strike had been reduced by implementing the proposed KTM method.

The decline of PGRF are close at two walking speeds, but there are still some residual impacts with faster walking speeds, e.g. 2.8s and 4.7s in Fig. 8 (b). There are two causes for these impacts: Firstly, due to the sensor accuracy, tracking errors of the controllers, etc, the standing leg could not fully imitate the movement of the SAM. Secondly,

![Fig. 7. GRF data obtained from shoes sensor when walking at about 0.3 m/s. (a) The GRF data when walking with clinical gait. (b) The GRF data when walking with modulated gait.](image1)

![Fig. 8. GRF data obtained from shoes sensor when walking at about 0.6 m/s. (a) The GRF data when walking with clinical gait. (b) The GRF data when walking with modulated gait.](image2)

### Table II

**Mean PGRF at Two Walking Speeds.**

| Walking Speed | PGRF of CG (mean ± SD) | PGRF of MG (mean ± SD) | Decline  |
|---------------|------------------------|------------------------|----------|
| 0.3 m/s       | 1278.4 ± 133.9 N       | 1046.8 ± 81.6 N        | 18.1%    |
| 0.6 m/s       | 1425.4 ± 179.4 N       | 1158.5 ± 120.6 N       | 18.7%    |
we adjust stiffness and damping of SAM only by walking speed and the parameters are not optimal for shock absorption. Although walking speed has the most significant effect on knee stiffness. However, there are other factors, such as age [22], weight, and height, that are not taken into account.

2) Usability of MG: Micro force duration can be analyzed as swing phases, e.g. 2.5s to 3.7s in Fig. 7 (a). There are no obvious GRF mutations during swing phases in Fig. 7 (b) and Fig. 8 (b), which suggest that swing feet did not scratch the ground when walking with MG. As shown in Fig. 9, the knee trajectories of MG were online modulated after strikes, while the hip trajectories were consistent with CG. After feet strike, there are subtle differences in the max knee angle modulated in each MG because of the variation of response by contact velocity. the maximum flexion angle of the knee joint is 22.3° at 6.6s in Fig. 9 (b), which is close to the maximum knee flexion angle of about 20° suggested in [21]. By adjusting the preload of the spring in the SAM model, the maximum knee flexion angle can be effectively limited. In addition, the MG follows the clinical gaits during swing phases, thus ensuring stable walking.

3) Limitations of the proposed approach: Firstly, we only recruited one paraplegic patient to verify the proposed approach, the feasibility and efficiency of the proposed KTM should be evaluated through the participation of numerous paraplegic subjects. Secondly, more precise human-exoskeleton models and SAM parameters based on individual differences should be taken into account. Thus, there are still some works that need to be addressed in the future.

IV. CONCLUSIONS AND FUTURE WORKS

In this paper, we proposed an approach to reduce the PGRF by KTM for walking assistance lower limb exoskeletons. In the KTM, a spring-damping system based SAM is proposed, which introduced the humanoid leg stiffness. When the impact works on the human-exoskeleton system, the SAM can effectively absorb the shock. Combined with the motion of the clinical gait and the SAM, the TM can construct the modulated gaits. The proposed KTM involves active trajectory modulation to imitate the humanoid leg compliance mechanism and without complex and bulky structural design. Finally, the proposed approach has been tested on a lower limb walking assistance exoskeleton robot named AIDER. Experimental results indicate that the proposed approach endows the exoskeleton with the ability to walk with reduced PGRF.

In the future, we would like to explore more effective methods for SAM parameter optimization. As well as extending the proposed gait planning framework to be more robust. Apply it to different scenes, such as downstairs, hills, etc.

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REFERENCES

[1] T. A. Astorino and E. T. Harness, “Effect of chronic activity-based therapy on bone mineral density and bone turnover in persons with spinal cord injury,” Arthritis Rheumatol, vol. 113, no. 12, pp. 3027–3037, 2013.
[2] C. Tefertiller, K. Hays, J. Jones, A. Jayaraman, C. Hartigan, T. Bushnik, and G. Forrest, “Initial outcomes from a multicenter study utilizing the indego powered exoskeleton in spinal cord injury,” Topics in Spinal Cord Injury Rehabilitation, vol. 24, no. 1, pp. 78–85, 2018.
[3] D. B. Fineberg, P. Asselin, N. Y. Harel, I. Agranova-Breyer, S. D. Kornfeld, W. A. Bauman, and A. M. Spungen, “Vertical ground reaction force-based analysis of powered exoskeleton-assisted walking in persons with motor-complete paraplegia,” The Journal of Spinal Cord Medicine, vol. 36, no. 4, pp. 313–321, 2013.
[4] K. E. Zelik and A. D. Kuo, “Human walking isn’t all hard work: evidence of soft tissue contributions to energy dissipation and return,” Journal of Experimental Biology, vol. 213, no. Pt 24, pp. 4257–64, 2010.

[5] J. M. Donelan, R. Kram, and A. D. Kuo, “Mechanical work for step-to-step transitions is a major determinant of metabolic cost of human walking,” Journal of Experimental Biology, vol. 205, no. Pt 23, pp. 3717–3727, 2003.

[6] Y. Iida, H. Kanehisa, Y. Inaba, and K. Nakazawa, “Activity modulations of trunk and lower limb muscles during impact-absorbing landing,” Journal of Electromyography & Kinesiology, vol. 21, no. 4, pp. 602–609, 2011.

[7] C. S. Simonsen, E. G. Celius, C. Brunborg, C. Tallaksen, and S. M. Moen, “Bone mineral density in patients with multiple sclerosis, hereditary ataxia or hereditary spastic paraplegia after at least 10 years of disease - a case control study,” Bmc Neurology, vol. 16, no. 1, p. 252, 2016.

[8] K. E. Zelik, T. W. P. Huang, P. G. Adamczyk, and A. D. Kuo, “The role of series ankle elasticity in bipedal walking,” Journal of Theoretical Biology, vol. 346, pp. 75–85, 2014.

[9] T.-w., P., Huang, K., A., Shorter, P., G., Adamczyk, and A. and, “Mechanical and energetic consequences of reduced ankle plantar-flexion in human walking,” Journal of Experimental Biology, vol. 218, no. 22, pp. 3541–3550, 2021.

[10] P. Malcolm, R. E. Quesada, J. M. Caputo, and S. H. Collins, “The influence of push-off timing in a robotic ankle-foot prosthesis on the energetics and mechanics of walking,” Journal of neuroengineering and rehabilitation, vol. 12, no. 1, pp. 1–15, 2015.

[11] K. Koyama, J. Umezawa, T. Kurihara, H. Naito, and T. Yanagiya, “The influence of position and area of shock absorbing material of shoes on ground reaction force during walking,” Ifmbe Proceedings, vol. 31, pp. 262–265, 2010.

[12] J. Ueda, M. Turkseven, E. Kim, Q. Lowery, and M. Mayo, “Shock absorbing exoskeleton for vertical mobility system: Concept and feasibility study,” in 2018 IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS), 2018, p. 3342–3349.

[13] J. Park, D. Lee, k. Park, and k. Kong, “Reduction of ground impact of a powered exoskeleton by shock absorption mechanism on the shank,” in 2021 International Conference on Robotics and Automation (ICRA), 2021, pp. 2085–2090.

[14] H. Yue and K. Mombaur, “Analysis of human leg joints compliance in different walking scenarios with an optimal control approach,” Ifac International Workshop on Periodic Control Systems, vol. 49, no. 14, pp. 99–106, 2016.

[15] A. Seyfarth, S. Lipfert, J. Rummel, M. Maus, and D. Maykranz, Walking and Running: How Leg Compliance Shapes the Way We Move. Modeling, Simulation and Optimization of Bipedal Walking, 2013.

[16] F. Russell, Y. Takeda, P. Kormushev, R. Vaidyanathan, and P. Ellison, “Stiffness modulation in a humanoid robotic leg and knee,” IEEE Robotics and Automation Letters, vol. 6, no. 2, pp. 2563–2570, 2021.

[17] N. W. Liu, “The effect of muscle stiffness and damping on simulated impact force peaks during running,” Journal of Biomechanics, vol. 32, no. 8, pp. 849–856, 1999.

[18] W. Fu, Y. Liu, and G. Zhao, “Surface effects on plantar pressure characteristics in jogging,” in 2011 International Conference on Future Computer Science and Education, 2011, pp. 93–96.

[19] J. Cardoso, C. Baptista, C. D. Sartor, A. Elias, and A. C. Mattiello-Sverzut, “Dynamic plantar pressure patterns in children and adolescents with charcot-marie-tooth disease,” Gait & Posture, vol. 86, pp. 112–119, 2021.

[20] P. G. Adamczyk and A. D. Kuo, “Redirection of center-of-mass velocity during the step-to-step transition of human walking.” Journal of Experimental Biology, vol. 212, no. Pt 16, pp. 2668–2678, 2009.

[21] C. Araab, C. Ab, D. Jr, and E. Fcb, “Leg and lower limb dynamic joint stiffness during different walking speeds in healthy adults - sciencedirect,” Gait & Posture, vol. 82, pp. 294–300, 2020.

[22] H. Hong, S. Kim, C. Kim, S. Lee, and S. Park, “Spring-like gait mechanics observed during walking in both young and older adults,” Journal of Biomechanics, vol. 46, no. 1, pp. 77–82, 2013.